DEVELOPMENT OF NEW WHIPLASH PREVENTION SEAT

Masahide Sawada
Junji Hasegawa
Toyota Motor Corporation.
Japan
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ABSTRACT

Whiplash, or soft tissue cervical injury, is a common injury incurred by occupants of passenger cars in rear-end collisions. Despite much investigation into the cause of such injury, no single mechanism describes it completely. Proposed criteria focus on the relative motions of the head and the thorax, while few case studies have been made on the motions of the cervical vertebrae. Recently, the human body finite element model called "THUMS"(Total HUman Models for Safety) and the use of X-ray cineradiography devices by volunteers have accelerated the investigation into the motions of the cervical vertebrae.

Seats have been developed that are specially designed to reduce impact on the neck in rear-end collisions by simultaneously restraining the head and body of the occupant and controlling their motion relative to each other. We have developed a seat that also reduces local strain of the neck by preventing the rotation of the head, and that uniformly distributing the loads on the cervical vertebrae.

A finite element model was used to simulate rear-end collisions under the same conditions as sled tests using a BIO-RID II dummy, with a THUMS human model placed on our newly developed seat. Prior to the simulation, the validity of the THUMS was investigated by comparing its head and neck motions with those in experiments. The validated THUMS predicted a reduction of local strain in the neck on the newly developed seat.

Having succeeded in reducing both the injury values to the dummy and the local strain of the neck of the THUMS, we predict that our new seat design to help reduce whiplash injury.

1. INTRODUCTION

Soft tissue cervical injury (whiplash) is a common injury resulting from rear-end collisions in passenger cars. As shown in Figure 1 [1-2], rear-end collisions account for about 50% of accidents resulting in injury, although only a small number of them are fatal. About 80% of injured occupants suffer neck injury, and the reduction of whiplash is,

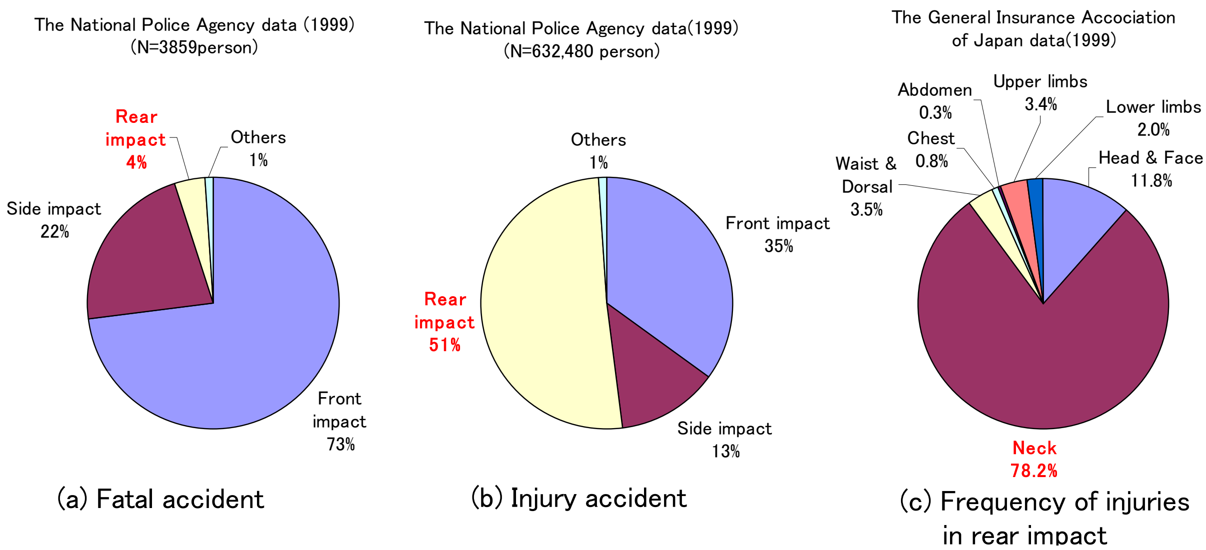


Fig.1 Realities of rear-end collision.

therefore, an important issue.

Krafft, et al. [4] analyzed the acceleration pulses of actual collisions using an in-vehicle data recorder. Based on this data, acceleration pulses obtained from sled tests conducted by rating organizations, including Folksam, IIWPG and ADAC, were considered. So far, a triangular pulse sled of ΔV = 16km/h is the most widely adopted test. In this test, the equivalent collision velocity is 32 km/h for a vehicle-to-vehicle collision between two cars of the same weight, which represents only a 60th percentile collision in Japan, as shown in Figure 2 [3]. If collision velocities of up to 50 km/h are examined, as much as 90% of all such collisions can be represented. In other words, when converted into ΔV for a collision between two cars of the same weight, the ΔV = 25 km/h.

Various attempts have been made to clarify the injury mechanism of whiplash. However, whiplash injury cannot be described with only a single mechanism, and so a variety of criteria have been proposed. Bostron, et al. [5] have proposed a criterion called a neck injury criterion (NIC) based on the variation in the pressure of the spinal fluid within the cervical spinal canal. Schmitt, et al. [6] have proposed Nkm focusing on the shear force and bending moment of the upper neck, whereas

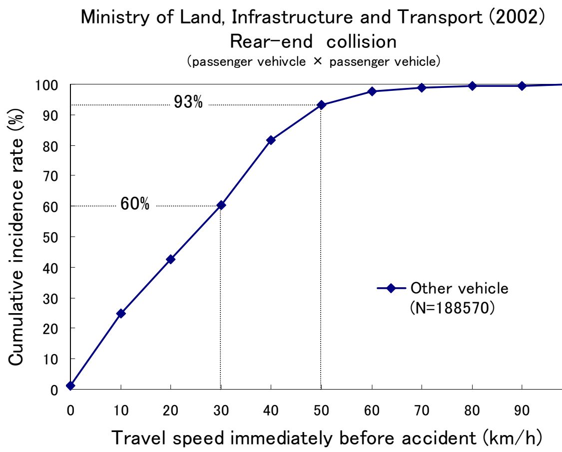


Fig.2 Cumulative incidence rate of travel speed immediately before accident.

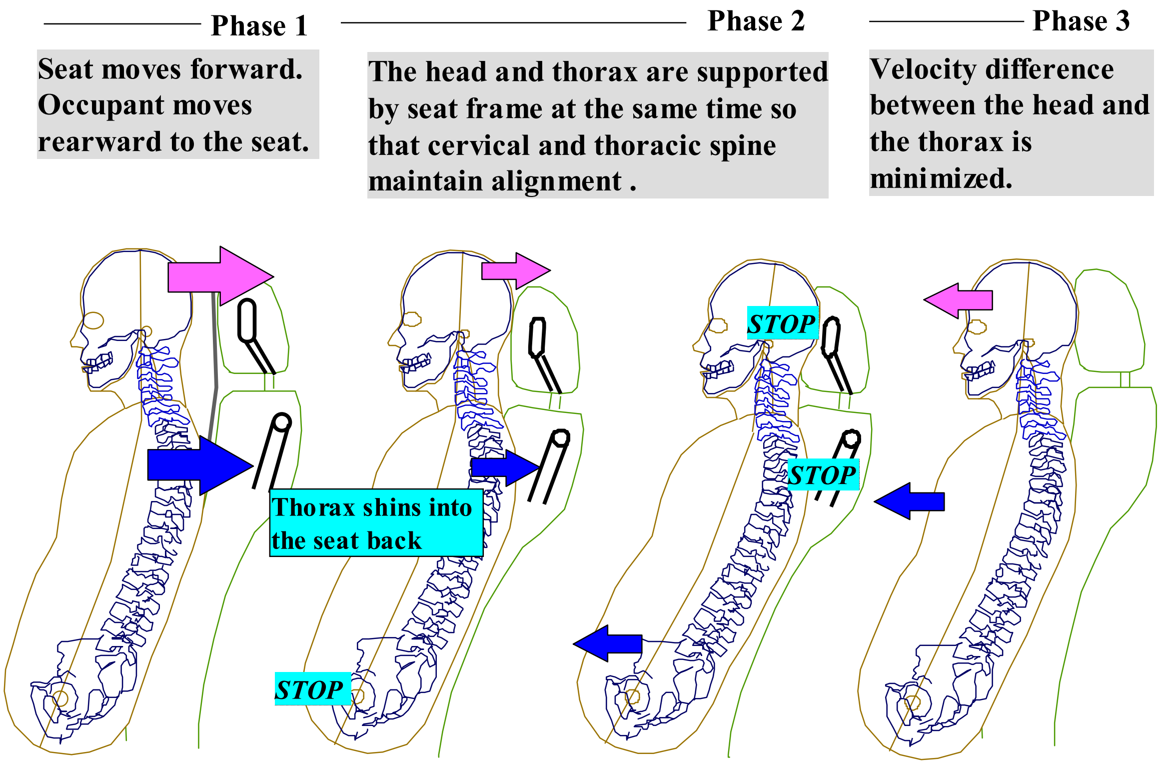


Fig.3 Whiplash injury lessening concept.

Heitplatz, et al. [7] have suggested a lower neck load index (LNL) that correlates with insurance claims for cervical vertebrae injuries, and emphasizes the shear force, axial force, and bending moment of the lower neck. Viano, et al. [8] have developed an advanced a neck displacement criterion (NDC) as a criterion for the movable range of the head and neck based on tests conducted on volunteers. Additionally, Panjabi, et al. [9] have presented IV-NIC for evaluating neck injury based on the ratio of it to the physiological limit rotating angle of the cervical vertebra joints.

Deng et al. [10] and Ono et al. [11] have analyzed the motions of the cervical vertebrae of corpses and of volunteers using X-ray cineradiography devices. On the other hand, Ejima et al. [12] and Hasegawa [13] have used human FE models to perform in-depth analyses of the stress and strain of the cervical vertebrae. Additionally, Lee et al. [14] have attributed cervical facet capsule distraction as a cause of neck pain by whiplash per their experiments on rats. Research into the mechanism of whiplash is thus shifting focus away from the relative motions of the head and the thorax and neck loads, toward the relationship of the local motions of the neck to whiplash.

We have developed a WIL seat [15] designed to reduce load on the neck based on a unique concept of the prevention of whiplash achieved by restraining the head and body of the occupant simultaneously and thereby controlling relative motions in a rear-end collision, as shown in Figure 3. Other companies have developed the shock absorption seat [16] and the active head restraint seat [17], both of which are intended to reduce loads on the neck. We have developed a seat that not only achieves the abovementioned objectives, but can also control local strain of the neck by preventing the rotation of the head and by uniformly distributing load on the cervical vertebrae. We also verified the effects of the seat through experiments using a BIO-RID II dummy and finite element (FE) analyses on a human FE model (THUMS).

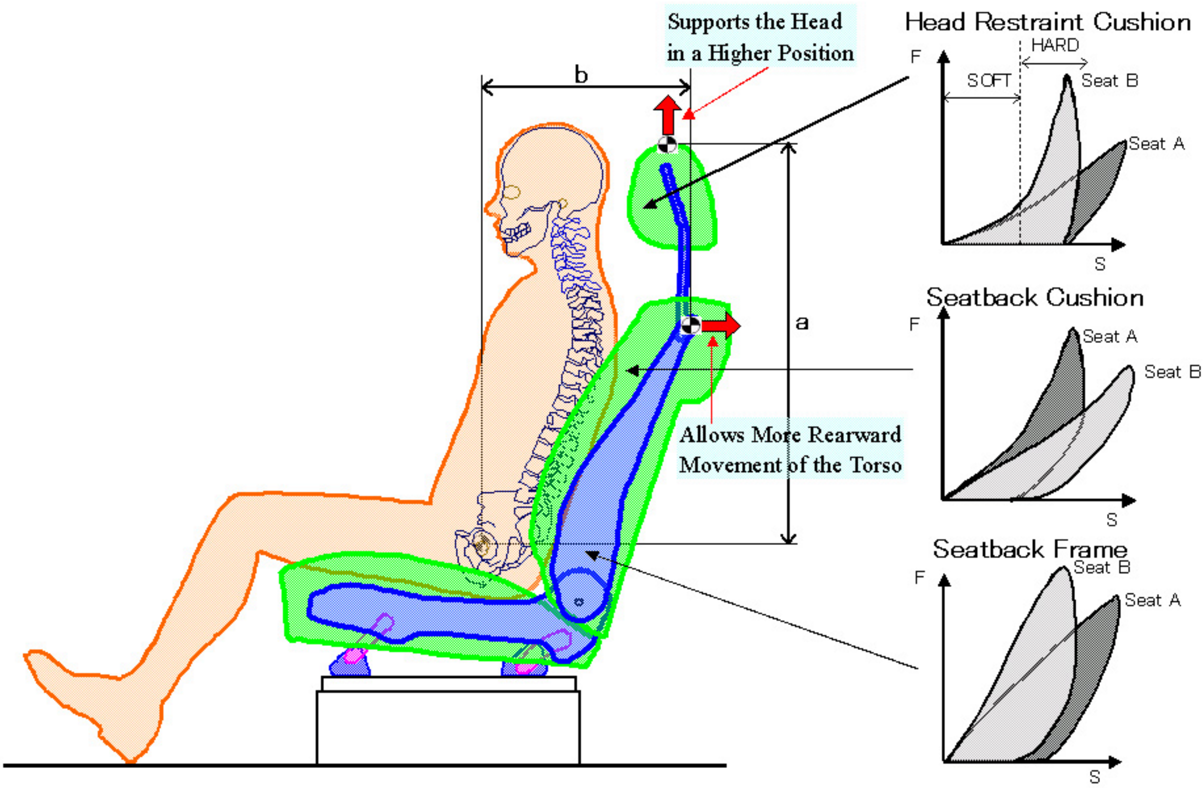


Fig.4 Comparison between a conventional seat (Seat A) and the newly developed seat (Seat B).

