

# An inflatable belt system in the rear seat occupant environment: investigating feasibility and benefit in frontal impact sled tests with a 50<sup>th</sup> percentile male ATD

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**ABSTRACT** – Frontal-impact airbag systems have the potential to provide a benefit to rear seat occupants by distributing restraining forces over the body in a manner not possible using belts alone. This study sought to investigate the effects of incorporating a belt-integrated airbag ("airbelt") into a rear seat occupant restraint system. Frontal impact sled tests were performed with a Hybrid III 50th percentile male anthropomorphic test device (ATD) seated in the right-rear passenger position of a 2004 mid-sized sedan buck. Tests were performed at 48 km/h (20 g, 100 ms acceleration pulse) and 29 km/h (11 g, 100 ms). The restraints consisted of a 3-point belt system with a cylindrical airbag integrated into the upper portion of the shoulder belt. The airbag was tapered in shape, with a maximum diameter of 16 cm (at the shoulder) that decreased to 4 cm at the mid-chest. A 2.5 kN force-limiter was integrated into the shoulder-belt retractor, and a 2.3 kN pretensioner was present in the out-board anchor of the lap belt. Six ATD tests (three 48 km/h and three 29 km/h) were performed with the airbelt system. These were compared to previous frontal-impact, rear seat ATD tests with a standard (not-force-limited, not-pretensioned) 3-point belt system and a progressive force-limiting (peak 4.4 kN), pretensioning (FL+PT) 3-point belt system. In the 48 km/h tests, the airbelt resulted in significantly less ( $p < 0.05$ , two-tailed Student's t-test) posterior displacement of the sternum towards the spine (chest deflection) than both the standard and FL+PT belt systems (airbelt: average  $13 \pm 1.1$  mm standard deviation; standard belt:  $33 \pm 2.3$  mm; FL+PT belt:  $23 \pm 2.6$  mm). This was consistent with a significant reduction in the peak upper shoulder belt force (airbelt:  $2.7 \pm 0.1$  kN; standard belt:  $8.7 \pm 0.3$  kN; FL+PT belt:  $4.4 \pm 0.1$  kN), and was accompanied by a small increase in forward motion of the head (airbelt:  $54 \pm 0.4$  cm; standard belt:  $45 \pm 1.3$  cm; FL+PT belt:  $47 \pm 1.1$  cm). The airbelt system also significantly reduced the flexion moment in the lower neck (airbelt:  $169 \pm 3.3$  Nm; standard belt:  $655 \pm 26$  Nm; FL+PT belt:  $308 \pm 19$  Nm). Similar results were observed in the 29 km/h tests. These results suggest that this airbelt system may provide some benefit for adult rear seat occupants in frontal collisions, even in relatively low-speed impacts. Further study is needed to evaluate this type of restraint system for different size occupants (e.g., children), for out-of-position occupants, and with other occupant models (e.g., cadavers).

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## INTRODUCTION

The refinement of restraints for rear seat passengers remains a developing field in automobile safety. Rear seat occupants are present in 13-17% of all frontal towaway collisions, and 23% of all fatal frontal collisions (Parenteau and Viano 2003). Unlike in the front seat where an airbag and a knee bolster are present, the seatbelt must provide the restraint necessary to decelerate rear seat occupants during collisions. As a result, the shoulder belt is the most

common source of injury to restrained adult rear seat occupants in frontal collisions; 76% of AIS3+ injuries in restrained rear seat occupants over the age of 13 occur in the thorax (Parenteau and Viano 2003). Thoracic injuries from belt loading are especially problematic for the elderly due to increased thoracic fragility (Kent et al. 2005), increased belt use (Glassbrenner et al. 2004), and increased risk of morbidity and mortality following otherwise minor injuries such as rib fractures (Kent et

al. 2008). Frampton and Lenard (2009) identified the reduction in seatbelt loads in frontal collisions – especially for the elderly (and including rear seat passengers) – as a potential area for reduction in AIS 3+ injuries through the improvement in passive safety system design.

### **Improved Rear Seat Restraints – Previous Studies**

Over the past several years, several studies have begun investigating the efficacy and potential benefit that may be gained by employing advanced passive safety technologies into the restraints of the rear seat. The main goal of such restraints in a frontal collision is to reduce chest loading while still adequately managing the forward motion of the rear seat occupants. In a series of 50 km/h frontal impact sled tests, Zellmer et al. (1998) reported that a pretensioned belt system with a 5.5 kN force limiter could reduce internal chest deflection of a Hybrid III 50<sup>th</sup> percentile male (AM50) anthropomorphic test device (ATD, dummy) compared to a standard belt, while maintaining the 300 mm forward chest motion limit of the ECE-R 16 regulation. In a parametric study using MADYMO simulations with ATD models in frontal impact collisions, Kent et al. (2007) reported that there were many different combinations of pretensioner and force-limiter characteristics that may reduce both chest deflection and head excursion in frontal collisions up to at least 48 km/h.

Forman et al. (2008) expanded on the MADYMO study of Kent et al. (2007) study with a series of 29 km/h and 48 km/h frontal impact, rear seat sled tests with various sizes of ATDs. Tests were performed with either a standard (not force-limited, not pretensioned) 3-point belt system or with a force-limiting 3-point belt with a retractor pretensioner (FL+PT system). To satisfy the static rear seat restraint test requirements of U.S. Federal Motor Vehicle Safety Standard (FMVSS) 209, the force-limiter used in that study was of a progressive, dual-stage design. It was designed to pay the shoulder belt out of the retractor at an initial force limit of 3 kN up to a defined amount of belt payout, followed by a secondary force limit of 4.5 kN for the remainder of the belt payout. In their 48 km/h Hybrid III AM50 tests, that study found that the FL+PT system decreased average peak upper shoulder belt loads from 8.7 kN (with the standard belt) to 4.4 kN. This was accompanied by decrease in internal chest deflection from 33 mm to 23 mm. Based on these measured chest deflections, that study predicted that the FL+PT system would decrease the risk of AIS 3+ chest injury in a 65 year old male from 28% (with the standard belt) to 11%. Due to the presence of the

pretensioner, the decrease in belt forces resulted in only a modest increase in forward head excursion.

Despite these initially encouraging results, subsequent cadaver studies suggested that chest injury may be problematic even for the FL+PT system described above. In a series of 48 km/h frontal impact sled tests with human cadavers, Forman et al. (2009) found that the same progressive FL+PT system as used in the dummy tests resulted in AIS 4 chest injuries in two out of the three cadavers tested. This was only a slight improvement from tests with a standard belt, which resulted in AIS 4 chest injuries in three out of three cadavers tested (Michaelson et al. 2008).

