Evaluation of Pediatric ATD Biofidelity as Compared to Child Volunteers in Low-Speed Far-Side Oblique and Lateral Impacts

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Objective: Motor vehicle crashes are a leading cause of injury and mortality for children. Mitigation of these injuries requires biofidelic anthropomorphic test devices (ATDs) to design and evaluate automotive safety systems. Effective countermeasures exist for frontal and near-side impacts but are limited for far-side impacts. Consequently, far-side impacts represent increased injury and mortality rates compared to frontal impacts. Thus, the objective of this study was to evaluate the biofidelity of the Hybrid III and Q-series pediatric ATDs in low-speed far-side impacts, with and without shoulder belt pretightening.

Methods: Low-speed (2 g) far-side oblique (60°) and lateral (90°) sled tests were conducted using the Hybrid III and Q-series 6- and 10-year-old ATDs. ATDs were restrained by a lap and shoulder belt equipped with a precrash belt pretightener. Photoreflective targets were attached to the head, spine, shoulders, and sternum. ATDs were exposed to 8 low-speed sled tests: 2 oblique nontightened, 2 oblique pretightened, 2 lateral nontightened, 2 lateral pretightened. ATDs were compared with previously collected 9- to 11-year-old (n = 10) volunteer data and newly collected 6- to 8-year-old volunteer data (n = 7) tested with similar methods. Kinematic data were collected from a 3D target tracking system. Metrics of comparison included excursion, seat belt and seat pan reaction loads, belt-to-torso angle, and shoulder belt slip-out.

Results: The ATDs exhibited increased lateral excursion of the head top, C4, and T1 as well as increased downward excursion of the head top compared to the volunteers. Volunteers exhibited greater forward excursion than the ATDs in oblique nontightened impacts. These kinematics correspond to increased shoulder belt slip-out for the ATDs in oblique tests (ATDs = 90%; volunteers = 36%). Contrarily, similar shoulder belt slip-out was observed between ATDs and volunteers in lateral impacts (ATDs = 80%; volunteers = 78%). In pretightened impacts, the ATDs exhibited reduced lateral excursion and torso roll-out angle compared to the volunteers.

Conclusions: In general, the ATDs overestimated lateral excursion in both impact directions, while underestimating forward excursion of the head and neck in oblique impacts compared to the pediatric volunteers. This was primarily due to pendulum-like lateral bending of the entire ATD torso compared to translation of the thorax relative to the abdomen prior to the lateral bending of the upper torso in the volunteers, likely due to the multisegmented spinal column in the volunteers. Additionally, the effect of belt pretightening on occupant kinematics was greater for the ATDs than the volunteers.

Keywords: ATD, pediatrics, biofidelity, kinematics, far-side impacts, human volunteers

Introduction

Motor vehicle crashes (MVCs) are the leading cause of fatal injury (Centers for Disease Control and Prevention 2010) and sixth leading cause of nonfatal injury (Centers for Disease Control and Prevention 2011) for children ages one to 18 years in the United States. Pediatric injury prevention in MVCs is enhanced through the use of biofidelic anthropomorphic test devices (ATDs) to ensure safety systems mitigate injuries in real children.
Traditionally, adult ATDs have been validated using postmortem human subjects (PMHS). However, due to specimen supply limitations surrounding pediatric PMHS impact testing, the pediatric ATD response corridors have typically been scaled from adult biomechanical data (Irwin and Mertz 1997), with limited consideration for the tissue and morphological changes that occur with age. Several studies have called into question the biofidelity of these ATDs (Kent et al. 2009; Lopez-Valdes et al. 2009; Menon et al. 2004, 2005; Sherwood et al. 2002). In the absence of PMHS tests, human volunteer testing becomes a viable option to evaluate pediatric ATD performance and has previously been used for adult ATD evaluation (Beeman et al. 2012; Beegeman et al. 1980).

Previously, we quantified the dynamic responses of pediatric volunteers (6–14 years) in low-speed (<4 g) frontal crash conditions (Arbogast et al. 2009; Seacrist, Arbogast, et al. 2012) and compared the responses of pediatric volunteers to the Hybrid III and Q-Series 6- and 10-year-old ATDs, noting differences in head and spine kinematics, seating environment reaction loads, and upper neck loads (Seacrist et al. 2010, 2013; Seacrist, Mathews, et al. 2012). In general, the ATDs exhibited reduced excursions, increased belt loading, and increased upper neck bending moment, resulting in an overestimation of the neck injury criteria.

Though these data provide valuable insight into the biofidelity of the pediatric ATDs in full frontal impacts, it is important to note that MVCs are not limited to only the frontal crash direction. Side-impact crashes represent a high mortality and injury burden for rear seat occupants with similar number of child fatalities compared to frontal impacts and rollovers (NHHTSA 2008) as well as an increased fatality rate compared to frontal crashes (Arbogast and Durbin 2013). Effective countermeasures for mitigating such injuries in near-side impacts include shoulder and side curtain airbags. However, limited countermeasures exist for far-side impacts. Consequently, far-side impacts account for 43% of Abbreviated Injury Scale (AIS) 3+ injuries and 25% of lateral impact fatalities (Gabler et al. 2005). Developing novel countermeasures for far-side impacts requires the use of biofidelic ATDs. Though the Hybrid III ATDs are traditionally considered frontal impact ATDs, they have been used to evaluate side impact response (Huot et al. 2005; Kapoor et al. 2008; Malott et al. 2004) due to the absence of a side impact Hybrid III child ATD. Although the Q-series ATDs were designed to be omnidirectional, and side-impact-specific versions of the Q-series 6-year-old (Q6s) and 3-year-old (Q3s) are available, there have been limited efforts to establish their side-impact performance.