2. TEST METHODS

2.1 Test samples

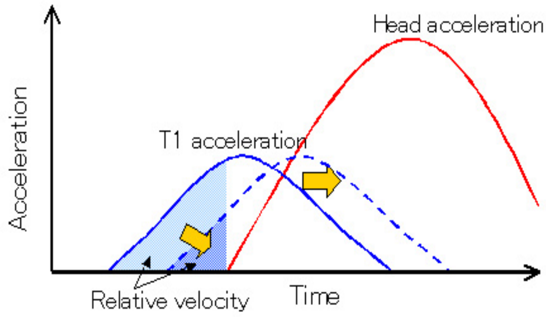
Figure 4 shows a comparison between a conventional seat (Seat A) and the newly developed seat (Seat B). One of the features of the new design is that the position (b) of the upper part of the frame and the stiffness of the seatback cushion were changed so that the thorax of the occupant sinks deeper into the seat in Phase 1 (until the head contacts the head restraint). This delays the onset of the rebounding motion of the thorax G (T1G) and reduces the velocity relative to head, as shown in Figure 5 (a). Similarly, the design of this newly developed seat (Seat B) aims to reduce the velocity relative to thorax G (T1G), but takes a different approach than the active head restraint, which causes the head G to rebound from the head restraint earlier as shown in Figure 5 (b).

We also reviewed the vertical position (a) of the head restraint and the F-S characteristic of the head restraint cushion to ensure restraint in Phase 2 (after the head contacts the head restraint). The balance of seat frame strength was also reconsidered, to provide reliable restraint performance at higher velocities up to ΔV = 25 km/h. These factors were changed with the aim of preventing the head of the occupant from extending over the head restraint even in a high-velocity rear-end collision by securely restraining the head at a higher position.

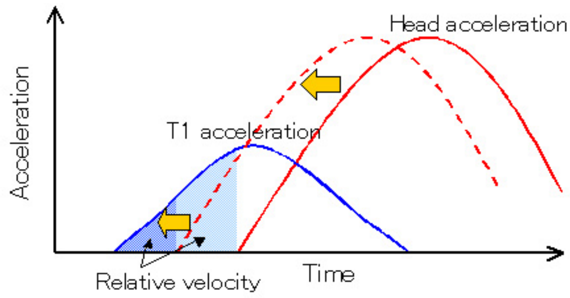
As mentioned earlier, currently there is no commonly accepted theory for the mechanism of whiplash at present, and a variety of criteria are proposed. In the following section, we verify the performance of the seat developed based on the abovementioned design through FE analyses using these proposed criteria and THUMS.

2.2 Test devices and conditions

An electrically controlled servo-hydraulic sled tester (Figure 6) was used to reliably generate free collision acceleration pulses. We conducted tests using a BIO-RID II (Figure 7) dummy, the most popular dummy for whiplash evaluation. The test method complies with the IIWPG test protocol [18]. Sled acceleration Pulses 1, 2, and 3 (shown in Figure 8) were used for testing. Pulse 1 is the most widely adopted acceleration pulse and a triangular pulse of ΔV = 16 km/h, tested by ADAC, Folksam and each



(a) Development concept



(b) Active head restraint system concept

Fig.5 Whiplash lesseing concept comparison.

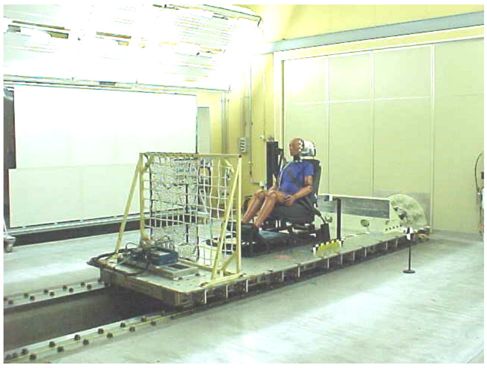


Fig.6 Electrically controlled servo-hydraulic sled tester.

Fig.7 BIO-RID II dummy.

IIWPG organization (IIHS and Thatcham). Since there is no standardized sled acceleration pulse of $\Delta V = 25$ km/h, we tested a triangular pulse of $\Delta V = 25$ km/h used by ADAC [19] for pulse 2, and a trapezoidal pulse of $\Delta V = 24$ km/h adopted by Folksam [20] for pulse 3.

3. RESULTS OF EXPERIMENTS AND DISCUSSION

This section discusses the results of conventional seat (Seat A) and the newly developed seat (Seat B), with sled acceleration pulses of pulses 1 to 3.

3.1 Discussion based on new criteria $T1G_{HRC}$, $T1V_{HRC}$ and $T1S_{HRC}$

Figure 9 shows the analysis results of thorax G (T1G). As shown in Figure 9 (a), there is no significant difference in $T1G_{max}$ between the conventional and new seats. However, the new seat produced lower values for T1G ($T1G_{HRC}$) until the head contacts the head restraint (T_{HRC}), for velocity variation $T1V_{HRC}$ (Expression 1) and for displacement $T1S_{HRC}$ (Expression 2) as shown in Figures 9 (b) to (d). This result indicates that the relative G, V, and S of the head and thorax are low in Phase 1 and head and thorax are more uniformly restrained. Figure 10 shows a comparison of head G

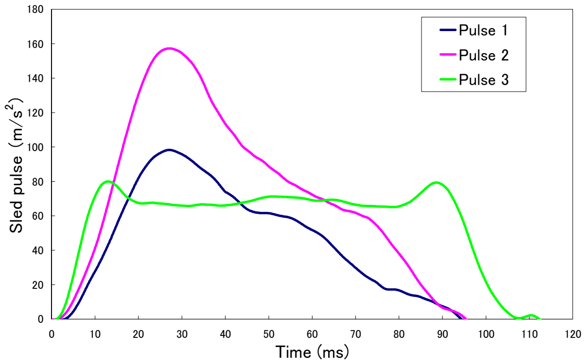
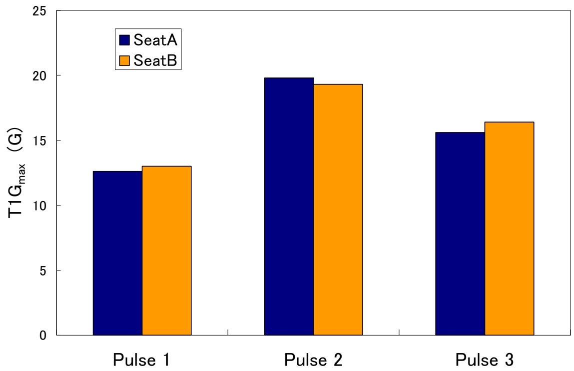
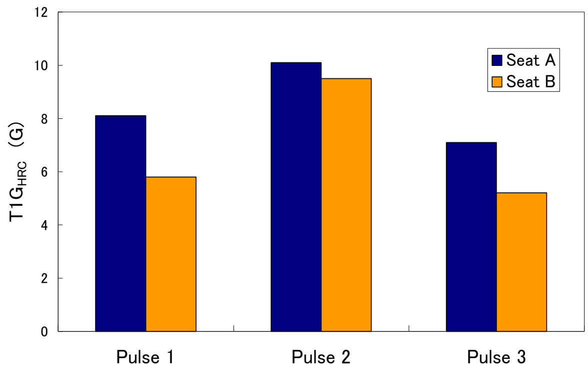


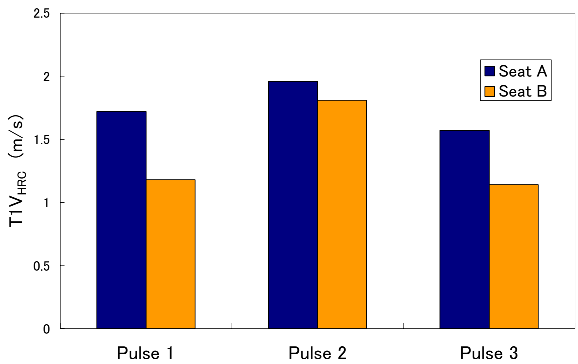
Fig.8 Test sled pulses.



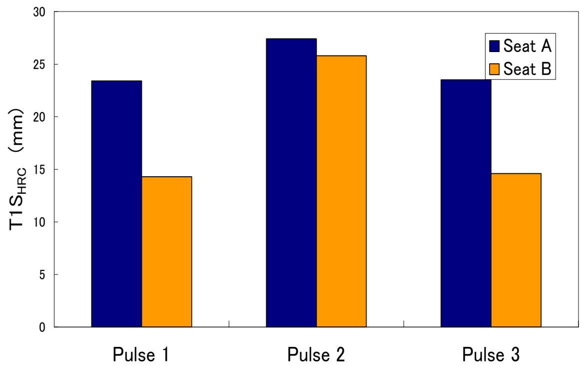
(a) $T1G_{max}$



(b) $T1G_{HRC}$

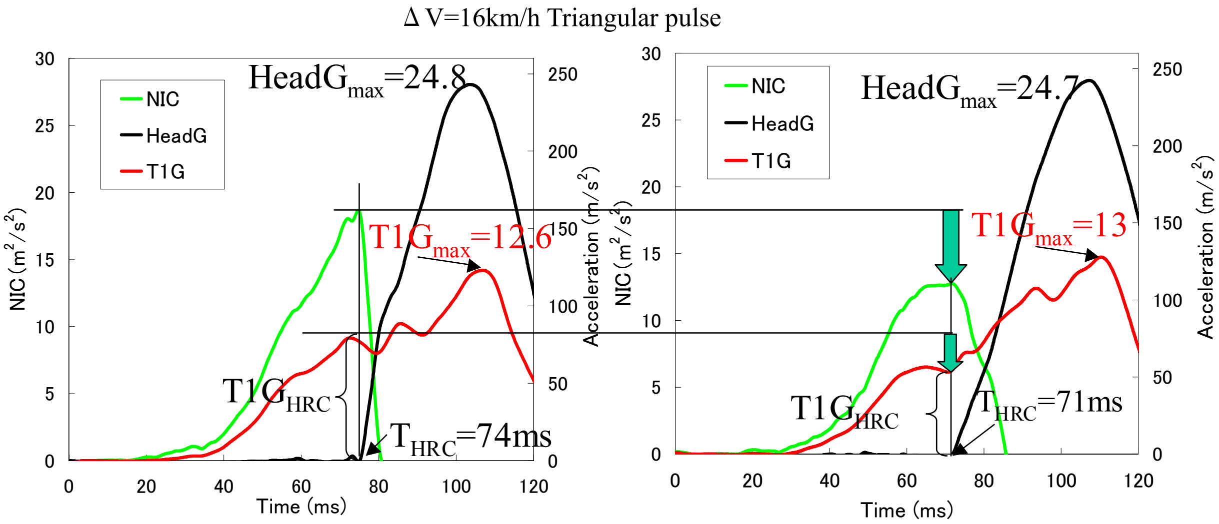


(c) $T1V_{HRC}$



(d) $T1S_{HRC}$

Fig.9 Examination with thorax G (T1G).



(a) Seat A (b) Seat B

Fig.10 Comparison of head G, thorax G, and NIC in Pulse 1.

and thorax G (T1G) in Phase 1 between the two seats. At maximum T1G, the head and the thorax are restrained, T1G(t) < HeadG(t), and the head is securely restrained and its extension restricted. Both seats had almost equal T_{HRC}, but $T1G_{HRC}$ (and $T1V_{HRC}$ and $T1S_{HRC}$) was lower in the new seat (Seat B). This indicates that the thorax sinks deeper into the seat in Phase 1 as intended.

$$T1V_{HRC} = \int_0^{T_{HRC}} T1G(t)dt \tag{1}$$

$$T1S_{HRC} = \int_0^{T_{HRC}} \int_0^{T_{HRC}} T1G(t)dt^2 \tag{2}$$

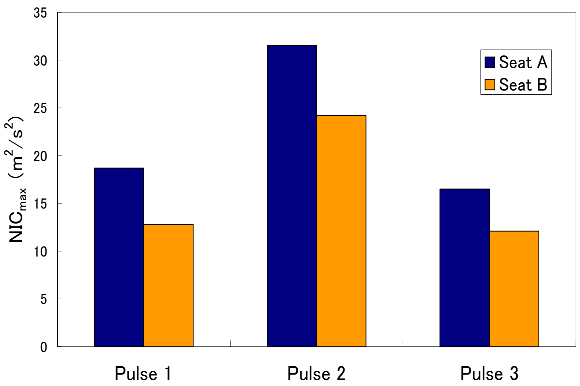


Fig.11 NIC examination.

3.2 Discussion based on conventional criteria

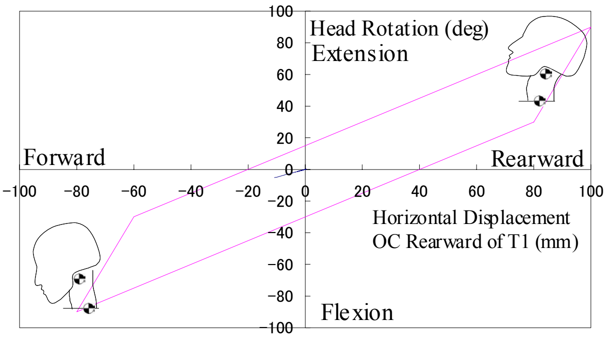
Figure 11 shows test results compiled based on the whiplash evaluation criterion NIC proposed by Bostron, et al. [5]. The NIC is a criterion for evaluating Phase 1 with emphasis placed on the relative motions of head G and thorax G (T1G) (Expression 3) and considered to be consistent with the new seat concept. NIC_{max} of the new seat (Seat B) was smaller than that of the conventional seat (Seat A) at the three sled pulses.

$$NIC(t) = 0.2 \cdot a_{rel}(t) + (V_{rel}(t))^2 \tag{3}$$

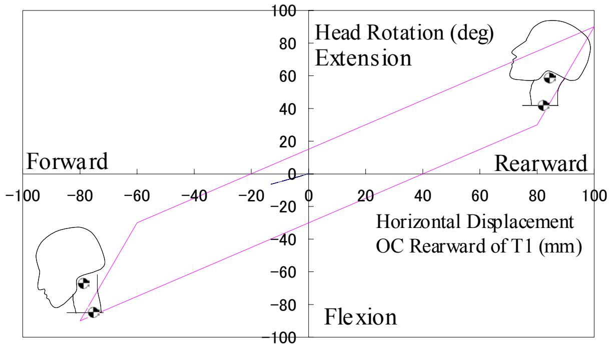
$$\text{where} \quad a_{rel} = T1G(t) - HeadG(t)$$

$$v_{rel} = \int a_{rel}(t)dt$$

Figure 12 shows the relationship between (a) horizontal displacement and head rotation at the NDC proposed by Viano, et al. [8] and between (b) horizontal displacement and vertical displacement. This criterion is considered to match the design concept of simultaneously restraining the head and the thorax and thereby preventing head rotation in Phase 2. Both seats securely prevent the vertical and rearward motions, rearward rotation of the head and the thorax, and are expected to provide a higher level of whiplash-reducing performance.