### **An “Airbelt” System**

This study seeks to investigate an alternative rear seat restraint system consisting of elements common to existing front seat restraints – a pretensioner and a force-limiter – combined with a novel restraint technology, a belt-integrated airbag. While belt-integrated airbags (“airbelts”) have been investigated (and even incorporated) to some extent in the aviation setting, few recent studies have investigated the possibility of incorporating airbelt systems into the automotive environment. In the early days of the development of inflatable restraints (in the 1970’s), NHTSA and collaborators studied the feasibility of inflatable belt restraints (in the front seat of automobiles) in a research program involving human volunteers, cadavers, and early-generation dummies (those studies are summarized by Digges et al. 1991). Despite promising results, that technology was not fully developed in favor of steering-wheel-hub-mounted driver airbags. In the 1980’s GM performed a series of (approximately) 60 km/h frontal sled tests with a Hybrid III ATD to study several designs of inflatable shoulder belts (Horsch et al. 1991). Those restraints did not include force-limiters, and resulted relatively high peak upper shoulder belt forces (around 10kN). Karigiri et al. (1999) reported a series of 53 km/h frontal sled tests with several sizes of Hybrid III ATDs restrained by a seat-integrated inflatable seatbelt. That study did not, however, report matched sled tests with conventional restraints for comparison (comparisons were only made to dummy responses in full-vehicle NCAP rigid barrier tests).

Through the design of the contour of an airbelt, restraining forces can be distributed over a larger area of the thorax than a standard diagonal belt (Horsch et al. 1991), and can be targeted to the stronger structures of the thorax (such as the shoulder). In the rear seat environment, an airbelt system may be also

able to provide restraint to the head and neck that is absent without a traditional frontal airbag. This study sought to investigate the feasibility and potential benefit that can be gained from an airbelt system in frontal impact ATD sled tests in a rear seat environment.

## METHODS

Six frontal impact sled tests with a Hybrid III 50<sup>th</sup> percentile male ATD were performed (Figure 1). Three were performed with a nominal change in velocity ( $\Delta V$ ) of 29 km/h; three were performed with a  $\Delta V$  of 48 km/h. The sled acceleration pulses were approximately trapezoidal in shape (Figure 2), and were chosen based on barrier tests of a mid-sized sedan (Forman et al. 2006a, Forman et al. 2006b).

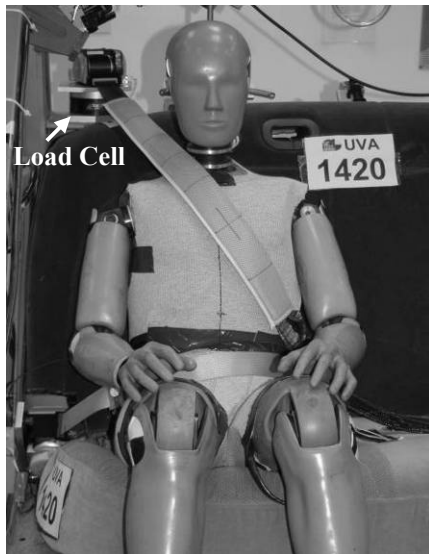


Figure 1: Passenger-side (top) and frontal (bottom) views of the Hybrid III AM50 ATD in its initial position, with the airbelt restraint installed.

The sled buck represented the rear-seat of a 2004 mid-sized sedan. The ATD was seated in the outboard, passenger's side position. The front seat

was removed for all tests to allow the capture of high-speed video from an anterior viewpoint. The seat cushion on which the dummy sat was replaced for each test. Dummy positioning was performed following the same procedure as Forman et al. (2008), which based the dummy initial position on the rear-seat occupant posture study of Reed et al. (2005). Although the front seat was removed for the tests, a reinforced mounting pedestal for the front seat remained on the buck. The rear surface of this pedestal was a rigid, flat, steel plate - this provided a positioning point for the feet (the feet were initially positioned with the tip of the shoes in contact with the plate), and provided restraint for the feet during the tests. A dummy neck shield (Melvin et al. 1993, Horsch et al. 1990) was not used.

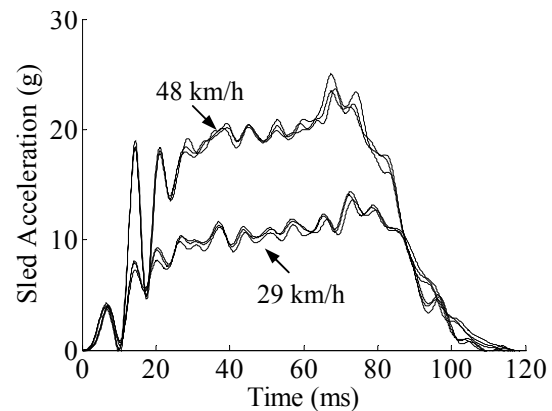


Figure 2: Sled acceleration time histories for the 48 km/h and 29 km/h tests.

## Restraints

The restraints in this study consisted of a three-point seatbelt with an airbag integrated into the upper portion of the shoulder belt (Figure 3). The restraint system had two retractors – one at the upper shoulder belt anchor and one at the outboard lap belt anchor. The upper shoulder belt retractor included a torsion-bar type force limiter, which limited the upper shoulder belt force to approximately 2.5 kN. The lap belt retractor (Figure 4) included a pretensioner, which produced a nominal 2.3 kN pretensioning force. Because this restraint was still in development, it did not include a traditional buckle on the inboard lap/shoulder belt anchor. The shoulder belt and the lap belt were separate sections of webbing, each tied directly to an inboard anchor bracket (Figure 4). This inboard anchor bracket also provided the mounting point for the airbelt inflator.