Given that the side-impact response of these ATDs are based on similar scaling methods and components as frontal ATD development, the differences observed in the frontal loading environment may persist in other crash directions. Consequently, there is a growing need to evaluate the biofidelity of the pediatric ATDs in far-side impacts by comparing them to biomechanical response data from real children, which is extremely limited in the literature. Therefore, the goal of the current study was to evaluate the biofidelity of the Hybrid III and Q-series 6- and 10-year-old pediatric ATDs compared to pediatric volunteers in low-speed far-side impacts. Since our previous side-impact volunteer data lacked appropriately aged children for comparison to the 6-year-old ATDs (Arbogast et al. 2012), a secondary goal of this study was to expand the range of pediatric volunteer data by conducting additional low-speed far-side sled tests with 6- to 8-year-old children.

Despite a similar biofidelic basis, the Hybrid III and Q-series ATDs have different mechanical designs. Prominent anthropometric differences include the Q-series having shorter stature, wider shoulders, decreased torso weight, and increased weight of the lower limbs (European Enhanced Vehicle-Safety Committee Q-dummies Report 2008; Waagmeester et al. 2009). The Hybrid III and Q-series ATDs also exhibit differing rubber neck column and cable construction. Additionally, the Hybrid III ribcage consists of 6 horizontal ribs, whereas the Q-series ribcage is constructed from a single inclined piece. The Q-series ribcage is also more cylindrical than the Hybrid III. Furthermore, the Q-series has a more rounded abdomen and exhibits a more slouched posture (Lubbe 2010). These differences have culminated in differing impact response, including belt interaction, chest deflection, and increased neck flexion (Lubbe 2010; Maltese et al. 2008; Schnottale et al. 2011). Modifications have also been made to Q6s relative to the Q6 aimed at improving side impact biofidelity. These include a segmented rubber neck with center cable to limit maximum rotation and provide torsional resistance, an updated thorax (shape, contour, thickness) to improve lateral compliance, and a modified hip joint allowing for some inward deflection (Wang 2009). Such differences in mechanical design and impact response warrant investigating the biofidelity of both the Hybrid III and Q-series ATDs.

In addition to biofidelic evaluation involving passive 3-point belt restraints, it is important to understand whether the pediatric ATDs accurately predict occupant response when exposed to active safety systems, such as pretensioners and novel precrash technologies. Pretensioners are an effective countermeasure designed to remove the slack from the seat belt and facilitate early application of the restraint load to the occupant by coupling the occupant to the vehicle early in the crash and increasing the amount of work done by the restraint (Bose et al. 2010; Forman et al. 2008; Grime 1979; Kent et al. 2007; Mitzkus and Eyrainer 1984; Viano 1988; Walz 2004). Previous adult PMHS (Kent et al. 2001) and ATD (Forman et al. 2008) laboratory tests, field data (Foret-Bruno et al. 1998), and New Car Assessment Program crash tests (Walz 2004) have suggested that pretensioning is beneficial for reducing injury to adult occupants in frontal impacts. In far-side impacts, both computational models and vehicle-to-vehicle crash tests revealed that pretensioning significantly reduced lateral excursion compared to a standard 3-point belt (Stolinski, Grzebieta, and Fildes 1998a; Stolinski et al. 1999; Stolinski, Grzebieta, and Fildes 1998b). Additionally, a limited number of quasistatic adult volunteer tests revealed that torso roll-out and shoulder belt slip-out were reduced with pretensioning (Douglas et al. 2007). However, data quantifying the effect of these countermeasures on pediatric occupants is limited in the literature. Given that children ages 6 years and older are also restrained using these vehicle-based safety
systems, there is a need to evaluate the effect of pretensioning on pediatric occupant kinematics.

Similar to pretensioners, novel seat belt pretighteners also provide retraction of the belt to limit occupant motion. However, unlike pretensioners that activate at the onset of impact, pretighteners are precrash devices that respond earlier to potential impacts, providing significantly lower loads in an effort to (1) place occupants in an optimal initial position should an impact occur and (2) provide low-load belt retraction for less severe impacts that do not warrant activation of pretensioners. These novel belt pretighteners provide an opportunity to evaluate the effect of “pretensioning” on pediatric volunteers in the context of low-speed impacts. A recent study evaluated the effect of belt pretightening on pediatric volunteers during low-speed oblique and lateral sled tests (Arbogast et al. 2012), revealing significant reductions in torso roll-out, as well as forward and lateral excursion in pretightened compared to nontightened trials. These data confirm previous high-speed modeling efforts (Douglas et al. 2007), and these data can be used to evaluate pediatric ATD biofidelity in pretensioned impacts. Therefore, a secondary goal of this study was to evaluate ATD biofidelity in response to belt pretightening by comparing their response to pediatric volunteers during pretightened low-speed far-side lateral and oblique sled tests.