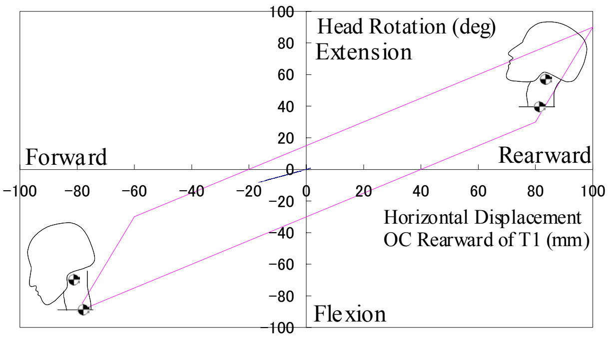


Seat A

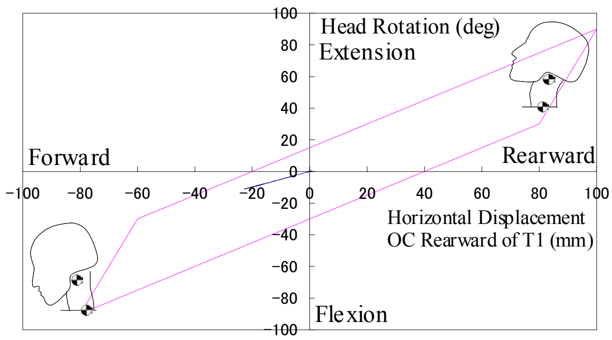


Seat B

(a) Pulse1 : Δ V=16km/h Triangular pulse

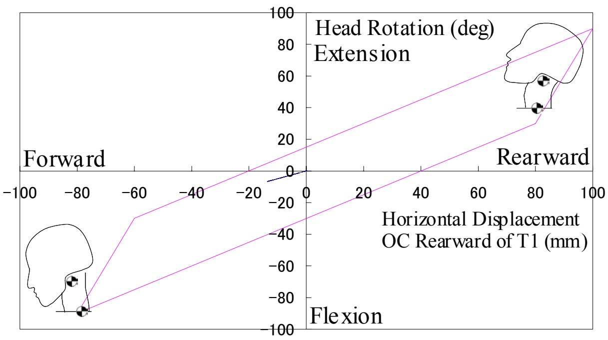


Seat A

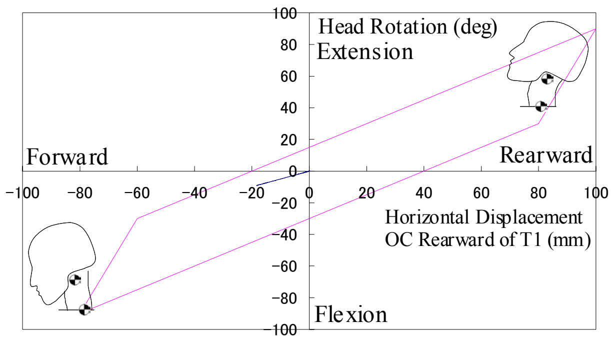


Seat B

(b) Pulse2 : Δ V=25km/h Triangular pulse



Seat A

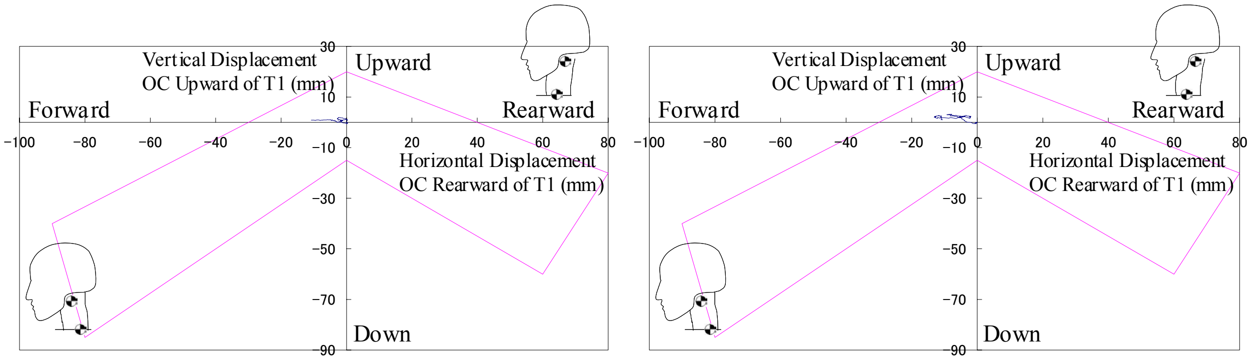


Seat B

(c) Pulse3 : Δ V=24km/h Trapezoidal pulse

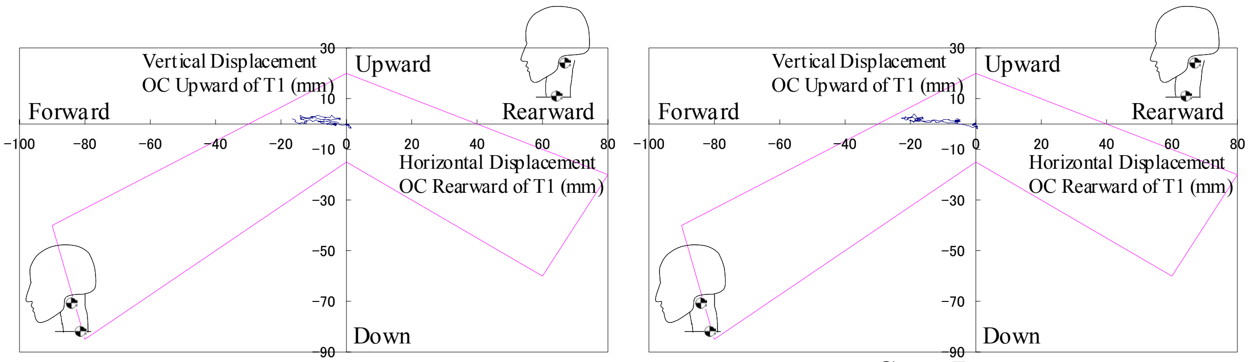
(a) Relationship between Horizontal Displacement and Head Rotation

Fig.12 NDC examination.



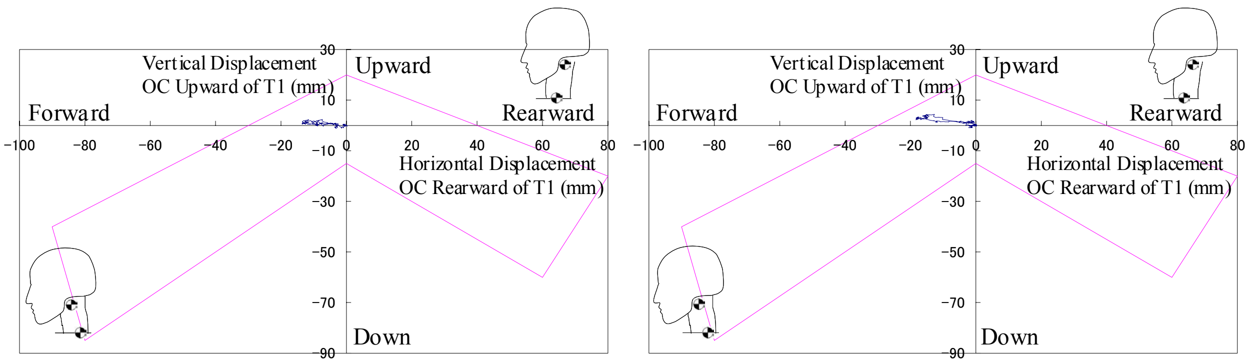
Seat A Seat B

(a) Pulse1 : Δ V=16km/h Triangular pulse



Seat A Seat B

(b) Pulse2 : Δ V=25km/h Triangular pulse



Seat A Seat B

(c) Pulse3 : Δ V=24km/h Trapezoidal pulse

(b) Relationship between Horizontal Displacement and Vertical Displacement

Fig.12 NDC examination.

Figure 13 shows the value of each component of the LNL (Expression 4) advanced by Heitplatz, et al. [7]. The LNL is a criterion focusing on the shear force, axial force, and bending moment of the lower neck in Phase 2. It draws attention as a criterion relating to the large angular variation and strain of the lower cervical vertebrae, such as C4-C5, C5-C6, and C6-C7, obtained from the results of the volunteer tests conducted by Sekizuka [15] and the results of the human FE analyses performed by Hasegawa [13]. The LNL indicates that the new seat (Seat B) produces lower values than the conventional seat (Seat A) at all three sled pulses and is, therefore, expected to yield a higher level of whiplash prevention.

Figure 14 shows the maximum value of lower neck extension moment My. This criterion was developed by focusing on the direct representation of load on the neck when it extends in Phase 2. This is also used in Expression 4 of the abovementioned LNL. The My_{max} of the new seat (Seat B) is lower than that of the conventional seat (Seat A) at all three sled pulses, suggesting that the former better prevents neck extension.

It can be seen in Figures 9 through 14 that the pulse shape exerts a significant effect on the injury value, for example there was a large difference in injury value between pulses 2 and 3 even though their ΔV is almost equal. On the whole, it was revealed that the injury value at pulse 2 was higher than that the value at pulse 3.

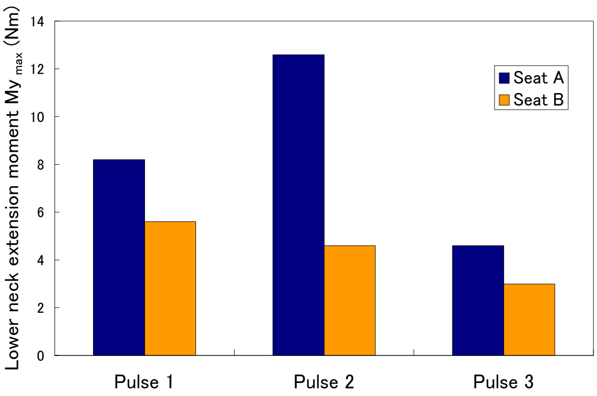
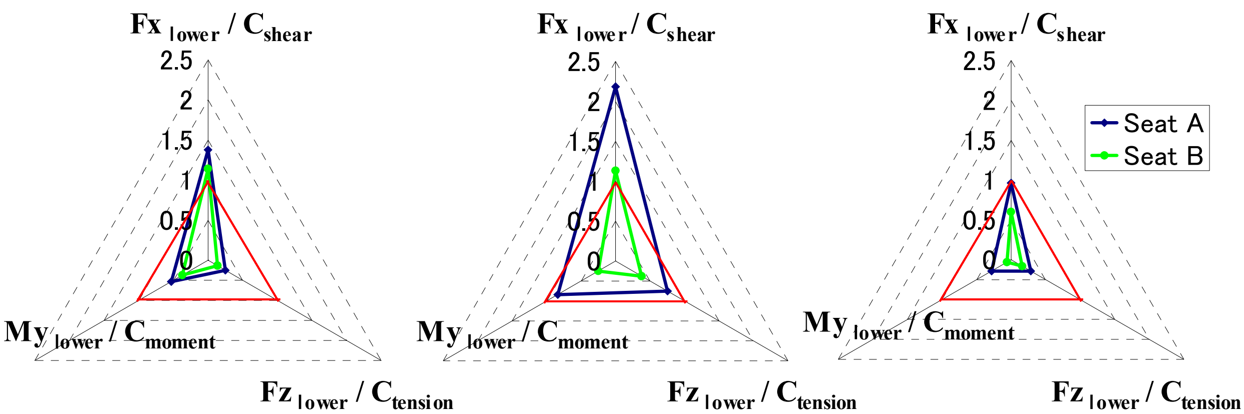


Fig.14 Lower neck extension moment My_{max} examination

$$\mathrm{LNL-index}(t)=\left|\frac{\sqrt{My_{lower}(t)^2+Mx_{lower}(t)^2}}{C_{moment}}\right|+\left|\frac{\sqrt{Fx_{lower}(t)^2+Fy_{lower}(t)^2}}{C_{shear}}\right|+\left|\frac{Fz_{lower}(t)}{C_{tension}}\right| \quad (4)$$

Intercept values : C_{moment}=15 , C_{shear}=250 , $C_{tension}$=900

In BIO-RID II dummy : $Mx_{lower}(t)=0$, $Fy_{lower}(t)=0$



(a) Pulse1
ΔV=16km/h Triangular pulse

(b) Pulse2
ΔV=25km/h Triangular pulse

(c) Pulse3
ΔV=24km/h Trapezoidal pulse

Fig.13 LNL examination.

4. FE ANALYSES

4.1 FE model and its verification

To confirm the ability of the new seat to reduce loads on the human neck and to analyze how load on the neck occurs, we used a human FE model called a total human model for safety (THUMS) to simulate rear-end collisions. Figure 15 shows an enlargement of the entire body and neck of the THUMS-AM50 OCCUPANT Ver.1.63-050304. This model is an upgraded version of conventional THUMS passenger model v1.5, and was subjected to significant improvements such as the added ability to represent the structure of the spinal cord and the cervical vertebrae joints in detail, in order to accurately reproduce the motions of the neck and to evaluate the strain of the soft neck tissues in rear-end collisions. We also prepared a seat model representing a conventional seat structure and one combining newly developed structural design attributes (Figure 16), both of which were given the strength characteristics obtained from component tests (Figures 17 and 18). A THUMS v1.63 was placed in almost the same posture as the BIO-RID II dummy in both seat models, and rear-end collisions were simulated by inputting acceleration pulses equivalent to those used for the sled experiments.