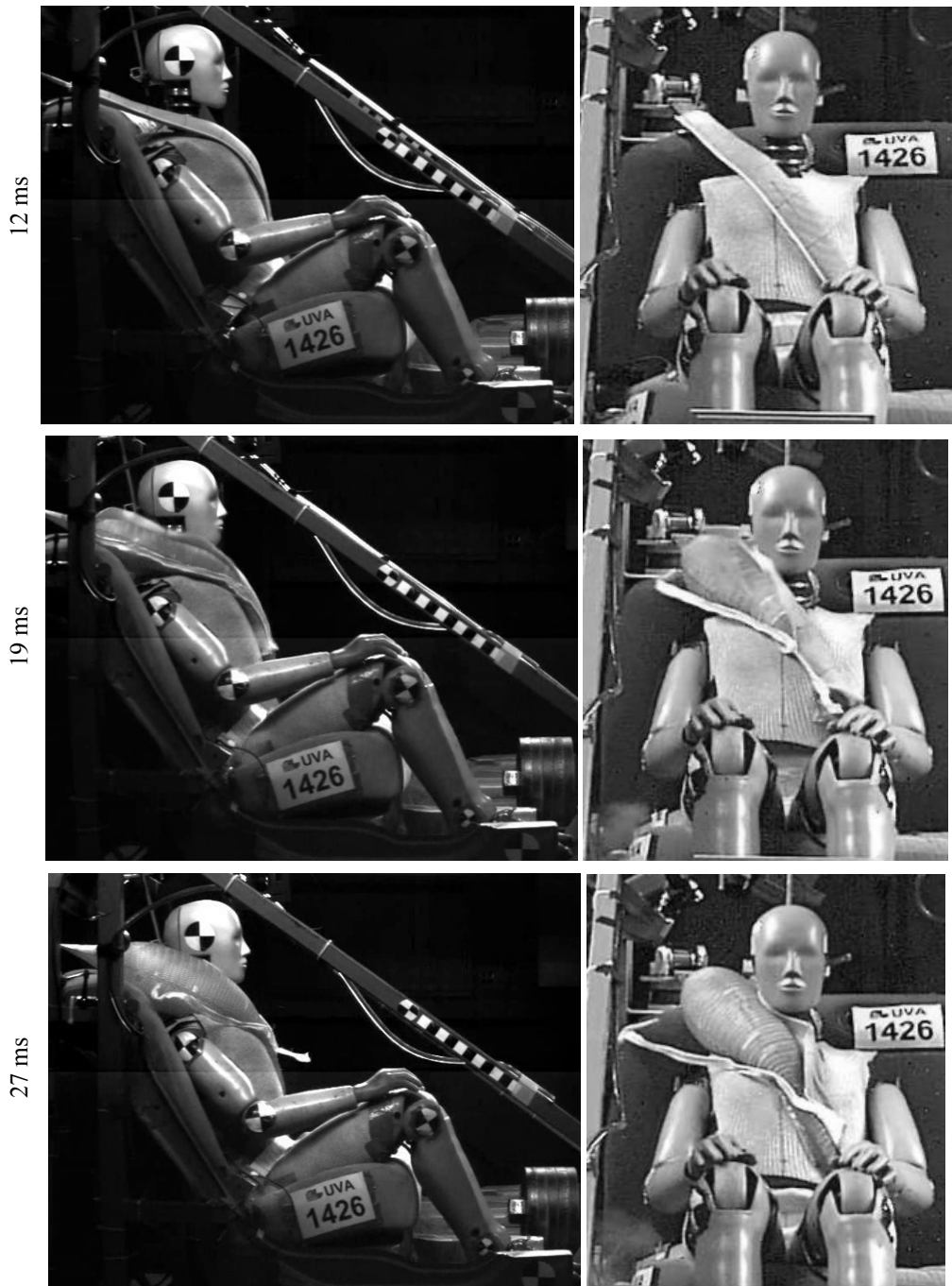


Figure 3: High speed video captures of a 48 km/h test showing the geometry and deployment of the belt-integrated airbag. The airbag and the lap belt pretensioner were both fired at  $t=12$  ms. The peak internal airbag pressure occurred at approximately  $t=27$  ms (estimated based on developmental tests).



Figure 4: Close-up views of the outboard lap belt retractor (top) and the inboard assembly of the lap and shoulder belt anchors and the airbelt inflator

The belt-integrated airbag was cylindrical in shape, with a maximum diameter of 160 mm. It was tapered such that the largest diameter was above or behind the shoulder, with the diameter decreasing to approximately 4 cm at mid-chest (Figure 3). This shape was chosen to increase the portion of the chest loading borne by the upper chest and shoulder (as opposed to the lower chest), to spread the load over a wider area of the upper chest, to help limit forward flexion of the neck, and to accommodate different sizes of occupants (the taper would tend to decrease the diameter of the airbag interacting with smaller occupants). The inflator was a hybrid type with a design pressure of 80 kPa. The total airbag volume was 7.2 L. The airbag was not vented. In developmental 48 km/h sled tests with an instrumented airbelt, the airbag reached a peak internal pressure of approximately 190 kPa (CFC 180) 15 ms after the firing of the airbag. This pressure then decreased to a plateau (steady) pressure of approximately 125 kPa for the duration of the test. In the current tests, both the airbag and the lap belt pretensioner were fired at 12 ms following the

initiation of the test ( $t_0$ ). The initial peak in lap-belt force caused by the pretensioner occurred at approximately 17 ms into the test (Figure 5). Considering the characteristics described above, it is estimated that the peak internal pressure of the airbelt occurred at 27 ms after  $t_0$ .

### Instrumentation and Analysis

The dummy measurements reported here include the acceleration resultant at the head center of gravity (c.g.), the chest c.g., and the pelvis; the tensile (z-axis) force and the flexion (y-axis) moment in the upper neck and the lower neck; the lateral (x-axis) moment in the lower neck; the shear (x-axis and y-axis) forces in the lower neck; and the posterior motion of the sternum relative to the spine (termed the “chest deflection”,  $C$ , as measured by the internal sternum slider). Tension in the outboard portion of the lap belt was measured by a belt tension gauge. Each of these signals were filtered according to the SAE J211 Channel Frequency Class (CFC) recommendations.

For the lower neck, the resultant shear force was calculated as the magnitude of the resultant between the x-axis and y-axis forces. For the upper neck, the neck injury criterion  $N_{ij}$  was calculated based on the algorithm used in the current FMVSS 208 frontal crash protection safety standard (Eppinger et al. 2000).  $N_{ij}$  is calculated based on a linear combination of the flexion-extension moment ( $M_y$ ) and the tension-compression force ( $F_z$ ) in the upper neck (Equation 1). For the tension-flexion formulation (used here), the equation constants (intercepts) are:  $M_{int} = 310$ ;  $F_{int} = 6806$ . The flexion moment is also corrected for the distance between the Hybrid III upper neck load cell and the location of the occipital condyles using the measured anterior-posterior (x-axis) upper neck shear force (method and code for calculation described in Eppinger et al. 2000).

$$N_{ij} = \frac{M_y}{M_{int}} + \frac{F_z}{F_{int}} \quad (1)$$

In all tests, the overall kinematics of the dummies were observed with off-board high-speed video captured at a rate of 1000 frames/second. For the 48 km/h tests, the forward head motion of the head c.g. (in the buck reference frame) was quantified from this video.

Because of the presence of the airbag, it was not possible to install a belt tension gauge on the upper portion of the shoulder belt. Thus, the force in the upper shoulder belt was measured with a six-axis

load cell mounted between the retractor and the sled (Figure 2). The force measurement of this load cell was compensated for the inertia of the retractor (and the sprung mass of the load cell) using acceleration measurements recorded by an accelerometer package mounted to the load cell. In post-processing, the effective mass of the retractor and load cell (as determined from sled tests with no subject) was multiplied by the acceleration measured at the load cell, and this was subtracted from the force measured by the load cell. The upper shoulder belt tension was then defined to be the force resultant after this inertial compensation.