Methods

This study protocol was reviewed and approved by the Institutional Review Boards at The Children’s Hospital of Philadelphia, Philadelphia, Pennsylvania, and Rowan University, Glassboro, New Jersey (the testing site).

Human Volunteer Data

A comprehensive description of the human volunteer testing can be found in Arbogast et al. (2012). Briefly, low-speed, noninjurious far-side oblique (60°) and lateral (90°) sled tests were conducted on healthy male pediatric volunteers between the ages of 9 and 14 years (n = 20). Subjects were seated in a pneumatically actuated, hydraulically controlled low-speed deceleration sled equipped with an onboard accelerometer. A safe, noninjurious crash pulse applicable to the pediatric population was derived from an amusement park bumper car impact. The peak sled acceleration and rise time were 1.88 g in 52.7 ms for oblique impacts and 1.91 g in 54.3 ms for lateral impacts. Subjects were restrained using a standard automotive 3-point belt system (Takata Corp., Tokyo, Japan) equipped with a pretightener that applied a 250 N pretightening force to the occupant. The belt geometry was such that the D-ring is located 296 mm rearward of the H-point, falling within the measured range of 2001–2008 U.S.-based vehicles (Reed et al. 2008). Lightweight belt webbing load cells (model 6200FL-41-30, Denton ATD Inc., Rochester Hills, MI) were attached to the shoulder belt and on the right and left locations on the lap belt. The initial position of the torso and knee angles was set to 110°. The height of the shoulder belt anchor was adjusted to provide similar fit across subjects; specifically, the shoulder belt angle at the D-ring (defined as the angle the shoulder belt makes with the horizontal) was set at 55°, as measured by an inclinometer, at initial position for each subject by raising or lowering the D-ring. The lap belt anchor locations were fixed at 55° throughout the test series. Photoreflective markers were placed on anatomical landmarks of interest including the head top, C4, T1, external auditory meatus (bilaterally), suprasternal notch, and xiphoid process and were tracked using an 8-camera 3D near-infrared video target tracking system (model Eagle 4, Motion Analysis Corporation, Santa Rosa, CA), which triangulated the position of each marker in 3D space based on the 2D images of each camera and the intensity of the marker reflection. The accuracy of this system was verified by a static and dynamic calibration procedure that resolved a 500-mm calibration distance to 0.1 mm. A high-speed video camera (MotionXtra HGTH, Redlake, San Diego, CA) focused on the coronal plane of the occupant recorded the qualitative relative movement of the torso and the shoulder belt. Volunteers were assigned to either the oblique or lateral test conditions. Two repetitive trials were conducted for both the nontightened and pretightened conditions such that each volunteer experienced 4 trials. A test matrix is shown in Table 1. Trials were randomized; no effect of trial order was observed (Arbogast et al. 2012).

Experimental Testing

Low-speed oblique and lateral sled tests were conducted using the Hybrid III 6, Hybrid III 10, Q6, Q6s, and Q10. For comparison to the 6-year-old ATDs, additional healthy male 6- to 8-year-old volunteers (n = 7) were recruited. Informed consent was obtained from a parent/guardian and informed assent was obtained from the child. All volunteer and ATD tests were conducted using the aforementioned test sled and crash pulses. Marker placement, initial position, and test methodology mimicked that used for the pediatric subjects (Figure 1). Shoulder belt and lap belt anchor angles were set to 55°; the

Table 1. Test matrix

| Test condition | Impact angle |
|----------------|--------------|
|                | Oblique (60°) | Lateral (90°) |
| Nontightened   | 2 Trials     | 2 Trials      |
| Pretightened   | 2 Trials     | 2 Trials      |

Fig. 1. Instrumentation and initial position comparison of a pediatric volunteer (left), Hybrid III 10 (middle), and Q10 (right).
shoulder belt was allowed to lie naturally along the thorax. Two repeated trials were conducted per test condition.

Volunteer testing of the 6- to 8-year-olds mirrored that described in Arbogast et al. (2012) with one adaptation. Pilot testing revealed that the 6- to 8-year-old volunteers were uncomfortable removing their shirt during testing. To improve subject comfort and retention, 6- to 8-year-old volunteers donned a sleeveless, form-fitting compression athletic shirt. The only markers affected by donning of the shirt were the suprasternal notch and xiphoid process; all other markers were placed directly on the skin. To evaluate the effect of the athletic shirt on occupant kinematics and belt interaction, 2 sled tests with and without the compression shirt were conducted using the Hybrid III 10-year-old ATD.

Data Acquisition and Processing

Accelerometer and load cells were sampled at 10,000 Hz using a T-DAS data acquisition system (Diversified Technical Systems Inc., Seal Beach, CA) with a built-in anti-aliasing filter (4,300 Hz). The sled acceleration, seat belt loads, and forces and moments at the seat pan and foot rest were filtered at SAE channel frequency class 60, according to SAE J211 (Society of Automotive Engineers 1995). Motion analysis data were acquired at 100 Hz, which provided the optimal sampling frequency and resolution for this test series, and analyzed using Motion Analysis Cortex 2.5 software (Motion Analysis Corporation, Santa Rosa, CA). High-speed video was collected at 1000 Hz.