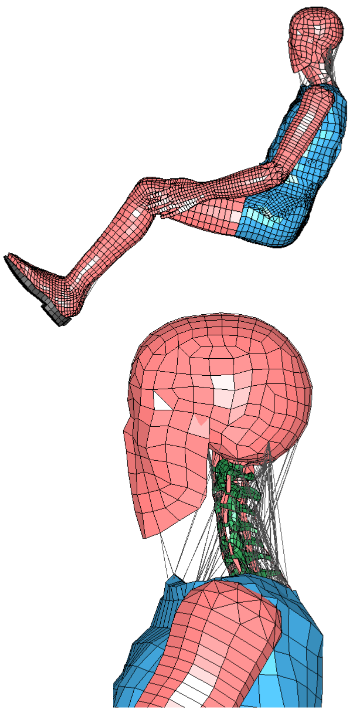


Fig.15 THUMS-AM50 OCCUPANT Ver.1.63-050304

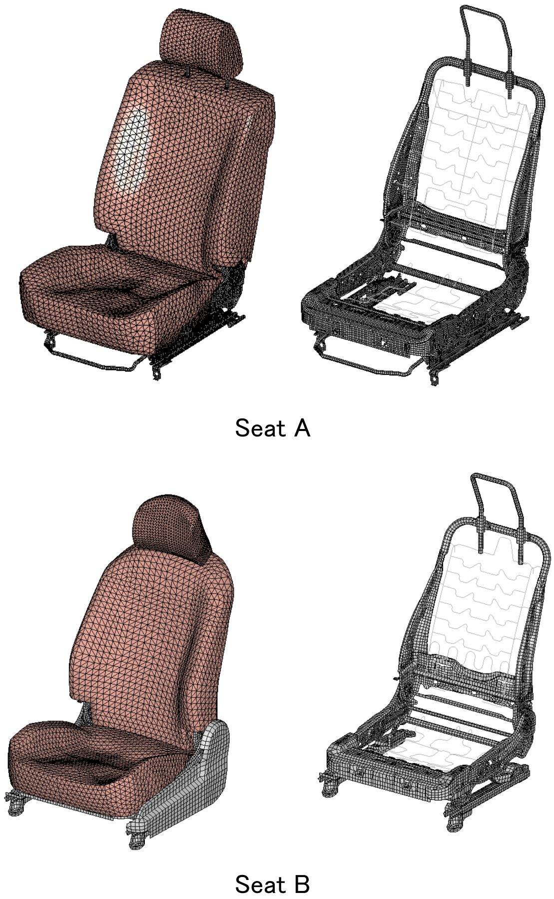


Fig.16 Seat FE model

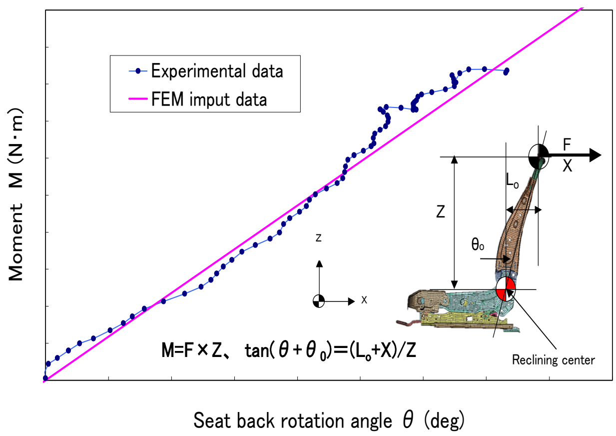
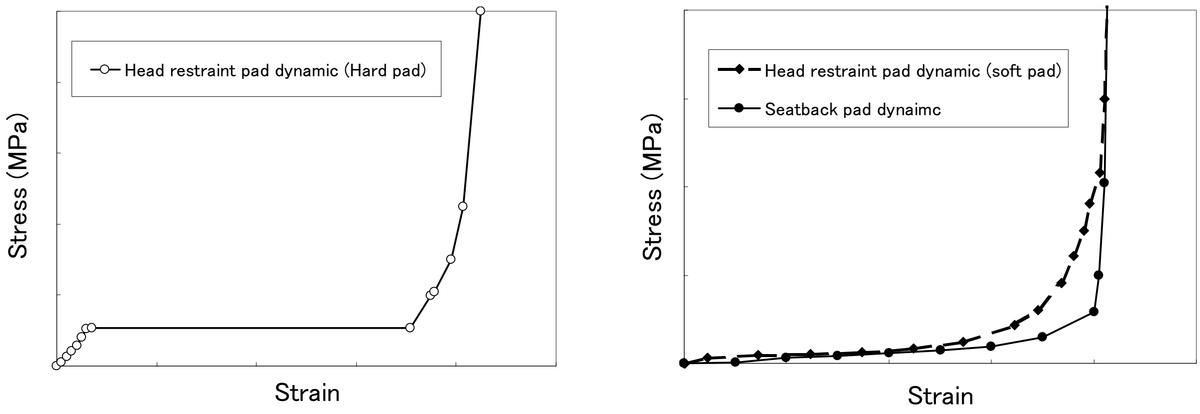


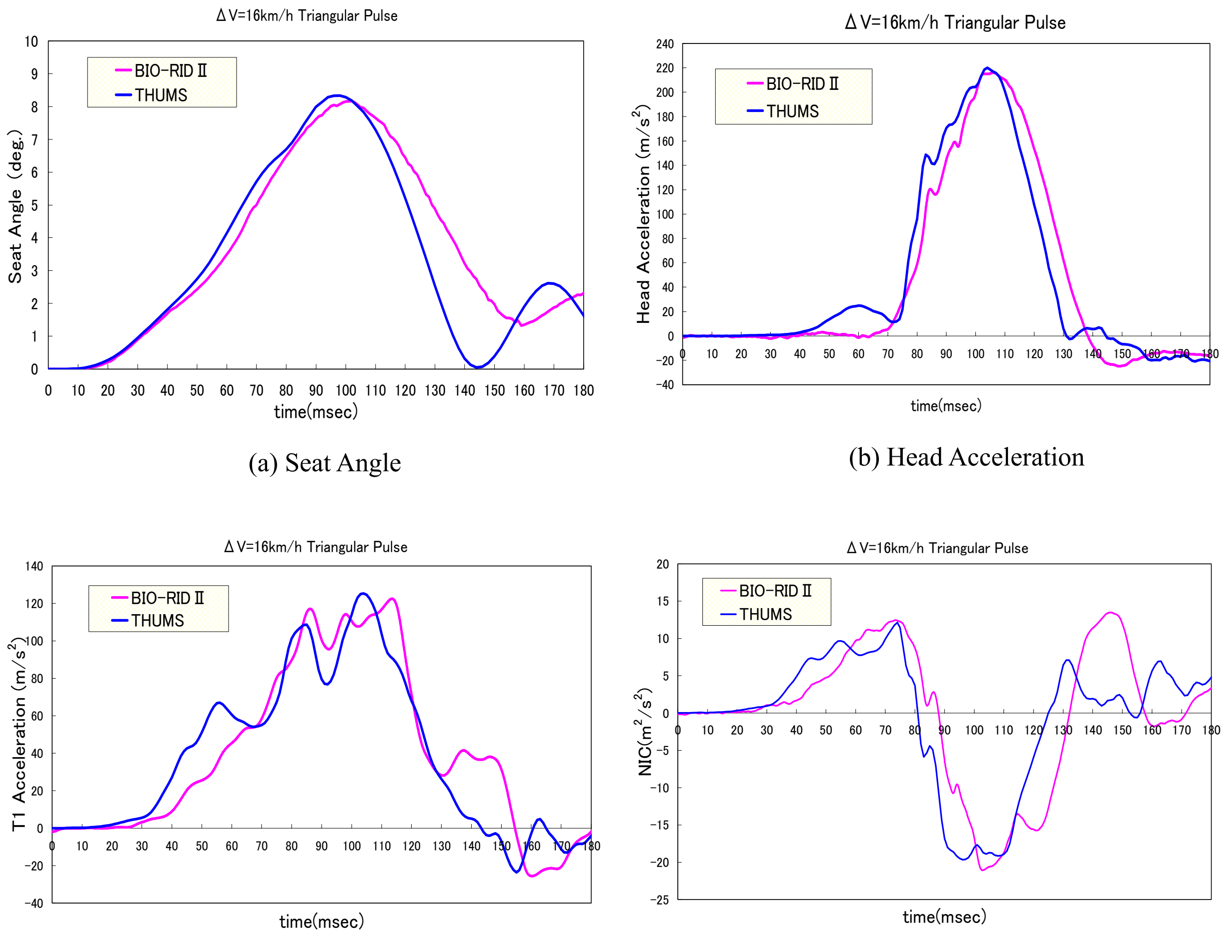
Fig.17 Seat back strength characteristic.



(a) Head restraint pad strength characteristic

(b) Head restraint pad and seatback pad strength characteristic

Fig.18 Head restraint and Seat back pad strength characteristic.



(a) Seat Angle

(b) Head Acceleration

(c) T1 Acceleration

(d) NIC

Fig.19 Correspondence verification of experiment and FE analysis.

Before examination, we first made a comparison of the response of each portion between the computer models and the BIO-RID II dummy under the same conditions to verify the prediction accuracy of the models. Figure 19 shows a comparison when Seat B (the new seat) was used and pulse 1 (ΔV = 16 km/h triangular pulse) was input. With regard to the seat angle representing seat deformation, the calculation results obtained from the THUMS v1.63 and the experimental results using the BIO-RID II dummy matched well, as shown in Figure 19 (a). Additionally, the acceleration pulses and peak levels of head G and thorax G (T1G) depicting the motions of the occupant's head and neck almost matched, as shown in Figures 19 (b) and (c). A high level of consistency was also obtained for the calculated NIC, as shown in Figure 19 (d). Although we could not make any comparison with data on human subjects in this examination, the THUMS v1.63 and the seat models are considered to be highly accurate in predicting the motions of the head and the neck in rear-end collisions because their overall motions matched well with the BIO-RID II dummy, which is regarded as having high biofidelity.

4.2 Results of FE analyses

Figure 20 shows the motions of the head and the neck and the strain distribution of the neck soft tissues at 106 ms after a collision. The seat B model was used for calculation, and pulse 2 (ΔV = 25 km/h triangular pulse) was input. Concerning the strain of the neck joint capsule (the soft tissue of the cervical vertebral joints) that is said to correlate with neck whiplash in rear-end collisions, it reached high at C5-C6 and C6-C7 vertebral (red area) and showed a similar tendency to the results of the volunteer tests performed by Sekizuka [15] and those of the human FE analyses conducted by Hasegawa [13].

Figure 21 shows the results of comparing neck soft tissue strain for the three input pulses between the conventional seat (Seat A) and the new seat (Seat B). The ratio of calculated strain output to the generally proposed criterion [21] was assigned to the vertical axis in the figure. The strain level was lower for the new seat than for the conventional seat for every input impulse.

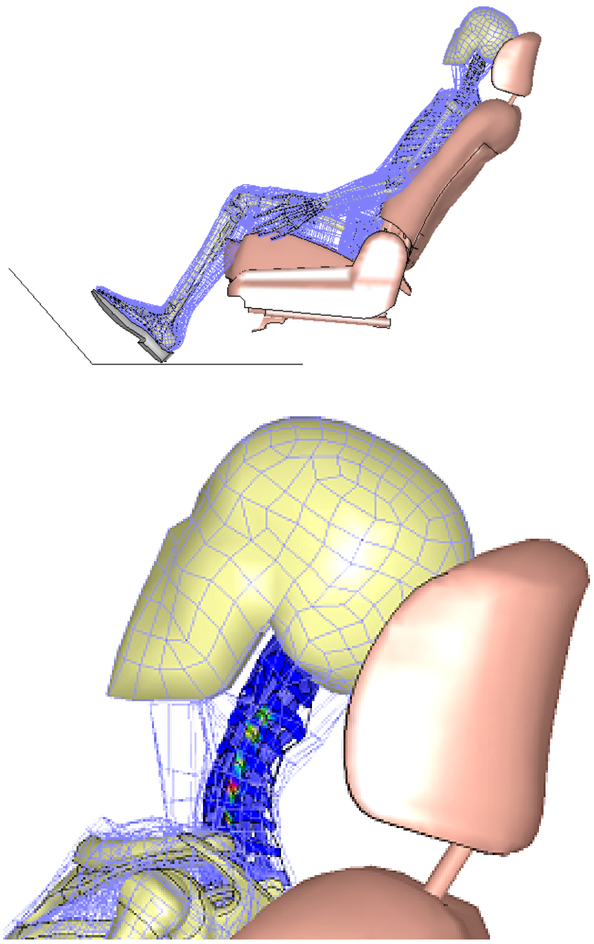


Fig.20 Example of strain distribution by FE analysis（106ms）.
(Seat B, pulse 2： ΔV=25km/h Triangular pulse)

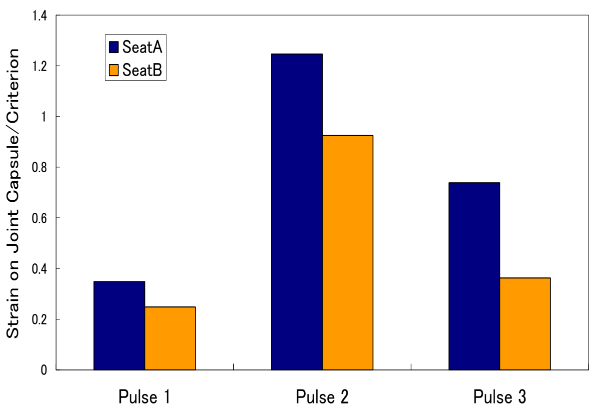


Fig.21 Strain on Joint Capsule / Criterion examination

5. CONCLUSIONS

1. We have developed a seat using the design concept of simultaneously restraining the head and thorax while preventing head rotation. The design employs higher strength seat components and an updated component layout.

2. The newly developed seat is able to simultaneously restrain the head and thorax, while minimizng the extension of the head, as intended. Measurements of simulated $T1G_{HRC}$, $T1V_{HRC}$, $T1S_{HRC}$, and NIC were used to confirm simultaneous restraint of the head and thorax; the NDC was used to measure head rotation; the LNL was used to measure load on the lower neck, and lower My_{max} was also used to ensure that the new seat could reduce whiplash injury.