Similar to the study of Forman et al. (2008), this study used the average peak chest deflections to estimate the chest injury risk that would result from these test and restraint conditions. To do this, this study used the age-dependent, Hybrid III AM50 chest injury risk function developed by Laituri et al. (2005). This injury risk function (Equation 2) predicts the risk of AIS 3+ chest injury based on the maximum internal sternum slider deflection ( $C_{max}$ ), and the age of the occupant. This function was used to estimate AIS3+ thoracic injury risk for a 45 year old and 65 year old occupant under these test conditions.

$$p(AIS3+) = \frac{1}{1 + e^{-(12.597 + 0.05861 \text{Age} + 1.568 C_{max}^{0.4612})}} \quad (2)$$

## RESULTS

The belt force time histories for the 29 km/h and 48 km/h airbelt tests are shown in Figure 5. Typical video captures at the time of maximum forward head excursion are shown in Figure 6. To compare the kinematics, Figure 6 also includes video captures from the standard belt and FL+PT belt, rear seat, Hybrid III AM50 tests of Forman et al. (2008). Anterior and lateral video captures illustrating the kinematics throughout the airbelt tests are shown in Figures 7 and 8.

Select peak data values (averaged across trials) are shown in Table 1. Table 1 also includes calculated values for the 3 ms clip peak resultant accelerations for the head c.g., chest c.g., and pelvis; the 15 ms Head Injury Criterion ( $HIC_{15}$ ); and the upper neck injury criterion  $N_{ij}$ . Each of these are compared to the tests of Forman et al. (2008) using a two-tailed Student's t-test (to test for actual differences in the restraint groups, as opposed to perceived differences resulting from random inter-test variation). The results were deemed to be (statistically) significantly different if the p value is less than or equal to 0.05.

Table 1 also includes the Hybrid III AM50 Injury Assessment Reference Values (IARVs) currently used in the FMVSS 208 frontal crash safety standard. For certain measures not included in FMVSS 208, Table 1 includes IARV estimates described by Mertz et al. (2003).

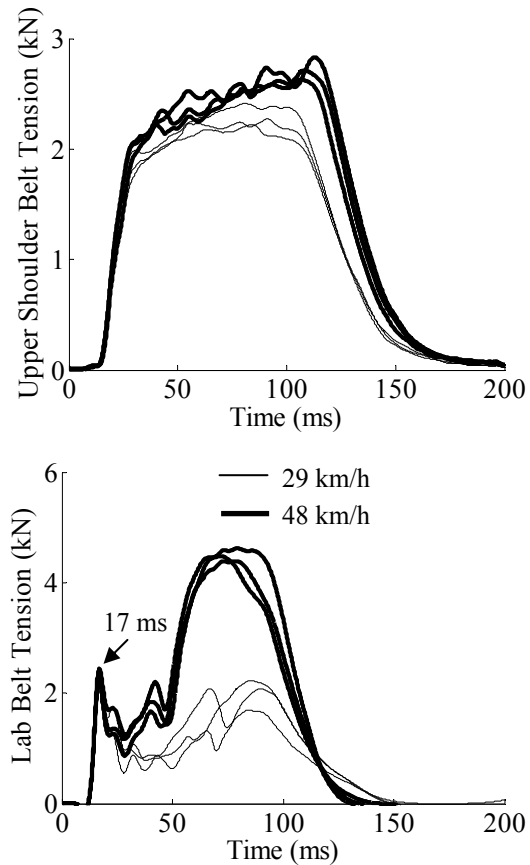


Figure 5: Belt force time-histories in the airbelt tests. Top: Upper shoulder belt force (measured at the anchor, inertially compensated for the acceleration of the load cell and the retractor). Bottom: Out-board lap-belt tension. Both the airbelt and the lap-belt pretensioner were fired at 12 ms. The initial peak in lap-belt tension (resulting from the pretensioner) occurred at approximately 17 ms.

In the 29 km/h airbelt tests, the peak and 3ms clip head acceleration, chest acceleration, pelvis acceleration, and lower neck tension were similar (not significantly different) to the FL+PT tests, but were significantly lower than the standard belt tests. The outboard lap belt tension, upper shoulder belt tension, upper neck tension, upper neck flexion moment,  $N_{ij}$ , lower neck flexion moment, lower neck resultant shear force, and chest deflection were significantly lower in the 29 km/h airbelt tests than both the FL+PT belt and the standard belt. In the 48 km/h tests, all of the dummy measures and restraint

forces shown in Table 1 – except for the lower neck lateral moment and the pelvis acceleration - were significantly lower with the airbelt system than with the FL+PT and standard belt restraints. The airbelt did result in greater lower neck lateral moments for all cases except when compared to the 48 km/h

standard belt tests, however this moment was small compared to the lower neck flexion moments. The airbelt system also resulted in a small, but statistically significant increase in forward head motion in the 48 km/h tests (average of 7 cm increase compared to the FL+PT system).