Data Reduction

These pediatric volunteer data were compared to the Hybrid III 6, Hybrid III 10, Q6, Q6s, and Q10 ATDs. The ATDs were matched to the pediatric volunteers based on age; the 6- to 8-year-olds and 9- to 11-year-olds were matched to the 6-year-old and 10-year-old ATDs, respectively. To account for variations in stature and mass within the age groups, trajectories and seat environment reaction loads of the volunteers and Q-series ATDs were scaled to the anthropometry of the corresponding Hybrid III ATD according to length scaling (Eq. (1); Langhaar 1951) and force scaling (Eq. (2); Eppinger et al. 1984), respectively:

$$\lambda_L = \frac{L_{\text{Hybrid III}}}{L_{\text{subject}}}$$

$$\lambda_F = \left( \frac{m_{\text{Hybrid III}}}{m_{\text{subject}}} \right)^{2/3}$$

where $\lambda_L$ is the ratio of seated height, $\lambda_F$ is the force scaling factor, and $\lambda_m$ is the ratio of body mass. Mean ($\pm$SE) age, mass, erect seated height, and scale factors for the volunteers and ATDs are listed in Table 2.

The time series motion analysis and T-DAS data were imported into MATLAB (Mathworks, Inc., Natick, MA) for data analysis using a custom algorithm. The right rear seat pan marker was designated as the origin for the local (sled) coordinate system. Peak excursion relative to event onset was computed for the head and neck markers. Pediatric volunteer trajectory corridors were developed based on the mean response and one standard deviation in the $x$, $y$, and $z$ directions. Coronal plane head rotation was computed as the change in angle of the vector created by the bilateral external auditory meatus markers, relative to horizontal. Lateral bending was computed as the change in angle of the vector created by the suprasternal notch and xiphoid process markers relative to vertical. Torso roll-out angle was defined as the projected angle in the coronal plane between the sternum and the shoulder belt (Arbogast et al. 2012). Shoulder belt slip-out was determined from coronal plane high-speed video data and defined as the width of the shoulder belt moving beyond the ipsilateral acromion. Peak shoulder belt load and lateral seat pan shear loads were also computed.

Statistical Analysis

Data were imported into Statistical Analysis System 9.3 (SAS Institute Inc, Cary, NC) for statistical analysis and analyzed using both descriptive and inferential statistical techniques. The experiment-wise error rate was held at the .05 level. A robust variance estimator accounted for clustering of trials due to the multiple trials of each pediatric volunteer. Specifically, the Taylor series linearization method estimated within- and between-subject variance by linking multiple trials with each subject. Unlike general estimating equations and linear mixed effect methods, this method requires no assumption about the correlation structure of the outcomes and uses less computation power. Adjusted 95% confidence intervals (CIs) were estimated for all parameters. Differences between the ATDs and the pediatric volunteers were assessed by comparing the ATD mean of the repeated trials to the corresponding pediatric volunteer 95% CIs.

Results

Peak ATD response and pediatric volunteer 95% CIs for the 6-year-old comparison in oblique and lateral impacts are listed in Tables 3 and 4, respectively. Peak ATD response and pediatric volunteer 95% CIs for the 10-year-old comparison in oblique and lateral impacts are listed in Tables 5 and 6, respectively. Shoulder belt slip-out is listed in Table A1 (see online supplement). Head and neck coronal ($y-z$) and transverse plane ($x-y$) trajectories for the ATDs and trajectory corridors for the volunteers are shown in the Appendix.

Both the 6-year-old and 10-year-old ATDs overestimated lateral ($\Delta Y$) excursion and downward ($\Delta Z$) excursion of the head compared to the corresponding volunteer cohorts in non-tightened oblique and lateral impacts, with the exception of the Q10 in lateral impacts. In contrast, the ATDs underestimated or exhibited similar forward ($\Delta X$) excursion compared to the volunteers in oblique impacts. Contrary to the volunteers who exhibited upward ($\Delta Z$) excursion of the C4 and T1, the ATDs generally exhibited downward ($\Delta Z$) excursions of these landmarks. The opposite trends were observed in pretightened trials, where the ATDs exhibited reduced excursion for nearly all landmarks as compared to the volunteers.
The ATDs overestimated head rotation compared to the pediatric volunteers in oblique and lateral nontightened impacts. In pretightened impacts, the ATDs exhibited similar or reduced head rotation compared to the volunteers.

The Hybrid III 6 and Q6 exhibited increased torso roll-out compared to the volunteers in nontightened oblique and lateral impacts. Contrarily, the Hybrid III 10 and Q10 exhibited similar torso roll-out in oblique impacts and reduced torso roll-out in lateral impacts compared to the volunteers. The Q6s exhibited similar torso roll-out to the volunteers in oblique and lateral nontightened impacts. The addition of pretightening resulted in the ATDs exhibiting decreased or similar torso roll-out to the volunteers.

In general, the ATDs overestimated lateral seat pan shear and underestimated or had similar belt loads compared to the volunteers in nontightened trials. In pretightened trials, the ATDs overestimated both lateral seat pan shear and shoulder belt loads with a few exceptions.