3. There is a great difference in injury value even between the triangular pulse and the trapezoidal pulse at the same ΔV. In general, when a car suffers a rear-end collision G, the collision acceleration pulse seems to depend on the bullet vehicle and the wrap rate. The triangular pulse is considered to be effective in dealing more severe collision conditions.

4. In rear-end collision simulations using THUMS-AM50 OCCUPANT Ver.1.63-050304, the strain distribution of the neck soft tissues was larger in the lower neck, as observed in experiments. We also investigated the strain of the neck joint capsule, and confirmed that the strain level was lower for the new seat than for the conventional seat. This result proves that our new design is effective in reducing load on the neck. The seat development concept we proposed was proven to the effective by experiments and FE analyses.

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IMPROVED SEAT AND HEAD RESTRAINT EVALUATIONS

Marcy Edwards
Sarah Smith
David S. Zuby
Adrian K. Lund
Insurance Institute for Highway Safety
United States
Paper Number 05-0374

ABSTRACT

Since 1995 the Insurance Institute for Highway Safety (IIHS) has measured and evaluated the static geometry of head restraints on vehicle seats. Geometry is important because a restraint positioned behind and close to the back of an occupant's head is a necessary first step toward reducing neck injury risk in rear crashes. In recent years head restraint geometry in new model passenger vehicles has improved steadily. However, a restraint that does not remain close to the head during a crash cannot effectively support the head and neck, so the effectiveness of a restraint with good static geometry may be reduced by poor dynamic response of a seatback or restraint cushion. In addition, the effectiveness of advanced seat and head restraints designed to move during a crash, either to improve geometry or reduce torso accelerations, can be evaluated only in dynamic tests. Thus, good geometry is necessary but, by itself, not sufficient for optimum protection. Dynamic evaluations using a test dummy also are needed to assess protection against neck injury in rear crashes.

Several insurance-sponsored organizations formed the International Insurance Whiplash Prevention Group to develop a seat/head restraint evaluation protocol, including a dynamic test. Tests using this protocol produce substantially different results among seat/head restraint combinations, even among those with active head restraints. IIHS published its first set of evaluations using the protocol in fall 2004. This paper describes the rationale behind the protocol and summarizes the results of IIHS testing so far.

INTRODUCTION

The Highway Loss Data Institute (HLDI) estimates that every year insurers pay approximately 1.7 million injury claims for which a neck sprain/strain (i.e. whiplash) is the most serious injury suffered by the claimant (HLDI, 2004). With an average cost of \$4,798 for these claims (Insurance Research Council, 2003), the total cost for crashes that result in nothing more serious than whiplash is \$8.2 billion, and this accounts for 25 percent of all crash injury claims dollars paid by insurers. This suggests a much larger whiplash problem than the federal government estimate of only 800,000 minor neck injuries occurring annually in the United States, of which 270,000 occur in rear crashes (National Highway Traffic Safety Administration (NHTSA), 2004). Bowie and Walz (1995) estimate that the total cost of U.S. whiplash injuries exceeds \$19 billion. These injuries are similarly costly in other countries: CAN\$ 409.7 million in British Columbia, Canada (Dayton, 1996); €2 billion in Germany (Langwieder and Hell, 2001); \$43.5 million in Sweden (Holm, 1996); and £1.6 billion in the United Kingdom (Batchelor, 2001). These substantial economic costs are in addition to the emotional and social costs of the pain and suffering associated with minor neck injury.

Vehicle seats and head restraints have been recognized for more than 35 years as the primary countermeasures against whiplash injuries in rear crashes. In 1969 the U.S. government issued Federal Motor Vehicle Safety Standard (FMVSS) 202 as an initial effort to reduce the number of whiplash injuries (NHTSA, 2001). The standard required that all front outboard seating positions in cars be equipped with head restraints that could be adjusted to at least 700 mm above the seat reference point. In 1991 the standard was extended to cover pickup trucks, sport utility vehicles, and vans. This effort was partly successful, with various evaluations of the regulation estimating a 14-18 percent reduction in neck injuries in rear crashes in cars with head restraints compared with earlier models without them (Kahane, 1982; O'Neill et al., 1972; States et al., 1972). One weakness of the early standard was that it did not set a minimum height requirement for adjustable restraints. Not surprisingly, Kahane (1982) found that fixed restraints, which were no shorter than 700 mm above the seating reference point, were more effective than adjustable ones, which often are left in their lowest adjustment positions.

The current European head restraint standard (UN-ECE Regulation no. 17), which applies to passenger vehicles sold in Europe, addresses the shortcoming of the U.S. standard by specifying a minimum height for all head restraints. It also requires head restraints to be taller. Restraints must be at least 750 mm above the H-point and include at least one adjustment position 800 mm above the H-point (United Nations Economic Commission for Europe, 2002). Recognizing that the current U.S. standard leaves many taller vehicle occupants unprotected, NHTSA proposed to upgrade FMVSS 202 in January 2001. The proposal, which was issued as a new safety standard in December 2004, adopted the same height requirements as ECE regulation 17 and added a backset requirement specifying that a restraint could be no farther than 55 mm behind the head of a dummy representing a 50th percentile male seat occupant (NHTSA, 2004). The new backset requirement reflects the simple physical fact that a restraint must be near the head to help support it early in a crash and accelerate it along with the torso. FMVSS 202a will apply to passenger vehicles built after September 1, 2008.

In an effort to encourage manufacturers to equip their vehicles with seats and head restraints better able to provide rear crash protection to a wider range of vehicle occupants, the Insurance Institute for Highway Safety (IIHS) began rating static head restraint geometry for public information in 1995. The measurement protocol used the Head Restraint Measuring Device (HRMD) developed by the Insurance Corporation of British Columbia (ICBC) to measure the static geometry (height and backset) of vehicle head restraints relative to the head of an average-size male (Gane and Pedder, 1996). Ratings (good, acceptable, marginal, or poor) were based on static geometry (Figure 1) and whether the restraints had locking adjustments. The rating procedure was modified and adopted by the Research Council for Automotive Repairs (RCAR) in 2000 and was the basis for head restraint ratings in Australia, Canada, the Untied States, and the United Kingdom until it was replaced in 2004 by a procedure that includes dynamic tests.

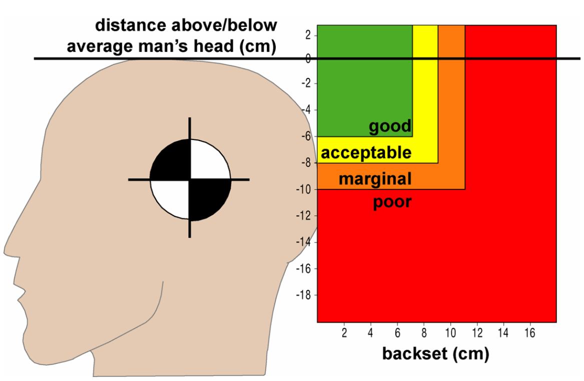

Figure 1. Head restraint geometry ratings

Ten years of IIHS static geometry ratings, combined with the more recent impending upgrade of the U.S. head restraint standard, effectively encouraged automakers to fit the U.S. vehicle fleet with seats and head restraints with better static geometry. As shown in Figure 2, the proportion of cars offering seats with good and acceptable head restraint geometry increased from 7 percent in 1995 to 78 percent in 2004.

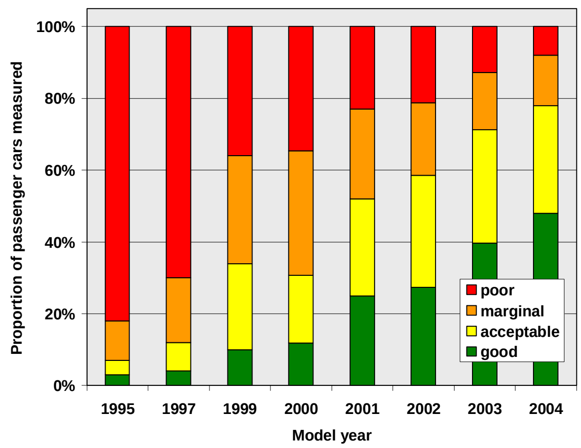

Figure 2. Evaluations of head restraint geometry, passenger cars, 1995-2004

In addition to improving static head restraint geometry, several automakers have developed seats and head restraints with other features intended to reduce whiplash injury risk in rear crashes. These features include yielding seatback cushions with strong perimeter frames (e.g., General Motors' Catcher's Mitt and Toyota's Whiplash Injury Lessening (WIL) system), energy-absorbing seats (e.g., Volvo's Whiplash Injury Prevention System (WHIPS)), and active head restraints. The yielding seatback cushion and energy-absorbing designs control the movement of an occupant's torso to reduce the stresses on the neck until the restraint can contact the head. Active head restraints include a mechanism to move the restraint closer to the head during a crash so it can help support the head earlier than a restraint that does not move. Studies have shown that several of these seat/head restraint designs are effective in reducing neck injury rates in rear crashes (Farmer et al, 2003; Jakobsson and Norin, 2004; Viano and Olsen, 2001).

Head restraints with better static geometry have been shown to reduce the risk and severity of neck injuries in rear crashes (Chapline et al., 2000; Farmer et al., 1999; Olsson et al., 1990). However, as the following example shows, not all restraints initially close to the head provide the same level of support for the head and neck in a rear crash.

Two seats from modern vehicles, the 2002 Ford Windstar and 2003 Pontiac Grand Am, were positioned so the static geometry of the restraints relative to a BioRID's head was similar (Figure 3). The seat/head restraints then were subjected to the same simulated rear crash; two tests were conducted with each design.

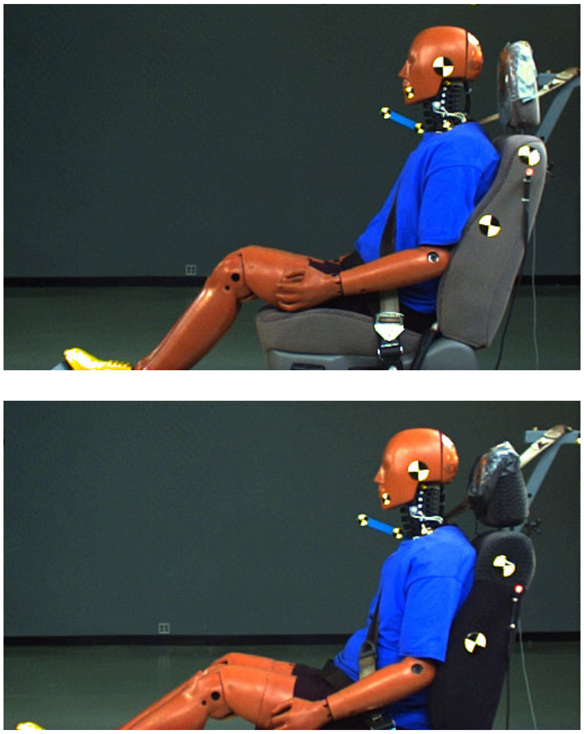

Figure 3. Photos (at T=0 ms) from tests of 2002 Ford Windstar (top) and 2003 Pontiac Grand Am (bottom)

Results indicated that the restraint in the Grand Am contacted the dummy's head earlier in the crash and provided better support than the restraint in the Windstar (Table 1). Figures 4 and 5 illustrate the two main reasons the Windstar seat and head restraint failed to provide the same level of support to the dummy's head and neck. First, although the restraints initially had the same backset, the head restraint in the Grand Am contacted the dummy's head at 60 ms into the crash, whereas the restraint in the Windstar did not contact the dummy's head until 40 ms later; rearward deflection of the Windstar's seatback kept the restraint from reaching the dummy's head sooner. Second, when the Windstar's restraint did contact the

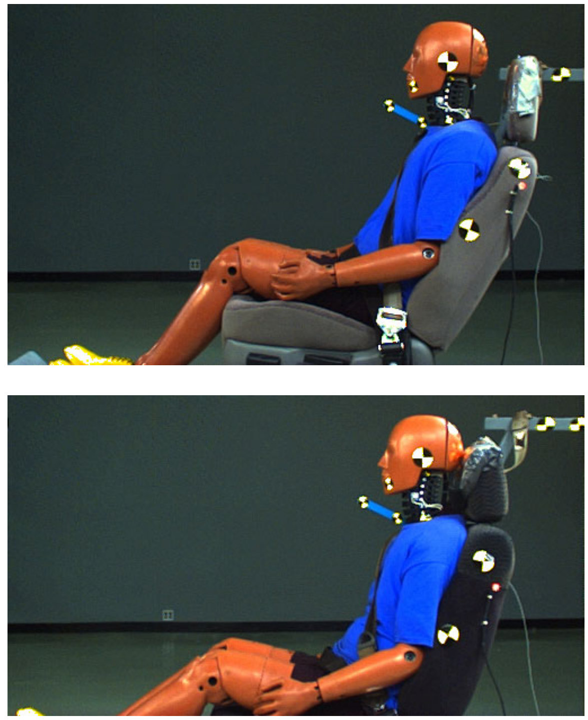

Figure 4. Photos (at T=60 ms) comparing seat movements in tests of 2002 Ford Windstar (top) and 2003 Pontiac Grand Am (bottom)

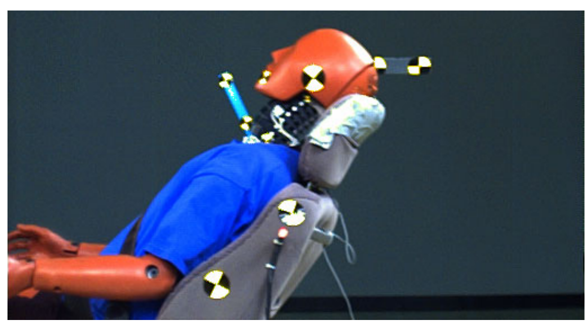

Figure 5. Photo (at T=168 ms) of head restraint contact showing compression of restraint from force of head, 2002 Ford Windstar

Table 1.
Comparison of 2002 Ford Windstar and 2003 Pontiac Grand Am seats tested with same dummy-to-head-restraint geometry

	2002 Ford Windstar		2003 Pontiac Grand Am	
	Test 1	Test 2	Test 1	Test 2
Time to head restraint contact (ms)	107	106	59	57
Upper neck shear force (N)	359	387	217	230
Upper neck tension force (N)	1084	1217	719	123
Neck injury criterion*	31	33	18	16

*Boström et al. (1996)

dummy's head, it offered little support because it was too soft. Thus, even if vehicle seats and restraints are required to meet more stringent geometric requirements, the level of whiplash protection will vary depending on other factors. The increasing proportion of new vehicle seats with good/acceptable static head restraint geometry and the advent of other whiplash protection features suggested a need for dynamic tests of seats to establish which designs are better able to provide beneficial support for occupants' heads and necks in rear crashes.