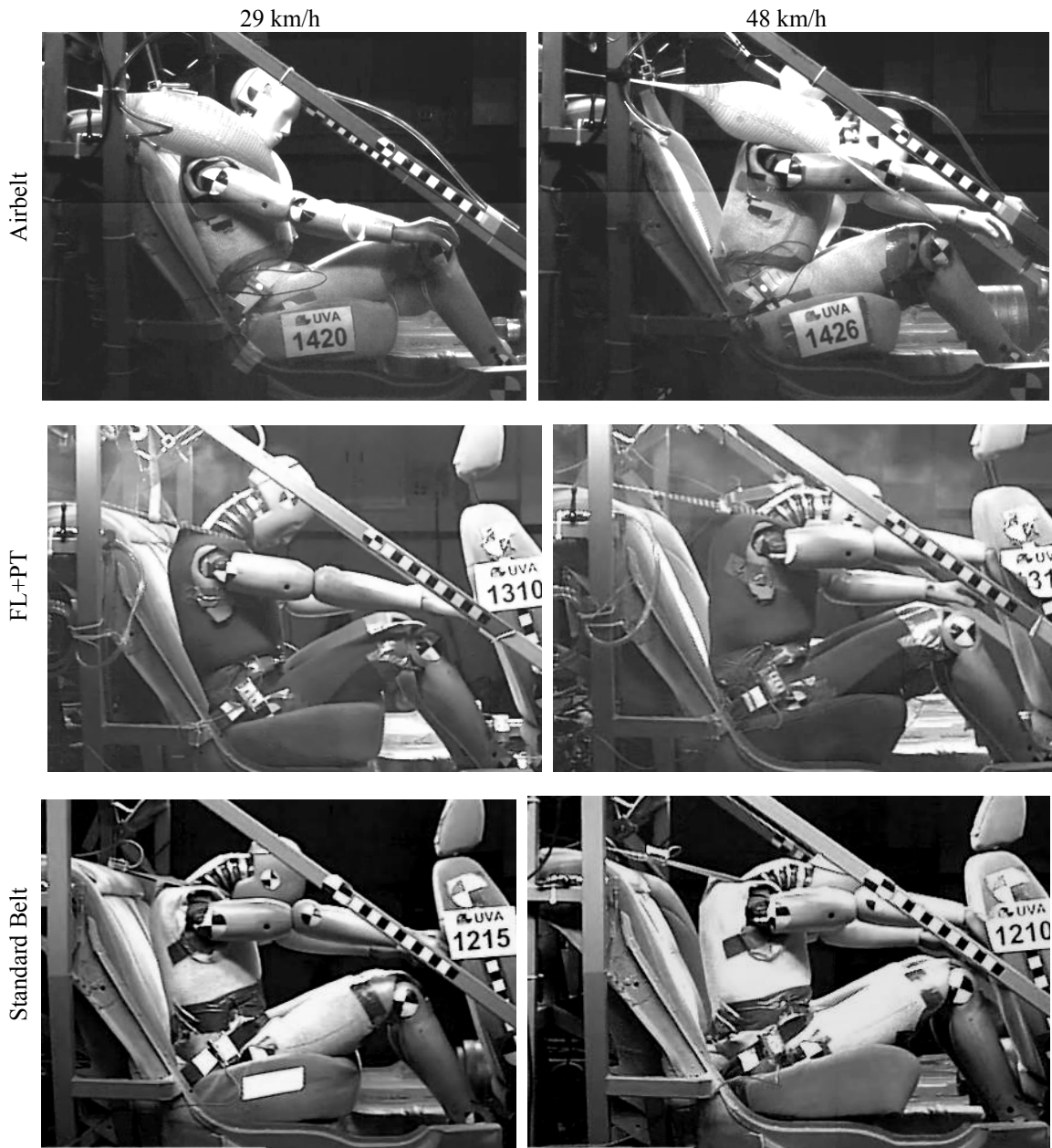


Figure 6: Video captures at the time of maximum forward head excursion for the airbelt tests, compared to the FL+PT and standard belt tests presented by Forman et al. (2008). Left: 29 km/h tests. Right: 48 km/h tests. In the Forman et al. (2008) tests, the front seat was installed in its OEM midtrack position.

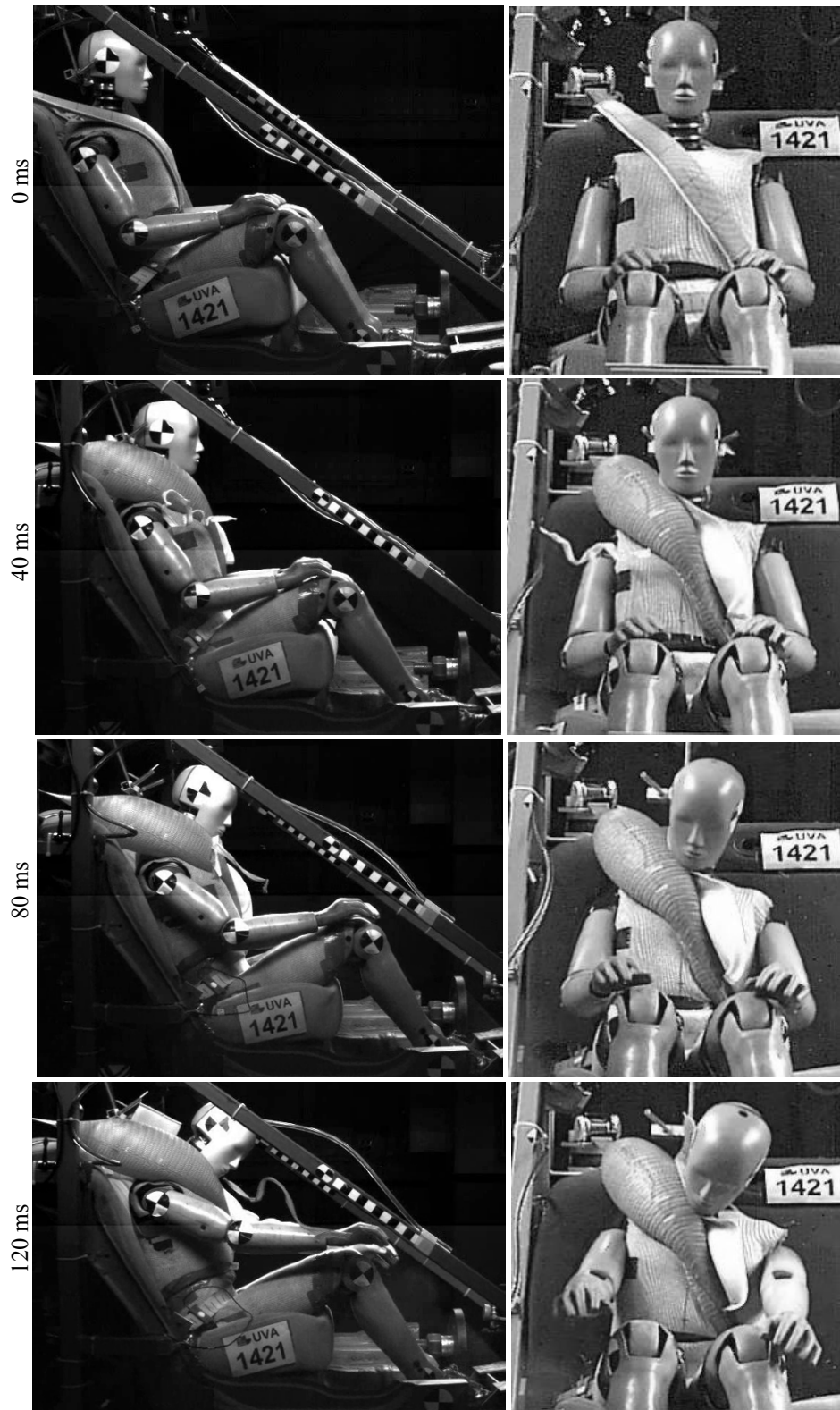


Figure 7: Lateral and anterior high-speed video captures of a 29 km/h airbelt test. Maximum forward excursion occurred at approximately 120 ms.





Figure 8: Lateral and anterior high-speed video captures of a 48 km/h airbelt test. Maximum forward excursion occurred at approximately 120 ms.