Increased ATD shoulder belt slip-out was observed in nontightened oblique impacts, with slip-out present in 90% of ATD trials compared to only 36% of pediatric volunteer trials. In contrast, similar frequency of shoulder belt slip-out was observed in nontightened lateral impacts (ATDs = 80%; volunteer = 78%). Pretightening had a substantial effect on shoulder belt slip-out, reducing ATD slip-out to 0% in both oblique and lateral directions and reducing volunteer slip-out to 8% and 6% in oblique and lateral impacts, respectively.

Comparison of the shirt vs. shirtless tests using the Hybrid III 10 indicated a marginal increase in torso roll-out angle of $3^\circ$ (shirt = 28$^\circ$; shirtless = 31$^\circ$) in oblique impacts and 4$^\circ$ (shirt = 29$^\circ$; shirtless = 33$^\circ$) in lateral impacts in the absence of a shirt. No differences in belt slip-out were observed between the shirt and shirtless trials; the Hybrid III 10 exhibited belt slip-out in nontightened trials and no belt slip-out occurred in pretightened trials with the shirt.

**Discussion**

This study sought to evaluate the biofidelity of the Hybrid III and Q-series 6- and 10-year-old ATDs in low-speed far-side impacts by comparing their kinematic response to age-matched pediatric volunteers. In general, the ATDs overestimated lateral and downward motion of the head and neck. This is contrary to our previous findings in frontal impacts.

**Table 2. Volunteer and ATD anthropometry and scale factors**

| Group       | Impact angle ($^\circ$) | Age (years) | Mass (kg) | 6-Year-old cohort (cm) | $\lambda_L$ | $\lambda_F$ |
|-------------|-------------------------|-------------|-----------|------------------------|------------|------------|
| Hybrid III  | 60/90                   | 6           | 23.4      | 63.5                   | 1.00       | 1.00       |
| Q/Qs 60/90 | 60/90                   | 6           | 22.9      | 60.1                   | 1.06       | 1.01       |
| Volunteers (n = 3) | 60                  | 7.4 ± 0.7   | 25.3 ± 2.5 | 66.4 ± 1.1          | 0.96 ± 0.02 | 0.96 ± 0.07 |
| Volunteers (n = 4) | 90                  | 7.8 ± 0.4   | 32.0 ± 1.8 | 69.7 ± 1.4          | 0.91 ± 0.02 | 0.81 ± 0.03 |
| Hybrid III  | 60/90                   | 10          | 35.3      | 72.4                   | 1.00       | 1.00       |
| Q           | 60/90                   | 10.5        | 38.5      | 77.8                   | 0.93       | 0.94       |
| Volunteers (n = 5) | 60               | 10.7 ± 0.4  | 37.0 ± 4.1 | 74.7 ± 1.8          | 0.97 ± 0.02 | 1.00 ± 0.07 |
| Volunteers (n = 5) | 90               | 10.8 ± 0.5  | 34.8 ± 0.9 | 74.8 ± 1.5          | 0.97 ± 0.02 | 1.01 ± 0.02 |

**Table 3. Comparison of peak responses: 6-year-old ATDs (oblique)**

|                  | Nontightened | Pretightened |
|------------------|--------------|--------------|
|                  | III 6 | Q6  | Q6s | 6-8 Years (95% CI) | III 6 | Q6  | Q6s | 6-8 Years (95% CI) |
| Head top $\Delta X$ (mm) | 107$^*$ | 63$^*$ | 76$^*$ | 120–132 | 36$^*$ | 33$^*$ | 15$^*$ | 58–96 |
| Head top $\Delta Y$ (mm)  | 350$^*$ | 321 | 348$^*$ | 243–323 | 107$^*$ | 160$^*$ | 154$^*$ | 204–238 |
| Head top $\Delta Z$ (mm)  | $-$143$^*$ | $-$72$^*$ | $-$76$^*$ | $-$23 to $-$52 | $-$26$^*$ | $-$24$^*$ | $-$24$^*$ | $-$39 to $-$51 |
| C4 $\Delta X$ (mm)       | 47$^*$ | 36$^*$ | 44$^*$ | 68–74 | 2$^*$ | 10$^*$ | 4$^*$ | 15–24 |
| C4 $\Delta Y$ (mm)       | 220 | 214 | 212 | 182–220 | 69$^*$ | 83$^*$ | 82$^*$ | 126–139 |
| C4 $\Delta Z$ (mm)       | $-$74$^*$ | $-$49$^*$ | $-$27$^*$ | 28–31 | $-$8$^*$ | $-$16$^*$ | $-$7$^*$ | $-$6 to 26 |
| T1 $\Delta X$ (mm)       | 31$^*$ | 24$^*$ | 38$^*$ | 56–59 | 1$^*$ | 3$^*$ | 4$^*$ | 7–11 |
| T1 $\Delta Y$ (mm)       | 197 | 214$^*$ | 189 | 164–202 | 63$^*$ | 66$^*$ | 77$^*$ | 106–115 |
| T1 $\Delta Z$ (mm)       | $-$53$^*$ | $-$4$^*$ | $-$23$^*$ | 30–39 | $-$11$^*$ | $-$14$^*$ | $-$9$^*$ | 0–25 |
| Head rotation ($^\circ$)  | 38$^*$ | 31$^*$ | 39$^*$ | 17–30 | 11$^*$ | 20 | 20 | 20–28 |
| Lateral bending ($^\circ$) | 45$^*$ | 37 | 36 | 19–40 | 13$^*$ | 5$^*$ | 12$^*$ | 16–28 |
| Torso roll-out ($^\circ$) | 51$^*$ | 45$^*$ | 41 | 38–41 | 16$^*$ | 11$^*$ | 17$^*$ | 21–36 |
| Shoulder belt load (N)    | 146 | 217$^*$ | 161 | 95–198 | 413$^*$ | 343 | 406$^*$ | 326–384 |
| Lap belt—left (N)         | 106$^*$ | 150 | 125$^*$ | 146–207 | 87 | 90 | 153$^*$ | 59–98 |
| Lap belt—right (N)        | 81$^*$ | 108 | 135 | 95–145 | 76 | 69 | 137$^*$ | 65–109 |
| Seat pan shear (N)        | $-$243$^*$ | $-$211$^*$ | $-$195$^*$ | $-$140 to $-$155 | $-$270$^*$ | $-$202 | $-$228$^*$ | $-$145 to $-$218 |