IIWPG SEAT/HEAD RESTRAINT EVALUATION

IIHS worked with the International Insurance Whiplash Prevention Group (IIWPG), formed in December 2000, to develop a vehicle seat and head restraint evaluation that included dynamic tests. IIWPG is comprised of research and testing organizations sponsored by automobile insurers, including Thatcham in the United Kingdom; Allianz Centre for Technology in Germany and the German Insurance Institute for Traffic Engineering; Folksam Insurance in Sweden; ICBC in Canada; Insurance Australia Group; and CESVIMap in Spain. The specific aims of the member groups vary, but their common objective is to use standardized testing of vehicle seats to encourage automakers to equip vehicles with seats that could help reduce whiplash injuries. The work of IIWPG included conducting many tests and considering all of the available research concerning whiplash injuries. The seat evaluation procedure adopted by IIHS reflects these efforts.

The IIWPG/IIHS evaluation procedure begins with an assessment of static geometry. The basic geometric requirements for seat and head restraint design, height and backset, are measured to produce a rating of good, acceptable, marginal, or poor, based solely on the adequacy of the restraint to accommodate large segments of the population. This rating procedure is detailed in the RCAR (2001) publication, "Procedure for Evaluating Motor Vehicle Head Restraints." Although the RCAR procedure assigns a good evaluation to all active head restraints, the IIWPG/IIHS static evaluation reflects the same measurement criteria as for nonactive restraints. The additional benefits of active head restraints, if any, are assessed through dynamic testing. Head restraints with geometric ratings of good or acceptable are tested in a simulated 16 km/h rear impact to determine a dynamic rating of how well they support the torso, neck, and head. The final overall rating of a seat is a combination of its geometric and dynamic ratings. Seat designs with geometric ratings of marginal or poor automatically receive an overall rating of poor. They are not subjected to dynamic testing because their geometry is inadequate to protect anyone taller than an average-size male.

The dynamic test consists of a rear impact using a crash-simulation sled and a BioRID IIg to represent an occupant. A sled test with standard crash pulse (Figure 6) is used rather than a full-vehicle test even though, in theory, full-vehicle test results could include the effect that a vehicle's rear structure might have on seat performance. However, in real-world rear crashes vehicles experience impacts with a wide range of vehicles at a variety of speeds such that seats in rear-struck vehicles will actually experience a wide range of crash pulses. The IIWPG procedure is designed specifically to assess the performance of seats and head restraints, not rear-end structures, the designs of which are driven by many factors other than neck injury prevention.

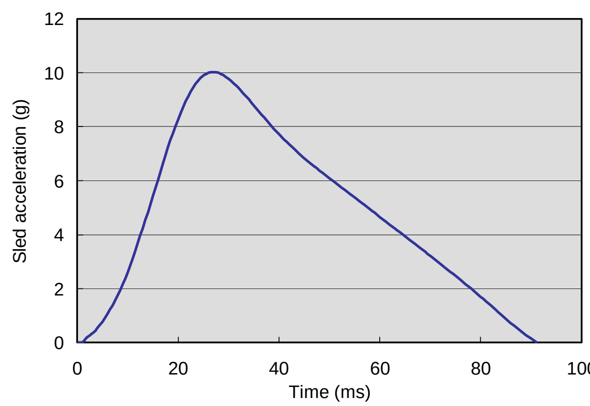

Figure 6. IIWPG sled pulse for dynamic tests of seats and head restraints

The performance criteria for the dynamic test are divided into two groups: two seat design parameters and two test dummy response parameters. The first seat design parameter, time to head restraint contact, requires that the head restraint or seatback contact the occupant's head early in the crash. This follows from the main reason for requiring a small static backset, which is to reduce the time during a rear crash until the head is supported by the restraint. Thus, the time-to-head-restraint-contact parameter ensures that initially good or acceptable static geometry is not made irrelevant by poor seat design. The second seat design parameter, forward acceleration of the seat occupant's torso (T1 acceleration), measures the extent to which the seat absorbs crash energy so that an occupant experiences lower forward acceleration. In some cases, seats designed to absorb crash energy may result in later head restraint contact times. Seats with features that reduce contact time or have effective energy-absorbing characteristics have been shown to reduce neck injury risk in rear crashes compared with seats with reasonably similar static geometry fitted to the same vehicle models (Farmer et al., 2003). The critical values of the seat design parameters have been set consistent with the performance of benchmark seats. The time-to-head-restraint-contact limit of 70 ms reflects head restraint contact times achieved by seats with active head restraint designs and good or acceptable static geometry. The T1 acceleration limit of 9.5 g is based on the maximum T1 accelerations recorded in tests of Volvo's WHIPS seats, which include energy-absorbing/force-limiting seatback hinges. Thus, these seat design parameters should encourage more automakers to adopt design principles that have been shown to be effective in the real world.

The two dummy response parameters, upper neck shear force and upper neck tension force, ensure that earlier head contact or lower torso acceleration actually results in less stress on the neck. The critical values of these neck forces are set according to the distribution of neck forces observed in current seats with good static geometry. The measured neck forces are classified low, moderate, or high depending on which region of Figure 7 the data points lie with respect to maximum neck shear and tension forces. The regions are bounded by curves representing the 30th and 75th percentiles of the joint probability distribution of neck shear and neck tension forces among seats with good geometry tested by IIHS or Thatcham in 2004. Thus the limits for low forces are achievable with current design knowledge.

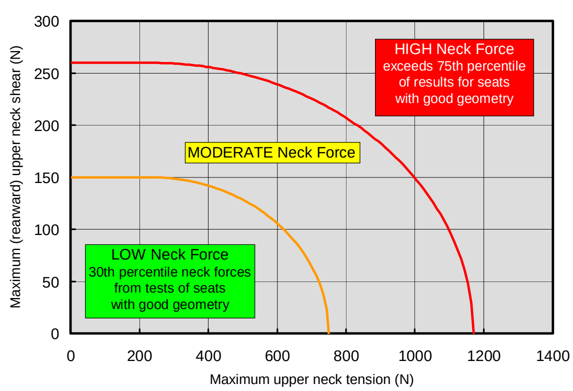

Figure 7. Rating for the joint distribution, maximum neck tension and maximum neck shear

To receive a good dynamic rating, a head restraint must pass at least one of the seat design parameters and also produce low neck forces. If neck forces are moderate or high, then the dynamic rating is only acceptable or marginal. If neck forces are high and neither seat design parameter is passed, then the dynamic rating falls to poor. Table 2 shows how the dynamic rating is determined, and Table 3 illustrates how the geometric and dynamic ratings are combined for an overall evaluation of seat design.

Table 2.
Dynamic rating requirements

Seat Design Criteria	Neck Force Classification	Dynamic Rating
T1 X-acceleration ≤9.5 g OR Time to head restraint contact ≤70 ms	Low	Good
	Moderate	Acceptable
	High	Marginal
T1 X-acceleration >9.5 g AND Time to head restraint contact >70 ms	Low	Acceptable
	Moderate	Marginal
	High	Poor

Table 3.
Formulation of overall rating

Geometric Rating	Dynamic Rating	Overall Rating
Good	Good	Good
	Acceptable	Acceptable
	Marginal	Marginal
	Poor	Poor
Acceptable	Good	Acceptable
	Acceptable	Acceptable
	Marginal	Marginal
	Poor	Poor
Marginal	No dynamic test	Poor
Poor	No dynamic test	Poor

RESULTS OF IIHS FIRST SEAT EVALUATION SERIES

IIHS's first evaluation series included only seats from 2004 and 2005 cars with current IIHS crashworthiness ratings for front or side impacts — a total of 97 seat/head restraint combinations from 79 different vehicle models. Forty-five seats had a static geometry rating of good, 28 were rated acceptable, 12 were marginal, and 12 were poor. Thus, 73 seats qualified for dynamic testing, and the remaining 24 seats received an overall rating of poor. A complete summary of the test results can be found in Appendix A.

Only 15 of the 73 seats tested passed the T1 acceleration criterion of 9.5 g. However, only 5 of these seats also had low neck forces, and they had either energy absorbing (WHIPS) or yielding seatback cushion (WIL) designs. Another 6 seats with low torso acceleration had high neck forces. Four of the seats with high neck forces also were among those with the largest seatback rotations: Ford Crown Victoria and Taurus, Lincoln Town Car, and Mercury Grand Marquis. The other two seats, Acura TL and Lexus GS, had relatively soft head restraint cushions that did not seem to offer enough support even after they contacted the dummy's head. Seats with good static head restraint geometry had lower T1 maximum accelerations on average than seats with acceptable geometry ($p<0.05$) (Table 4).

Table 4.
T1 maximum accelerations related to static geometry

Seat static rating	Good	Acceptable	All tests
Minimum T1 (g)	7.0	8.0	7.0
Maximum T1 (g)	16.2	17.0	17
Average T1 (g)	10.9	12.0	11.2

Eleven of the 73 seats tested passed the time-to-head-restraint-contact criterion of 70 ms. All but 2 of these seats were equipped with active head restraints. However, there also were 6 seats equipped with active head restraints that did not pass this criterion. None of the seats that passed produced high neck forces. Again, seats with good static head restraint geometry had significantly lower head restraint contact times on average than seats with acceptable geometry ($p<0.001$). The two nonactive seats that passed this criterion had good head restraint geometry (Table 5).

Table 5.
Head restraint contact times related to static geometry

Seat static rating	Good	Acceptable	All tests
Minimum time (ms)	53	64	53
Maximum time (ms)	126	133	133
Average time (ms)	84	100	92

The evaluation protocol takes into account that energy-absorbing seats are beneficial and that some designs may have delayed head restraint contact times. Results of this first series of seat evaluations indicate that seats meeting the T1 acceleration criterion had later head restraint contact times on average ($p<0.08$).

Thirteen of the seats tested produced low neck forces, 24 seats produced moderate neck forces, and the remaining 36 seats produced high forces. Of the 13 seats that produced low neck forces, 9 also passed either the T1 acceleration or head restraint contact time criteria. Three of the other 4 seats nearly passed one of the seat design criteria, with results just over the limit. Of the 13 seats with low neck forces, 12 had good static head restraint geometry. Both neck shear force and neck tension force were lower for seats with good static head restraint geometry ($p<0.001$) (Table 6).

Table 6.
Maximum upper neck forces related to static geometry

Seat static rating	Good	Acceptable	All tests
Minimum shear (N)	11	22	11
Maximum shear (N)	299	427	427
Average shear (N)	139	238	178
Minimum tension (N)	287	630	287
Maximum tension (N)	1365	1571	1571
Average tension (N)	750	1050	867

Of the 73 seats IIHS tested dynamically, only 8 earned an overall rating of good. Of the remaining 65 seats 16 were rated acceptable, 19 were marginal, and 30 were poor. Of the 8 seats with a good overall rating, 4 had active head restraints and 4 had energy-absorbing seats like Volvo's WHIPS. One seat with acceptable static head restraint geometry received a good dynamic rating, but its overall rating of acceptable reflects that it cannot be adjusted to protect the tallest seat occupants.

COMPARISON OF IIWPG/IIHS RATINGS WITH OTHER SYSTEMS

Since 2003, the Swedish Road Administration in conjunction with Folksam Insurance and Autoliv has published vehicle seat ratings based solely on dynamic tests. Ratings are derived from three tests at different speed/acceleration levels and from the scoring of three BioRID response parameters: NIC, Nkm, and head-rebound velocity (Krafft et al., 2004). Each of the three tests is assigned 5 points, so the maximum combined rating can be up to 15 points. Each of the three parameters evaluated in the tests is assigned points based on the magnitude of the value measured. The maximum point value assigned to NIC and Nkm for each test is 2, while head-rebound velocity is only assigned a maximum value of 1 point. When the points are combined from all three tests and all three rating parameters, a rating of Green+ (0-2.5 points), Green (2.6-5.0 points), Yellow (5.1-10.0 points) or Red (10.1-15.0 points) is assigned to the vehicle seat. Both the IIWPG/IIHS rating system and the SRA rating system have 4 rating categories; therefore, IIWPG/IIHS good can be compared with SRA Green+ and so on.

The mid-severity test that SRA conducts is similar to the IIWPG 16 km/h test. In order to compare the ratings systems for these tests, IIHS's first series of seat evaluations were scored according to the Swedish system. It was found that seat designs with the lowest point totals were those the IIWPG/IIHS system also rated good. In general, this partial application of the Swedish system to the IIHS test results showed good agreement with IIWPG/IIHS ratings. Seven seats had IIHS/IIWPG overall ratings that were two rating levels different from those suggested by the Swedish system for a single test. For fives models —Saab 9-2x and 9-3, Subaru Impreza, Nissan Altima, and Lincoln LS — the seat rating would have been two rating levels lower using the Swedish system compared with the IIWPG/IIHS procedure. For the other two models, Lexus LS 430 and Hyundai Elantra, seat ratings would have been two rating levels better using the Swedish system compared with the IIWPG/IIHS procedure. Among the 73 seats dynamically tested by IIHS, 6 also have been tested by SRA. All 6 of these seat designs received comparable ratings in both the SRA assessment and the IIWPG/IIHS assessment (Table 7).

Table 7.
SRA vs. IIHS ratings

Make and series	SRA rating	IIHS rating
2003 BMW 3-Series	Red	Poor
2003 Saab 9-3	Green +	Good
2003 Saab 9-5	Green	Acceptable
2003-04 Toyota Corolla	Green	Acceptable
2004 Volvo S40	Green +	Good
2004 Volvo V70/S80	Green +	Good

SUMMARY

Vehicle head restraint geometry has improved in recent years, and forthcoming safety regulations will reinforce these improvements. In addition, some automakers have equipped their vehicles with seats having other features intended to help reduce the risk of whiplash injury in rear crashes, some of which have proven to be effective. Consequently, ratings of vehicle seats for consumer information need to incorporate dynamic testing to differentiate among current seat designs and encourage the greater adoption of designs with additional anti-whiplash benefits. IIWPG has developed a rating system that addresses this need, and IIHS and other IIWPG members have begun publishing vehicle seat ratings using the IIWPG system.