Table 1: Peak Data Summary, Compared to Previous Tests§ (Avg. ± Std. Dev., All Accelerations are Resultants)

Subject	H3 AM50						IARV
	29 km/h			48 km/h			
Target ΔV							
Restraint	Airbelt	FL+PT	Std.	Airbelt	FL+PT	Std.	
Actual ΔV (km/h)	29.3±0.35	29.5±0.3	29.4±0.3	48.3±0.18	48.5±0.4	47.8±0.0	--
Head C.G. acceleration, g (CFC 1000)	17±0.4	18±2.4	32±0.5*	36±2.2	44±0.5*	64±2.5*	--
[3 ms clip]	[16±0.4]	[17±3.1]	[32±0.3]*	35±2.0	[43±0.4]*	[64±2.4]*	--
HIC <sub>15</sub>	15±1.4	18±7	77±1.3*	108±14	167±6*	439±36*	700‡
Chest c.g. acceleration, g (CFC 180)	15±1.4	12±4.9	24±0.4*	30±0.6	22±1.4*	39±0.4*	--
[3 ms clip]	[15±1.4]	[12±5.2]	[23±0.5]*	[28 ±0.3]	[21±1.4]*	[38±0.4]*	60‡
Pelvis acceleration, g (CFC 1000)	20±2.7	23±2.0	27±1.7*	41±1.8	38±0.9	38±0.4	--
[3 ms clip]	[20±2.7]	[22±2.0]	[26±2.2]*	37±0.6	[34±2.1]	[37±1.0]	--
Upper shoulder belt tension, kN (CFC 60)	2.3±0.09	2.8±0.06*	5.3±0.20*	2.7±0.10	4.4±0.07*	8.7±0.25*	--
Outboard lap belt tension, kN (CFC 60)	2.3±0.06	3.9±0.19*	4.2±0.14*	4.5±0.12	6.1±0.46*	7.1±0.15*	--
Upper neck tension, kN (CFC 1000)	0.58±0.04	0.69±0.01*	1.27±0.03*	1.42±0.11	1.89±0.10*	2.43±0.05*	4.17‡
Upper neck flexion (y-axis) moment, Nm (CFC 600)	11±1.9	38±2.2*	68±0.3*	40±0.7	77±6.0*	91±5.3*	190♦
Nij (tension-flexion)	0.13±0.01	0.19±0.01*	0.31±0.001*	0.29±0.002	0.39±0.01*	0.70±0.02*	1.0‡
Lower neck tension, kN (CFC 1000)	0.68±0.17	0.55±0.04	1.02±0.07*	1.13±0.10	2.02±0.06*	1.52±0.18*	4.17♦
Lower neck shear resultant (x and y axes), kN (CFC 1000)	0.46±0.02	0.63±0.05*	1.22±0.16*	0.82±0.07	1.43±0.11*	2.37±0.49*	3.10♦
Lower neck flexion (y-axis) moment, Nm (CFC 600)	99±9.2	181±9.3*	313±1.2*	169±3.3	308±19*	656±26*	380♦
Lower neck lateral (x-axis) moment, Nm (CFC 600)	55±1.0	18±4.6*	33±6.0*	60±6.3	47±2.8*	71±6.7	268♦
Chest Deflection†, mm (CFC 600)	9.2±1.1	13±0.3*	19±1.1*	13±1.1	23±2.6*	33±2.3*	63‡
Forward head motion, cm	NM	NM	NM	54±0.4	47±1.1*	45±1.3*	--

§ The airbelt results are compared to the standard belt and FL+PT results of Forman et al. (2008).

‡ Injury Assessment Reference Values used in the current FMVSS 208 frontal crash protection safety standard.

♦ IARV estimates described by Mertz et al. (2003)

\* Significantly different than the airbelt tests as determined by a two-tailed Student's t-test ( $p \leq 0.05$ ).

† Chest deflection: sternum slider measurements

NM = Not Measured

The peak chest deflections were also used to predict the risk of AIS 3+ chest injury for a 45 year old and a 65 year old male occupant using the injury risk function of Laituri et al. (2005). The predicted injury risks are shown in Table 2, compared to the injury risk predictions described by Forman et al. (2008) for their standard belt and FL+PT belt systems. In the worst case shown, this analysis suggests that the airbelt system would result in a 2% risk of AIS 3+ chest injury for a 65 year old male occupant in a 48 km/h collision, compared to 11% and 28% risks for the FL+PT and standard belt systems of Forman et al. (2008).

## DISCUSSION

### Overall Kinematics and Belt Forces

The peak upper shoulder belt forces in these tests were significantly less than in the standard belt and FL+PT belt tests of Forman et al. (2008). This is not necessarily a function of the belt-integrated airbag, but is instead a result of the force-limiter included in the shoulder belt retractor of this system. The force limiter in this system yielded and paid out belt at 2.0 to 2.5 kN (Figure 5). In contrast, the force-limiter used in the FL+PT tests of Forman et al. (2008) was of a progressive type, where it would yield at 3 kN

for a defined amount of belt payout and then increase the force limit to approximately 4.4 kN for the remainder of the belt payout. The standard belt of Forman et al (2008) had no force limiter, resulting in the 8.7 kN peak upper shoulder belt forces noted in Table 1.

The less aggressive force limiter of the airbelt system resulted in a small increase in forward excursion of the head of the dummy (7 cm increase compared to the FL+PT system) in the 48 km/h tests (Table 1). Despite this small increase, Figure 6 shows that the dummy's head still would not have contacted the rear surface of the front seat, were it installed in the OEM mid-track position. The airbelt system resulted (at least qualitatively) in less forward head excursion than the FL+PT belt in the 29 km/h tests, despite a decrease in peak upper shoulder belt tension. This suggests that the belt-integrated airbag facilitates earlier, more consistent engagement of the thorax than the pretensioner integrated into the shoulder belt retractor of the FL+PT system. This is consistent with the minimal increase in forward head excursion observed in the 48 km/h tests, despite a substantial decrease in peak upper shoulder belt force.

Table 2: Predicted AIS3+ chest injury risk based on the ATD internal chest deflections ( $C_{max}$ )

$\Delta V$ (km/h)	Restraint	$C_{max}$ avg. (mm)	AIS3+ Injury Risk	
			45 year old	65 year old
29	Airbelt	9.2	<0.05%	1%
	FL+PT	13	<0.05%	2%
	Std.	19	2%	6%
48	Airbelt	13	<0.05%	2%
	FL+PT	23	4%	11%
	Std.	33	11%	28%

As observed in previous rear seat sled test studies in this environment, this particular seat cushion and seat pan geometry renders it difficult to limit forward motion of the pelvis (Forman et al. 2008, Forman et al. 2009, Michaelson et al. 2008). Even with the FL+PT system used by Forman et al. (2008) (which included a pretensioner in the shoulder belt retractor), considerable forward motion of the pelvis was observed in 48 km/h tests with the Hybrid III AM50 (Figure 6). This forward motion of the pelvis can limit the forward rotation – in the standard belt tests the torso remained angled backwards throughout the entirety of the tests. In contrast, because of the presence of the lap belt pretensioner the airbelt system resulted in very little forward motion of the pelvis. This, coupled with the shoulder belt force-limiter, allowed greater forward rotation of the torso in the 48 km/h tests. The earlier engagement of the

pelvis also allowed a greater amount of ride-down time, which caused the peak lap belt force to be less than both the standard belt and FL+PT tests.