*Significant differences between the ATD and human volunteers at $P < .05$. 

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**Table 4. Comparison of peak responses: 6-year-old ATDs (lateral)**

|                      | Nontightened |                  |                  | Pretightened |                  |                  |
|----------------------|--------------|------------------|------------------|--------------|------------------|------------------|
|                      | HIII 6       | Q6               | Q6s              | HIII 6       | Q6               | Q6s              |
| Head top ΔX (mm)     | 10           | 12               | 9                | 3            | 6                | 2                |
| Head top ΔY (mm)     | 354*         | 365*             | 328*             | 273–308      | 138*            | 146*             | 183*             | 187–239        |
| Head top ΔZ (mm)     | −116*        | −93*             | −78*             | −27 to −55   | −19*            | −17*             | −33*             | −35 to −68     |
| C4 ΔX (mm)           | 2*           | 6                | 7                | 0*           | 2                | 2                |
| C4 ΔY (mm)           | 187          | 235*             | 193              | 187–209      | 75*             | 72*              | 90*              | 102–127        |
| C4 ΔZ (mm)           | −22*         | −46*             | −21*             | −16 to 14    | −4              | −7               | −9               | −21 to 3       |
| T1 ΔX (mm)           | 8            | 6                | 7                | 2            | 11               | 2                | 1                |
| T1 ΔY (mm)           | 171          | 195*             | 172              | 156–177      | 66*             | 60*              | 82               | 74–90          |
| T1 ΔZ (mm)           | −24*         | −35*             | −19*             | 0–20         | −3*             | −10*             | −13*             | −1 to 15       |
| Head rotation (°)    | 46*          | 42*              | 41*              | 22–29        | 15*             | 4°               | 13               | 13–32          |
| Torso roll-out (°)   | 45*          | 44*              | 35               | 30–41        | 18*             | 9°               | 17               | 16–35          |
| Shoulder belt load (N) | 202*       | 166              | 131*             | 163–184      | 378*            | 384*             | 422*             | 305–349        |
| Lap belt—left (N)    | 152*         | 205              | 143*             | 197–224      | 123             | 133*             | 218*             | 116–132        |
| Lap belt—right (N)   | 79*          | 154              | 131              | 124–177      | 75*             | 114              | 194*             | 96–134         |
| Seat pan shear (N)   | −280*        | −255*            | −249*            | −164 to −177 | −331*           | −281*            | −280*            | −220 to −229   |

*Significant differences between the ATD and human volunteers at $P < .05$.

**Table 5. Comparison of peak responses: 10-year-old ATDs (oblique)**

|                      | Nontightened |                  |                  | Pretightened |                  |                  |
|----------------------|--------------|------------------|------------------|--------------|------------------|------------------|
|                      | HIII 10      | Q10              | 9–11 Years (95% CI) | HIII 10      | Q10              | 9–11 Years (95% CI) |
| Head top ΔX (mm)     | 65*          | 21*              | 67–135           | 51*          | 3*               | 59–67            |
| Head top ΔY (mm)     | 294*         | 305*             | 219–246          | 136*         | 120*             | 146–174          |
| Head top ΔZ (mm)     | −57*         | −80*             | −12 to −36       | −17          | −17              | −9 to −34        |
| C4 ΔX (mm)           | 30           | 15*              | 26–70            | 9*           | 2*               | 21–27            |
| C4 ΔY (mm)           | 187          | 193              | 146–198          | 66*          | 59*              | 76–91            |
| C4 ΔZ (mm)           | −18*         | −36*             | 12–25            | 4*           | −6*              | 9–22             |
| T1 ΔX (mm)           | 30           | 15*              | 26–50            | 7*           | 2*               | 10–17            |
| T1 ΔY (mm)           | 137          | 173              | 132–175          | 47*          | 49*              | 49–70            |
| T1 ΔZ (mm)           | 5*           | −33*             | 11–22            | −4*          | −4*              | 8–20             |
| Head rotation (°)    | 34*          | 35*              | 10–16            | 19           | 17               | 17–20            |
| Lateral bending (°)  | 26*          | 28*              | 17–22            | 11           | 3                | 1–15             |
| Torso roll-out (°)   | 28           | 31               | 28–36            | 13           | 8                | 7–21             |
| Shoulder belt load (N) | 225*       | 242*             | 244–306          | 480*         | 330*             | 416–463          |
| Lap belt—left (N)    | 295*         | 169*             | 212–256          | 171          | 122*             | 131–194          |
| Lap belt—right (N)   | 95*          | 108*             | 119–158          | 67*          | 100*             | 104–128          |
| Seat pan shear (N)   | −305*        | −263*            | −204 to −224     | −295*        | −269*            | −227 to −269</TB> |

*Significant differences between the ATD and human volunteers at $P < .05$.