The IIWPG/IIHS system continues to emphasize the importance of static head restraint geometry by dynamically testing only those seats that meet certain geometric requirements. This decision recognizes that many current vehicles still are equipped with head restraints that are not high enough to help accelerate the heads of taller occupants in rear crashes and the fact that many head restraints with sufficient adjustment range cannot be locked into position or are too far behind the head to provide support early in a crash. In addition, government regulation requiring better geometry will not be in full effect for another 4 years. Adequate head restraint geometry and locks for adjustable restraints still are necessary first steps to provide protection against neck injuries in rear crashes.

Despite good or acceptable static geometry, two-thirds of the seats tested by IIHS failed to demonstrate adequate support for the head and neck in a simulated rear crash. These received dynamic ratings of marginal or poor. Thus improvement in dynamic performance is needed. In that regard, it is encouraging that 23 of the seats with good or acceptable dynamic ratings did not have special features such as active head restraints or energy-absorbing seatbacks. These results indicate that a good overall rating probably can be achieved without the addition of the more expensive special features if the static geometry is sufficiently good. However, the best rated seats in IIHS's initial series of tests were those equipped with some variation of the special features, which have been shown to be effective in real crashes.

As interest in minor neck injuries increases, other seat evaluation systems have appeared. A comparison of the IIWPG/IIHS system with that used in Sweden suggested that the two systems reward the same seat design strategies.

ACKNOWLEDGMENT

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APPENDIX A. Ratings and technical data for vehicle seat/head restraints

			Geometry			Seat design parameters			Neck forces				
Make and model	Model years	Seat type	Rating	Backset (mm)	Distance below top of head (mm)	Pass/Fail	Max. T1 accel. (g)	Head contact time (ms)	Force rating	Max. neck shear force (N)	Max. neck tension (N)	Dynamic rating	Overall rating
Acura TL	2004-05	All Seats	Good	45	40	Pass	9.5	106	High	250	882		
Acura TSX	2004-05	All seats	Good	60	52	Fail	10.5	105	High	175	1253	Poor	Poor
Audi A4	2004-05	Seats that adjust manually	Good	42	60	Fail	12.9	88	High	247	905	Poor	Poor
Audi A4	2004-05	Seats with power adjustment	Acceptable	52	62	Fail	12.9	92	High	296	1075	Poor	Poor
Audi S4	2004-05	All seats	Good	62	50	Fail	11.2	91	High	248	911	Poor	Poor
Audi A6	2005	All seats AHR	Good	62	52	Pass	12.8	66	Moderate	183	674	Acceptable	Acceptable
BMW 3-Series	2002-05	All seats	Acceptable	68	80	Fail	11.3	102	High	309	988	Poor	Poor
BMW 5-Series	2004-05	Base seats	Acceptable	75	12	Fail	14.0	103	High	230	750	Poor	Poor
BMW 5-Series	2004-05	Sport/comfort seats with AHR	Good	60	15	Pass	9.9	64	Moderate	95	761	Acceptable	Acceptable
Cadillac CTS	2003-05	Seats without adjust-able lumbar	Acceptable	82	58	Fail	10.5	100	High	351	1148	Poor	Poor
Chevrolet Malibu	2004-05	All seats	Good	28	52	Fail	12.1	71	Low	14	732	Acceptable	Acceptable
Chrysler Sebring	2003-05	Seats with power recline	Good	45	52	Pass	8.6	116	Moderate	188	636	Acceptable	Acceptable
Chrysler 300	2005	All seats	Good	20	52	Pass	12.0	53	Moderate	79	787	Acceptable	Acceptable
Dodge Neon	2001-05	Seats with adjustable head restraints	Acceptable	65	68	Fail	12.6	118	High	256	1298	Poor	Poor
Dodge Stratus	2003-05	Without lumbar	Good	55	58	Fail	11.5	113	High	262	696	Poor	Poor
Dodge Stratus	2003-05	With lumbar	Acceptable	48	62	Pass	8.8	117	Moderate	173	630	Acceptable	Acceptable
Ford Focus	2001-05	All seats	Good	58	40	Fail	12.8	98	Moderate	97	923	Marginal	Marginal
Ford Taurus	2004-05	All seats	Acceptable	60	75	Pass	9.3	116	High	263	675	Marginal	Marginal
Ford Crown Victoria	2003-05	Seats with adjustable lumbar	Good	55	60	Pass	8.4	126	High	238	724	Marginal	Marginal
Honda Civic	2003-05	All seats	Good	62	45	Fail	12.5	99	High	299	1149	Poor	Poor
Honda Accord	2003-05	EX models standard seats	Good	55	45	Fail	11.3	88	High	45	1365	Poor	Poor
Honda Accord	2003-05	LX models standard seats	Acceptable	72	50	Fail	13.6	93	High	81	1168	Poor	Poor
Hyundai Elantra	2001-05	All seats	Acceptable	78	62	Fail	14.3	110	High	427	1490	Poor	Poor
Hyundai Sonata	2001-05	GL models	Acceptable	80	68	Fail	9.9	104	High	161	975	Poor	Poor
Hyundai XG350	2002-05	All seats	Acceptable	72	65	Fail	13.6	74	High	314	1235	Poor	Poor
Infiniti I35	2002-04	All seats AHR	Acceptable	85	70	Fail	16.5	77	High	399	1571	Poor	Poor

			Geometry			Seat design parameters			Neck forces				
Make and model	Model years	Seat type	Rating	Backset (mm)	Distance below top of head (mm)	Pass/Fail	Max. T1 accel. (g)	Head contact time (ms)	Force rating	Max. neck shear force (N)	Max. neck tension (N)	Dynamic rating	Overall rating
Infiniti G35	2005	All seats AHR	Acceptable	28	62	Fail	9.7	105	High	220	782	Poor	Poor
Infiniti Q45	2005	All seats AHR	Good	48	52	Fail	9.9	101	Moderate	191	864	Marginal	Marginal
Jaguar X-type	2004-05	All seats	Good	42	35	Fail	11.0	110	High	292	749	Poor	Poor
Jaguar S-type	2005	All seats	Good	38	48	Pass	8.5	75	Low	70	369	Good	Good
Kia Spectra	2005	All seats AHR	Good	68	40	Fail	10.1	87	Low	89	609	Acceptable	Acceptable
Kia Optima	2001-05	Seats that adjust manually	Acceptable	68	68	Fail	13.3	87	High	235	1394	Poor	Poor
Kia Amanti	2005	All seats AHR	Good	65	53	Fail	9.6	77	Low	105	588	Acceptable	Acceptable
Lexus IS	2001-05	All seats	Good	42	45	Fail	12.0	78	Moderate	79	934	Marginal	Marginal
Lexus ES	2004-05	All seats	Acceptable	75	68	Fail	10.9	102	High	179	1041	Poor	Poor
Lexus GS	2003-05	All seats	Good	52	52	Pass	9.5	96	High	237	1009	Marginal	Marginal
Lexus LS	2001-05	All seats	Good	40	25	Fail	11.1	94	Moderate	96	839	Marginal	Marginal
Lincoln LS	2003-05	All seats	Good	68	15	Pass	8.4	75	Moderate	184	636	Acceptable	Acceptable
Lincoln Town Car	2003-05	All seats	Acceptable	80	68	Pass	7.8	134	High	271	859	Marginal	Marginal
Mazda 3	2004-05	Base seats	Good	42	58	Fail	11.9	86	Moderate	116	984	Marginal	Marginal
Mazda 3	2004-05	Seats with adjustable lumbar support	Acceptable	48	65	Fail	10.4	104	Moderate	138	839	Marginal	Marginal
Mazda 6	2003-05	Seats without adjustable lumbar	Good	50	58	Fail	10.7	98	Moderate	151	966	Marginal	Marginal
Mazda 6	2003-05	Seats with adjustable lumbar	Acceptable	35	68	Fail	11.1	104	High	192	1178	Poor	Poor
Mercedes C class	2004-05	Seats with auto-adjust head restraints	Good	45	-15	Fail	12.1	74	Moderate	193	836	Marginal	Marginal
Mercedes E class	2004-05	Seats with auto-adjust head restraints	Good	30	5	Pass	11.8	63	Moderate	126	813	Acceptable	Acceptable
Mercury Sable	2004-05	All seats	Acceptable	52	65	Fail	10.7	118	High	377	1220	Poor	Poor
Mercury Grand Marquis	2003-05	All seats	Acceptable	72	65	Pass	8.7	125	High	228	735	Marginal	Marginal
Mini Cooper	2002-05	All seats	Good	32	32	Fail	11.0	88	Moderate	175	653	Marginal	Marginal
Mitsubishi Lancer	2002-05	All seats	Good	60	48	Fail	11.9	82	Moderate	186	773	Marginal	Marginal
Mitsubishi Galant	2004-05	Cloth seats	Acceptable	85	75	Fail	13.4	100	High	280	887	Poor	Poor
Nissan Altima	2005	All seats AHR	Acceptable	80	60	Pass	9.7	64	Moderate	221	660	Acceptable	Acceptable
Saab 9-2X	2005	All seats AHR	Good	38	48	Pass	10.2	62	Low	29	403	Good	Good
Saab 9-3	2005	All seats AHR	Good	60	33	Pass	16.2	64	Low	11	287	Good	Good
Saab 9-5	2005	All seats AHR	Good	45	50	Fail	11.0	72	Low	12	337	Acceptable	Acceptable
Saturn ION	2003-04	Cloth seats	Good	48	55	Fail	11.0	88	High	252	601	Poor	Poor
Saturn ION	2003-04	Leather seats	Acceptable	68	75	Fail	10.8	84	High	238	672	Poor	Poor
Subaru Impreza	2005	All seats AHR	Good	40	53	Pass	10.9	65	Low	43	462	Good	Good

			Geometry			Seat design parameters			Neck forces				
Make and model	Model years	Seat type	Rating	Backset (mm)	Distance below top of head (mm)	Pass/Fail	Max. T1 accel. (g)	Head contact time (ms)	Force rating	Max. neck shear force (N)	Max. neck tension (N)	Dynamic rating	Overall rating
Subaru Impreza WRX	2004-05	All seats	Acceptable	75	45	Fail	9.9	101	Moderate	137	678	Marginal	Marginal
Subaru Legacy	2005	All seats AHR	Good	50	58	Pass	10.4	67	Moderate	80	725	Acceptable	Acceptable
Subaru Outback	2005	All seats AHR	Good	50	58	Pass	10.4	67	Moderate	80	725	Acceptable	Acceptable
Suzuki Aerio	2002-04	All seats	Good	62	32	Fail	12.3	85	Moderate	164	745	Marginal	Marginal
Suzuki Forenza	2004	All seats	Acceptable	75	72	Fail	14.3	94	High	340	1168	Poor	Poor
Suzuki Verona	2004	All seats	Acceptable	60	75	Fail	12.4	102	High	268	1138	Poor	Poor
Toyota Corolla	2003-04	All seats	Acceptable	40	62	Pass	8.5	91	Low	22	686	Good	Acceptable
Toyota Corolla	2005	All seats	Acceptable	38	65	Fail	15.3	85	High	97	1333	Poor	Poor
Toyota Camry	2004?	Cloth seats	Good	60	60	Fail	13.4	96	Moderate	115	1073	Marginal	Marginal
Toyota Camry	2004?	Leather seats	Acceptable	42	65	Fail	11.8	106	High	178	1166	Poor	Poor
Toyota Avalon	2001-04	All seats	Acceptable	68	65	Fail	15.0	98	High	130	1408	Poor	Poor
Volkswagen New Beetle	2004-05	Seats without adjustable lumbar	Good	35	52	Pass	10.8	69	Moderate	128	579	Acceptable	Acceptable
Volkswagen New Beetle	2004-05	Seats with adjustable lumbar	Good	42	48	Pass	8.9	69	Low	88	530	Good	Good
Volvo S40	2004-05	All seats	Good	15	35	Pass	8.3	84	Low	46	583	Good	Good
Volvo S60	2004	All seats	Good	40	30	Pass	7.0	92	Low	56	663	Good	Good
Volvo S80	2005	All seats	Good	15	55	Pass	9.0	87	Low	37	550	Good	Good
Seats Not Tested													
Acura RL	2001-04	All seats	Poor	68	108							Not tested	Poor
BMW 5 series	2004-05	Seats with adjustable thigh support	Poor	108	18							Not tested	Poor
Buick Century	2001-05	Cloth seats	Marginal	60	100							Not tested	Poor
Buick Century	2001-05	Leather seats	Poor	90	130							Not tested	Poor
Buick Regal	2001-04	All seats	Poor	95	125							Not tested	Poor
Buick LeSabre	2003-05	All seats (AHR)	Poor	75	130							Not tested	Poor
Buick Park Avenue	2003-05	All seats	Poor	135	110							Not tested	Poor
Cadillac Seville	2001-04	All seats	Poor	110	105							Not tested	Poor
Chevrolet Cavalier	2001-05	All seats	Poor	125	135							Not tested	Poor
Chevrolet Classic	2003-05	Cloth seats	Marginal	60	90							Not tested	Poor
Chevrolet Classic	2003-05	Leather seats	Poor	65	105							Not tested	Poor
Chevrolet Impala	2001-05	Cloth bucket seats	Poor	90	130							Not tested	Poor
Chevrolet Impala	2001-05	Leather bucket seats	Marginal	60	90							Not tested	Poor
Chrysler Sebring	2003-05	Seats that recline manually	Marginal	98	58							Not tested	Poor
Honda Civic Hybrid	2003-05	Base seats	Marginal	92	75							Not tested	Poor