### Neck Shear, Tension, and Lateral Bending, and $N_{ij}$

Some of the largest concerns - and potential benefits - for this airbelt system concern injury to the neck. Because of its proximity to the head and neck, loading of the neck is a valid concern with this type of belt-integrated airbag system. Shear, lateral bending, and tensile loading of the neck (from engagement of the head) have the potential to occur during the deployment of the airbag. The results of these tests indicate, however, that this airbelt system resulted in a significantly lower peak shear resultant force in the lower neck compared to both the standard belt and the FL+PT belt in both test speeds. In the 48 km/h standard belt tests, the peak lower neck shear ( $2.37 \pm 0.49$  kN) approached the IARV estimate of 3.1 kN (Mertz et al. 2003); the peak shear with the airbelt ( $0.82 \pm 0.07$  kN) was substantially lower. The airbelt also resulted in significantly less tension (in both the upper and lower neck) compared to the standard belt in both test speeds; less upper and lower neck tension than the FL+PT system in the 48 km/h tests; less upper neck tension than the FL+PT system in the 29 km/h tests; and similar (not significantly different) lower neck tension to the FL+PT system in the 29 km/h tests. In all cases, however, the peak upper and lower neck tensions were substantially lower than their respective IARVs.

The airbelt system did result in greater peak lower neck lateral moment for most cases (except when compared to the 48 km/h standard belt tests). These lateral moments were small, however, relative to the peak flexion moments – the greatest average peak lateral moment in the airbelt tests ( $60 \pm 6.3$  Nm) was approximately 60% of the lowest observed average peak flexion moment in the lower neck ( $99 \pm 9.2$  Nm, airbelt, 29km/h), and was less than 20% of the average peak lower neck flexion moments resulting from the 48 km/h FL+PT and standard belt tests ( $308 \pm 19$  and  $656 \pm 26$  Nm, respectively). In all cases, the peak lateral bending moments were well below the IARV estimate of 268 Nm (Mertz et al. 2003).

The airbelt system also resulted in a significant reduction in the upper neck injury criterion,  $N_{ij}$  (tension-flexion), compared to both the standard belt and the FL+PT belt under both test speeds. Both the  $N_{ij}$  and the peak upper neck tension forces were, however, below the FMVSS 208 Injury Assessment Reference Values for all tests. This is consistent with

the absence of upper cervical spine injuries in the studies of Michaelson et al. (2008) and Forman et al. (2009), which described standard belt and FL+PT belt cadaver sled tests matched with the 48km/h dummy tests described herein.

Neck shields are commonly used in studies of close-proximity (e.g., out-of-position) deployment of frontal airbags into Hybrid III ATDs (e.g., Prasad et al. 2008). These neck shields are designed to prevent the airbag from entering the cavity between the rear surface of the ATD's chin and the front surface of its neck during the membrane phase of the deployment (Banglmaier et al. 2005). Other than preventing this intrusion of the airbag into the chin cavity, the other fundamental design requirement of ATD neck shields is that they not affect measured neck loads in test situations that do not result in intrusion of the airbag into the chin cavity (although some variation exists in the literature, the most recent publications indicate that this requirement should hold true for all test situations, not just neck calibration tests; Banglmaier et al. 2005). In the current study, the limited size of the airbelt prevented it from deploying into the chin cavity of the ATD. As a result, it is unlikely (by the very definition of the design requirements, Banglmaier et al. 2005) that a properly designed ATD neck shield would have influenced the neck loads or moments measured in the current study.

The results above suggest that the airbelt should not increase the tensile and shear loading of the neck of an adult under these collision conditions, and in some cases may reduce it compared to the standard and FL+PT belt systems studied here. The airbelt does, however, have the potential to engage vulnerable structures on the lateral and anterior aspects of the neck (such as the carotid arteries, the vagus nerve, and possibly the trachea) that are not usually engaged in frontal collisions by properly-positioned shoulder belts. It is currently not possible to assess injury risk to these structures using a Hybrid III ATD – further work is needed to assess the risk of injury to these structures with a different occupant model (possibly through tests with human cadavers). Lastly, these tests only investigated one type of collision (frontal) with one size of occupant (50<sup>th</sup> percentile male) – more work is needed to study neck injury risk in other collision scenarios, with other occupant sizes (particularly children), and with out-of-position deployment.

### **Lower Neck Flexion**

One of the potential benefits of this airbelt system is its ability to provide restraint to the head to reduce inertial loading of the neck. Severe lower neck

injuries (dislocations, ligament tears, spinal cord injury) were observed in the standard belt, 48 km/h rear seat cadaver sled tests of Michaelson et al. (2008). These were attributed to excessive forward flexion of the neck, caused by the rapid deceleration and backwards lean of the torso combined with a lack of an airbag to provide supplemental restraint to the head. This was consistent with the relatively large lower neck flexion moments observed in the matched Hybrid III standard belt tests ( $656 \pm 26$  Nm), which were 72% greater than the current estimate for a lower neck flexion moment IARV (380 Nm, Mertz et al. 2003). A reduction in neck injuries was observed in the FL+PT cadaver study of Forman et al. (2009), however one C2 fracture (out of three tests) was observed. This was also consistent with the observed Hybrid III lower neck moment ( $308 \pm 19$  Nm), which was less than (but approached) the IARV. Although it is unknown to what extent the injuries observed in those cadaver tests represent what occurs in living humans, some field data do suggest that inertial-loading neck injuries can occur in restrained occupants in frontal collisions when an airbag is not present to restrain the head (Frampton and Lenard 2009, Hassan et al. 1996, Huelke et al. 1992, Forman et al. 2009).

The airbelt system resulted in a reduction in the peak flexion angle of the lower neck (Figure 6), through a combination of providing restraint to the head (especially in the 29 km/h tests) and by allowing greater forward rotation of the torso (in the 48 km/h tests). The airbelt provided a “ride-down” period of energy absorption through the payout of belt from the shoulder-belt force limiter (similar in effect to a front-passenger airbag, which provides a ride-down period through venting of the airbag). This system resulted in a nearly 50% reduction in the peak lower neck flexion moments in both the 29 km/h and 48 km/h tests when compared to the FL+PT system, and resulted in reductions of 68% and 75% compared to the 29 km/h and 48 km/h standard belt tests (respectively). As a result, the peak lower neck flexion moments with the airbelt were substantially lower than the current IARV estimate (45% of the IARV in the 48 km/h tests).