**Table 6. Comparison of peak responses: 10-year-old ATDs (lateral)**

|                      | Nontightened |                  |                  | Pretightened |                  |                  |
|----------------------|--------------|------------------|------------------|--------------|------------------|------------------|
|                      | HIII 10      | Q10              | 9–11 Years (95% CI) | HIII 10      | Q10              | 9–11 Years (95% CI) |
| Head top ΔX (mm)     | 1*           | 3*               | 15–28            | 10           | 1*               | 7–26             |
| Head top ΔY (mm)     | 341*         | 244*             | 262–322          | 151*         | 140*             | 166–204          |
| Head top ΔZ (mm)     | −91*         | −60*             | −22 to −59       | −16          | −15*             | −16 to −38       |
| C4 ΔX (mm)           | 0*           | 1*               | 7–17             | 3*           | 3*               | 6–11             |
| C4 ΔY (mm)           | 181          | 135*             | 162–234          | 61*          | 69*              | 79–102           |
| C4 ΔZ (mm)           | −26          | −16              | −31 to −14       | 2            | −5               | −7 to −15        |
| T1 ΔX (mm)           | 3*           | 1*               | 7–14             | 4*           | 3*               | 6–15             |
| T1 ΔY (mm)           | 142*         | 116*             | 150–203          | 54           | 58               | 50–79            |
| T1 ΔZ (mm)           | −8*          | −13*             | −2 to 18         | 8            | −5*              | 5–15             |
| Head rotation (°)    | 45*          | 29*              | 16–25            | 23           | 13*              | 17–32            |
| Lateral bending (°)  | 31*          | 18*              | 21–30            | 8            | 6*               | 8–16             |
| Torso roll-out (°)   | 29*          | 20*              | 34–40            | 10*          | 11*              | 13–23            |
| Shoulder belt load (N) | 239*       | 274              | 247–313          | 412          | 378*             | 412–445          |
| Lap belt—left (N)    | 285          | 213*             | 248–302          | 158          | 139              | 139–199          |
| Lap belt—right (N)   | 108*         | 139*             | 174–219          | 89*          | 117*             | 132–191          |
| Seat pan shear (N)   | −354*        | −317*            | −246 to −273     | −397*        | −381*            | −314 to −340     |

*Significant differences between the ATD and human volunteers at $P < .05$. 
Premeditation of only the maximum excursions does not fully capture the global kinematic differences between the ATDs and volunteers. The Hybrid III and Q-series lumbar spine allows for limited relative shear motion, essentially coupling the thorax and pelvis. Consequently, at these low load conditions, the ATD torso behaves like a pendulum, resulting in the torso tipping laterally from the pelvis. Contrarily, the flexible and multisegmented pediatric human spine may allow the thorax to translate and remain more upright before rotation, yielding lower overall excursions. This is evidenced by the flatter portion of the volunteer trajectories that precedes the rotation (Figures A1 and A2, see online supplement), as well as the downward excursions exhibited by the C4 and T1 markers in the ATDs compared to the upward motion of these landmarks in the volunteers (Tables 3–6). Of interest, the WorldSID 5th percentile female ATD, unlike the Hybrid III and Q-series ATDs, does contain a flexible lumbar spine designed to simulate shear motion between the upper and lower torso. Testing of the WorldSID 5th in our low-speed sled is currently underway to determine whether this revision improves biofidelity in side impacts.

Pretightening resulted in substantial reductions of lateral and downward motion for both the volunteers and the ATDs. However, the effect of belt pretightening was greater on the ATDs than the volunteers. The coupling of the upper and lower thorax may again be responsible for the increased effectiveness of pretightening on the ATDs. Given that the pretightener acts on both the lap and shoulder belts, the coupled ATD thorax could experience increased restraint compared to the multisegmented pediatric thorax. These findings have important implications for the design of low-speed countermeasures and precrash devices, because the ATDs could potentially overestimate the effectiveness of belt pretighteners.

These kinematics correspond to increased shoulder belt slip-out for the ATDs compared to the volunteers in oblique, nontightened impacts. This is likely due to the presence of an articulating shoulder in the volunteers, which, coupled with increased forward motion in the oblique impacts, helped retain the shoulder belt on the torso. Similar shoulder belt slip-out was observed between the ATDs and volunteers in lateral impacts where a lack of forward motion likely minimized the influence of shoulder articulation on belt retention. Pretightening virtually eliminated shoulder belt slip-out in the oblique and lateral directions for both the ATDs and volunteers, suggesting that pretightening is an effective method for minimizing shoulder belt slip-out in low load far-side impacts.