These results suggest that this type of system would tend to decrease the inertial flexion loading of the lower neck relative to the standard belt and FL+PT belts (studied herein) under these collision conditions, for a 50<sup>th</sup> percentile male. Anatomical and mechanical differences may, however, cause the airbelt to interact with the head and neck of a human in a different manner than with the Hybrid III (Melvin et al. 1993). Determining the actual effect of

the airbelt on the neck injury risk requires further study (possibly with tests with another occupant model).

### **Chest Injury Risk**

Chest injury mitigation is of particular concern in the restraint of adult rear seat occupants (Kent et al. 2007, Kuppa et al. 2005). The most commonly injured region in restrained rear seat adult occupants in frontal collisions is the chest, with 76% of AIS3+ injuries in persons over the age of 13 occurring in the thorax (Parenteau and Viano 2003). In their 48 km/h frontal cadaver sled tests, Michaelson et al. (2008) found that the standard (not force-limited, not pretensioned) rear seat seatbelt resulted in substantial chest injuries in adult cadavers of advanced age, with each of the three subjects receiving AIS 4 chest injuries. Even with the FL+PT system described above, Forman et al. (2009) observed AIS 4 chest injuries in two out of the three cadavers tested at 48 km/h in this rear seat environment. This was attributed to the greater secondary force limit of that FL+PT system, which was designed to satisfy the static test requirements of FMVSS 209.

The results of this study suggest that the airbelt system results in a significant reduction in chest loading (compared to the standard and FL+PT systems studied), which may result in a decrease in the risk of chest injury to an adult occupant in these collision scenarios. In the 48 km/h tests, the airbelt system resulted in decreases in internal chest deflection of 43% and 60% compared to the FL+PT and standard belt tests (respectively) of Forman et al. (2008). This was likely the result of the decreased aggressivity of the load limiter in the shoulder belt of the airbelt system – the average peak upper shoulder belt force with the airbelt system was 2.7 kN, compared to 4.4 kN with the FL+PT system and 8.7 kN with the standard belt. This may have also been affected, though, by the geometry of the belt-integrated-airbag. The airbag integrated into the upper portion of the shoulder belt was tapered in shape so that it engaged the shoulder and the upper portion of the chest to a greater extent than the middle and lower portion of the chest. This may also have been affected by the lap belt pretensioner, which would tend to offload the chest by increasing the proportion of the whole-body restraining force that was borne by the pelvis.

Despite the chest injuries in the standard belt and FL+PT cadaver sled tests described above, the dummy chest measures (both deflection and 3 ms clip acceleration) did not exceed the FMVSS 208 chest injury IARVs in any of the test conditions studied

here (Table 1). This is likely the result of the specificity of the Hybrid III chest injury prediction to loading distribution (Kent et al. 2003a), and increased thoracic fragility with age (Kent et al. 2003b). As a result, the potential differences in thoracic injury risk resulting from the three restraint systems studied here cannot be assessed using the FMVSS 208 IARVs alone (especially if it desired to study effects on injury risk as a function of age). Some insight may be gained using the belt-loading-specific, age dependent thoracic injury risk function of Laituri et al.(2005) (Equation 2, Table 2). Even though the airbelt loading is not strictly belt loading (the bag distributes the load over a greater area of the upper chest than a standard shoulder belt), the Laituri et al. risk curve will tend to result in a conservative estimate of thoracic injury risk (will err towards predicting a greater risk) – as the load is distributed over a larger area of the thorax, the injury risk for a given value of Hybrid III sternum slider deflection tends to decrease (Kent et al. 2003a).

Based on the injury risk function of Laituri et al. (2005), these tests suggest that this airbelt system may result in a 2% risk of AIS 3+ chest injury for a 65 year old male occupant in a 48 km/h collision, compared to 11% and 28% risks for the FL+PT and standard belt systems of Forman et al. (2008). It is important to remember, however, that these estimations of chest injury risk are still approximations. The 48 km/h cadaver sled tests of Michaelson et al. (2008) and Forman et al. (2009) each suggested an increased estimate of chest injury risk for an elderly person, with 3/3 cadavers sustaining AIS 4 chest injuries with the standard belt and 2/3 cadavers sustaining AIS 4 chest injuries with the FL+PT belt. Independent of the absolute chest injury risk, however, these results suggest that the relative chest injury risk in these conditions would be less with this airbelt system than with either the FL+PT system or the standard rear seat belt system described by Forman et al. (2008).

### **CONCLUSION**

Six frontal impact sled tests (three 29 km/h and three 48 km/h) were performed with a Hybrid III 50<sup>th</sup> percentile male ATD in a rear seat environment. The restraints used consisted of a three-point belt including a belt-integrated airbag, a 2.5 kN force limiter in the shoulder belt retractor, and a pretensioner in the outboard lap belt. These tests resulted in substantial, statistically significant decreases in peak upper shoulder belt force (18% decrease in the 29 km/h tests; 48% decrease in the 48 km/h tests), outboard lap belt force (41%, 29 km/h; 25%, 48 km/h), lower neck flexion moment (45%, 29

km/h; 45%, 48 km/h), lower neck resultant shear force (26%, 29 km/h; 43%, 48 km/h), and chest deflection (29%, 29 km/h; 43%, 48 km/h) compared to previous tests with a force-limiting, pretensioning (FL+PT) 3-point belt in the same environment. The airbelt system also resulted in substantial, statistically significant decreases in peak upper shoulder belt force (57%, 29 km/h; 69%, 48 km/h), outboard lap belt force (45%, 29 km/h; 37%, 48 km/h), lower neck flexion moment (68%, 29 km/h; 74%, 48 km/h), lower neck shear resultant (62%, 29 km/h; 65%, 48 km/h), and chest deflection (52%, 29 km/h; 61%, 48 km/h) compared to previous tests with a standard belt in the same environment. The airbelt system qualitatively resulted in a decrease in maximum forward neck flexion angle, a decrease in forward motion of the pelvis, an increase in forward rotation of the torso (in the 48 km/h tests). The airbelt also resulted in a statistically significant, but small magnitude increase in maximum forward head excursion compared to the other restraints in the 48 km/h tests (7 cm increase compared to the FL+PT belt, 9 cm increase compared to the standard belt). Based on the measured chest deflections, it was predicted that the airbelt system would result in a substantial decrease in chest injury risk for an adult occupant in a 48 km/h collision, compared to both the FL+PT and standard belt systems. More work is needed to evaluate this system with other occupant models (e.g., cadavers), for other occupant sizes (e.g., children), for out-of-position occupants, and in other collision scenarios.

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