Also contrary to our findings in frontal impacts (Seacrist, Mathews, et al. 2012), the ATDs exhibited increased seat pan shear force combined with similar or reduced belt loads compared to the volunteer cohorts. In frontal impacts the increased seat pan shear observed in the volunteers was attributed to the compliance of the human thigh and buttock creating a larger contact area with the seat pan. However, it is possible that the human tissue compliance is less influential along the lateral axis of the thigh in these side impacts. This behavior may also stem from the decoupled human torso. Due to the coupled ATD torso, any lateral translation exhibited by the ATDs would have to occur through pelvic sliding, transferring additional load to the seat pan. However, given the ability of the human torso to translate independently from the pelvis, the volunteers may have exhibited reduced pelvis motion and, consequently, decreased seat pan shear compared to the ATDs. Further study is needed to determine the cause of these differences in seat environment loads.

In oblique impacts, the Hybrid III ATDs exhibited increased forward excursion relative to their Q-series equivalents. Obviously, the Hybrid III series are designed specifically for frontal impacts, whereas the Q-series is intended for omnidirectional testing. The increased focus on frontal response in the Hybrid III series may be the source of this difference. Interestingly, the Q10 exhibited reduced torso roll-out in lateral impacts compared to oblique impacts as well as the other ATDs. Though this may seem counterintuitive, the increase can be attributed to the frontal component in the oblique tests, which allows the Q-series ATD to pivot around the shoulder belt rather than simple lateral motion relative to the belt in the lateral impact tests. Additionally, the more inclined Q-series chest results in the shoulder belt crossing the thorax more cranially, potentially restricting thorax excursion in lateral impacts. Compared to the Q6, the Q6s exhibited reduced torso roll-out and T1 lateral excursion that was similar to the volunteer cohort. This reduced torso roll-out could be due to the improved thorax shape and lateral compliance relative to the Q6 (Wang 2009).

Unlike adult ATDs, which represent a specific size and gender (e.g., 5th percentile female, 50th percentile male, 95th percentile male), the pediatric ATDs are intended to represent a specific age. Therefore, the current study matched volunteers to the pediatric ATDs based on age, using force and length scaling to account for anthropometric differences. Though previous studies have called into question the accuracy of these methods when scaling from adult to pediatric data (Kent et al. 2009; Lopez-Valdes et al. 2009; Menon et al. 2004, 2005; Sherwood et al. 2002), the volunteer cohorts in this study were of similar age and, therefore, of similar anatomic maturation to that which the ATD is designed to represent. Hence, the effect of scaling on occupant response in the current study should not introduce errors associated with age-based differences.

Though this study focused on far-side impacts, the findings do have relevance for near-side impacts because the response will be similar up to the point of contact with the vehicle structure in near-side impacts. The observed differences between the ATD response including decreased out-of-plane motion,
increased head rotation, and torso “tipping” compared to the volunteers may lead to nonbiofidelic interaction with side curtain airbags or the vehicle structure.

Several limitations warrant discussion. First, the current study used a standard 3-point seat belt adjusted to the height of the ATDs and pediatric volunteers. The setup was not designed to mimic any specific automobile but rather to provide an automotive-like environment in which to compare the ATDs and humans. In a vehicle environment, the 6- to 8-year-old volunteers and ATDs would be restrained using a belt-positioning booster. The addition of such child restraints would shift the occupant center of mass further upward relative to the point of impact, potentially yielding increased excursion. Consequently, the current study may slightly underpredict the head excursion in real-world crashes for the 6- to 8-year-old cohort. Additionally, because human tissue is viscoelastic, these findings may not be generalizable to other loading rates, such as those seen in motor vehicle crashes. Because pediatric volunteers cannot be tested at crash-relevant rates, future work should explore novel computational methods to determine whether the observed ATD–volunteer differences in kinematics and reaction loads remain at increased loading rates. Such efforts have been conducted for frontal impacts (Dibb et al. 2014). Lastly, a 6-year-old ATD would ideally have been used to evaluate the influence of the shirt donned by the 6- to 8-year-olds but was unavailable at the time of volunteer testing. Therefore, a Hybrid III 10 was used. Given that the goal was to evaluate the influence of the shirt on belt-to-torso kinematics and that the Hybrid III ATDs use the same vinyl skin, the use of the Hybrid III 10 vs. Hybrid III 6 should not significantly influence our findings.

To our knowledge, this is the first study to evaluate the biofidelity of the 6- and 10-year-old pediatric ATDs in side impacts by comparing their response to child volunteers. In general, the ATDs overestimated lateral excursions and head rotation while underestimating forward excursion when compared to age-matched volunteers. The ATDs also exhibited primarily lateral bending or tipping compared to translation of the thorax relative to the abdomen followed by lateral torso rotation in the volunteers, likely due to the decoupled lumbar spine in the volunteers. These differences led to increased shoulder belt slip-out in the ATDs compared to the volunteers. Belt pretightening had a greater effect on the ATD kinematics compared to the volunteers, which could potentially lead to overestimations of the pretightener’s effectiveness on pediatric occupants in regulatory tests. Overall, the findings suggest that the pediatric ATDs exhibit suboptimal biofidelity under these loading conditions. Additionally, though some metrics were statistically similar between the ATDs, no particular ATD was substantially more biofidelic than the other ATDs. These data provide insight into the biofidelity of the pediatric ATDs in far-side impacts and have important implications for the design and evaluation of low-speed countermeasures, such as novel precrash belt pretighteners.

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Supplemental Materials

Supplemental data for this article can be accessed on the publisher’s website.

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