			Geometry			Seat design parameters			Neck forces				
Make and model	Model years	Seat type	Rating	Backset (mm)	Distance below top of head (mm)	Pass/Fail	Max. T1 accel. (g)	Head contact time (ms)	Force rating	Max. neck shear force (N)	Max. neck tension (N)	Dynamic rating	Overall rating
Mitsubishi Galant	2004-05	Leather seats	Marginal	70	85							Not tested	Poor
Nissan Sentra	2002-05	Base seats	Marginal	72	95							Not tested	Poor
Nissan Maxima	2004	Cloth seats (AHR)	Poor	130	75							Not tested	Poor
Nissan Maxima	2004	Leather seats (AHR)	Marginal	98	72							Not tested	Poor
Pontiac Grand Am	2001-05	Cloth seats	Marginal	70	83							Not tested	Poor
Pontiac Grand Prix	2004	All seats	Marginal	98	82							Not tested	Poor
Pontiac Bonneville	2003-05	Bench seats	Poor	135	115							Not tested	Poor
Pontaic Bonneville	2003-05	Leather seats	Marginal	105	100							Not tested	Poor
Saturn L series	2001-05	All seats	Poor	80	125							Not tested	Poor
Volkswagen Passat	2001-05	All seats	Marginal	100	58							Not tested	Poor

FULLY-ADAPTIVE SEATBELTS FOR FRONTAL COLLLISIONS

Timothy J. Paulitz
Donald M. Blackketter
Karl K. Rink
National Institute for Advanced Transportation Technology
Department of Mechanical Engineering
University of Idaho
United States
Paper Number 05-0127

ABSTRACT

The goal of this paper was to demonstrate the potential for a fully adaptive restraint system to significantly reduce injuries. To accomplish this, a three-bodied model of a 50th percentile Anthropometric Test Dummy (ATD) in a 35 mph frontal collision was made using Lagrangian Dynamics. The model was verified against test data obtained from NHTSA. Viscoelastic and constant force seatbelt models were created, and the results were compared for a 1998 Chevy Malibu. The simulation accurately reproduced the shape and magnitude of pelvis, chest, and head accelerations. The constant force seatbelt reduced pelvis, chest, and head accelerations by 56%, 62%, and 63%, respectively. The peak lap belt force was reduced by 60%. Relative head rotation was reduced by 16 degrees. A simple control concept was explored and demonstrated the feasibility of an adaptive constant force restraint system. Such restraint systems can make large reductions to risk of injury by significantly reducing forces and accelerations on the occupant.

INTRODUCTION

In 2003, there were over 2.8 million people injured and 32000 killed in over 6.3 million motor vehicle accidents in the US. The restraint system is intended to reduce the risk of these injuries and deaths, but the seatbelt alone only reduces risk of injury by 45-50% for front seat occupants.[18] Air bags, pretensioners and load limiting devices have been introduced to remove slack and better couple the occupant to the vehicle during ride down, resulting in reduced seatbelt forces being exerted on the occupant. While these devices are beneficial, most are not adaptive to each crash and occupant.[17] Therefore it is believed that an adaptive seatbelt and restraint system, using sensor data collected before and during a crash, could potentially reduce injuries by tuning the restraint for each occupant, vehicle, and crash severity.[1, 12]

The primary goal of this work was to demonstrate the potential for an adaptive restraint system to reduce occupant injuries by showing that a constant force seatbelt can reduce forces and accelerations on an occupant. This was shown in the case of a frontal collision of a passenger car into a fixed-rigid barrier at 35 mph with no offset. It is assumed that an adaptive system can be designed and controlled to provide an optimal force level as determined prior to the start of the crash. There was no physical testing carried out to verify these results. Simple analyses were used to contribute to the work and concepts that are necessary for further development of this technology. The concepts developed here can be expanded to an entire adaptive system that would include the air bag. This work does not promote the elimination of air bags, but rather that an adaptive system offers many improvements over current seatbelt restraints. Controlling the seatbelt is preferred because it already contacts a restrained occupant at the beginning of the crash. Therefore, a dynamical analysis was carried out to determine to what extent seatbelt design affects injuries, as inferred from seatbelt forces and occupant accelerations. A simple control example is also shown which demonstrates the basic ideas formed here, and allows basic conclusions on adaptive restraints to be drawn.

BACKGROUND INFORMATION

There are many methods and approaches used in the literature to simulate occupants in a variety of crash scenarios. Most occupant models reflect the mass and geometry of a 50th percentile male ATD since it is used most frequently in crash tests.[9] It is necessary to note that even though an ATD is classified into a percentile, rarely is an actual occupant classified under the same percentile in all body characteristics. Happee[8] et al have noted that through the use of scaled crash-dummy models it is possible to simulate crash events for occupants with anthropometry not currently represented in available ATDs. The assumption of scalability can allow conclusions reached through simulation to be extended to occupant anthropometry not investigated.

It was found that two-dimensional models could adequately simulate restraint systems, and vehicle and occupant response without delving too deeply into the complexities that arise from three-dimensional analyses. This is especially true when computation capabilities are limited or large-scale simulations are unwarranted.[7, 22]

Many mathematical models of the human body emphasize the main body components[13], which are treated as an articulated assembly of rigid bodies defined to realistically represent the geometry and response of the occupant and vehicle.[22, 23]

Functional mathematical models of the seatbelt[13, 23] are preferred because they better model the viscoelastic nature of seatbelts, and they are less restrictive than mechanical spring-mass-damper models[23, 25].[3]

There are two methods to achieve a constant restraint force. The first method is to alter the force-deflection characteristics of the seatbelt such that the force remains constant while the belt elongates, resulting in a seatbelt having elastic-perfectly plastic load-elongation properties.[22] The second method to attain a constant restraint force is to physically control the seatbelt with an actuator so that the load remains constant. The actuator would quickly react in a crash to pretension the belt to some predetermined value, at which time would actively control how much webbing is released or collected to maintain that force. The constant force seatbelt considered here will assume the controlled constant force, although without concern to how it is attained or maintained.

Miller[15, 16] shows that variable load limiting would produce significant improvements in injury by tailoring to the needs of the occupant based on anthropometry. However, the results do not strongly emphasize the benefits of continuously variable load limiting. Here ‘continuously’ variable is used to draw a difference between systems with infinite settings and those systems with only two or three fixed levels[4]. Overall, a majority of the published work on constant force restraints relies heavily on repeated full-scale crash tests and focuses more on a discrete approach to occupant protection using existing technology.[1, 15] For this reason, this work focused on a simulation-based constant force restraint that is assumed to be infinitely variable and is provided by some adaptive and controllable system.

The most applicable work done in the area of adaptive restraint systems is by Hesseling. Hesseling[9, 10] uses an optimal control method to find a controllable restraint system to optimize the chest and head accelerations of a 50th percentile ATD. The important difference between work by Hesseling and that by Miller is that the seatbelt tension and air bag vent diameter are completely controllable, rather than fixed at some optimal level that is parametrically determined.

This paper addresses several conclusions and recommendations made by Hesseling. Hesseling asserts, “…a manageable model that describes the relevant aspects of the dummy, the restraint system, the vehicle and their interactions with an acceptable accuracy would be…useful.”[9] In this work, a simplified model was created that can reproduce key aspects of the system, and it was used it to demonstrate the benefits of an adaptive constant force restraint. Control systems were not the focus of this work. However, some basic conclusions on the matter will be discussed.

The most applicable work done in the area of constant force restraints is done by Crandall.[5] Crandall uses optimization to find an optimal restraint force to minimize the Thoracic Injury Criteria. The optimal force is similar to the controlled constant force model discussed previously. Like Hesseling, Crandall’s work shows that the optimal restraint force may not be constant, and requires an initial pulse that then reduces to lower values. The restraint force found by Crandall is close in shape to a constant force restraint; if used, results in a near optimal response of the chest acceleration. The goal of this work was not to derive an optimal solution, but rather focus on the established work that shows that a constant force restraint (CFR) offers drastic performance improvements in frontal collisions over currently available systems.[17]

Adaptive restraint systems will be closely tied to pre-crash detection.[12, 24, 27] A typical progression of a crash is depicted in Figure 1. With current technology, the occupant is not well restrained until the pretensioner tightens the belt roughly 15 ms into the crash. This time delay is due to a combination of effects from sensing the crash, to the pretensioner response time. In this example describing pre-crash technology, the pretensioning could begin at t=-15 ms, so that the belt is already taut and pretensioned as determined by the information collected by the sensors before crash initiation at t=0. The benefit of integrating adaptive restraint systems with pre-crash technology is twofold. The adaptive restraint can actively control the response of the occupant by removing slack and better coupling the occupant to the vehicle before the crash begins, and it also reduces the magnitude of the required restraint force.

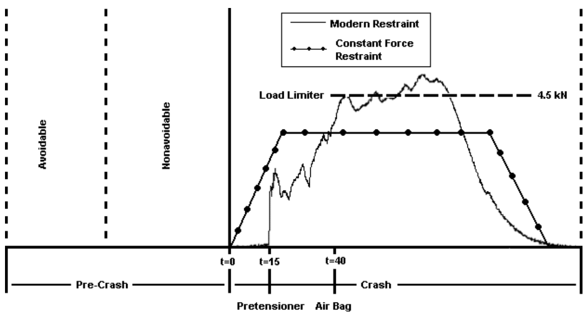

Figure 1: Crash event comparison of an adaptive restraint system with pre-crash sensors to current restraint systems.

There are a few load limiter designs on the market that can adapt to the crash environment, but do so only at discrete, predetermined levels.[4] Load limiters that can offer continuously variable levels of restraint have been developed, at least in theory, at some automotive companies. Takata[11] and Delphi Technologies, Inc[2] currently hold patents for load limiting devices that utilize magnetorheological (MR) fluid dampers to provide a continuously variable level of resistance. The variable resistance is typically obtained by allowing the fluid to flow through ports where electromagnets control the fluid's viscosity.

In 2003, TRW's Active Control Retractor became the first commercially available active seatbelt system. This system, utilizes pre-crash information from braking and stability control sensors to pretension the seatbelt with an electric motor. Takata also holds a patent for a variable pretensioning and load limiting system. Takata's adaptive system utilizes an electric motor for pretensioning and a MR-fluid load limiter, and together it could potentially provide the adaptive restraint necessary to fully realize the benefits of a constant force restraint. With the exception to those discussed above, it should be noted that a majority of this technology only appears in vague descriptions in the literature and patents. This author has no knowledge of physical systems existing, nor does there appear to be much testing or computer modeling of adaptive seatbelt restraint systems.

It is unclear at this point in time what form an adaptive restraint system will take when commercially available. With this in mind, the control for the adaptive CFR modeled in this paper was considered a 'black box' actuator. A model of how the adaptive control device responds can be added once it is available, and can be used to develop control laws. The 'black box' method also lends it to an open-ended design and simulation tool for restraint systems and controllers, without making any assumptions on the form of future technology. This simulation will be a useful tool to researchers to verify control strategies and develop alternative future restraint systems.

METHODS

The occupant was modeled with three rigid bodies and derived with Lagrangian Dynamics. The viscoelastic seatbelt was modeled as a system governed by a multivariable polynomial. The constant force seatbelt was modeled as a linear, piece-wise continuous, function of time. The vehicle was modeled as a deformable body governed by impact dynamics with a rigid barrier. The motion of the occupant and vehicle were found, but no interference between the two was included. A MATLAB program was written to simulate the system in a 35 mph frontal collision. This model was verified against real test data obtained from NHTSA.

Four primary assumptions were used in the model. It was assumed that the restraint forces act in only the direction of motion of the vehicle. This was done because of variance in seatbelt geometry between different vehicles. Also, assuming the forces only act in the horizontal directions reduced the complexity of the Equations of Motion (EOM), saving time in the derivation and solving of the equations. The occupant was modeled with rigid bodies, eliminating chest compression and torso bending. Friction was ignored since the coefficient of friction between body and seat will vary with each vehicle and model of ATD, giving conservative results.[7] All secondary collisions with air bag, dashboard, steering wheel, and seat interactions were ignored.[1] The model simulates the occupant-seatbelt interaction, and the adaptive restraint systems will be such that interior collisions do not occur, thus the need for modeling interior collisions is nonexistent. The locations of the dashboard, steering wheel, and air bag were tracked using the vehicle deceleration model and used as a reference when determining if a collision would occur.

Lagrangian Dynamics is a more favorable approach than Newtonian Dynamics for three main reasons. The first reason is that it provides a better understanding of the dynamics of the system because the system's kinetic and potential energies are used to derive the EOM. The second reason is that through the use of generalized coordinates, the number of equations that must be solved is reduced. The third reason is that it eliminates the need for including forces of constraint because they do no work on the system. The constraint forces include reactions forces from the seat and the joints of the body.

Thus the general approach to determining the EOM for the system is to formulate the Lagrangian *L* by deriving the kinetic energy *T*, potential energy *U*, and external forces *Q* in generalized coordinates *q*, where $L(\tilde{q},\dot{\tilde{q}},t) = T(\tilde{q},\dot{\tilde{q}},t) - U(\tilde{q},t)$ and $\tilde{Q}(\tilde{q},\dot{\tilde{q}},t)$.[19] The EOM can be determined with equation 1 and solved. Several iterations of one and two-bodied occupant models were used to arrive at the final three-bodied model used in this paper, shown in Figure 2, and Tables 1-3. Note that in Table 1 the mass of each model component is found by the product of the mass of the occupant by the percent mass of that component. Also note, that the total mass of the modeled body is not the total mass of the occupant, since it is assumed that a portion of the mass from the arms and lower legs is ignorable.[22]

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}_i}\right) - \left(\frac{\partial L}{\partial q_i}\right) = Q \qquad (1).$$

Table 1:

Mass and inertial values for a three-bodied occupant model, see Figure 2

Mass [N]	% Masses			Radii of Gyration k [m]	
Total	Pelvis	Torso	Head	Torso	Head
766.43	.2983	.5148	.07777	.2840	.1464

Table 2:

Size dimensions [m] for a three-bodied occupant model

Pelvis	Chest		Head	
Length	Height	Width	Height	Width
.59	.508	.254	.254	.2032

Table 3:

Joint Parameters for a three-bodied occupant model

	Hip	Neck		
		Low	ϕ_{limit} [Deg]	High
Spring [N m]	100	7	±30	70
Damper [N m s]	10	10	NA	

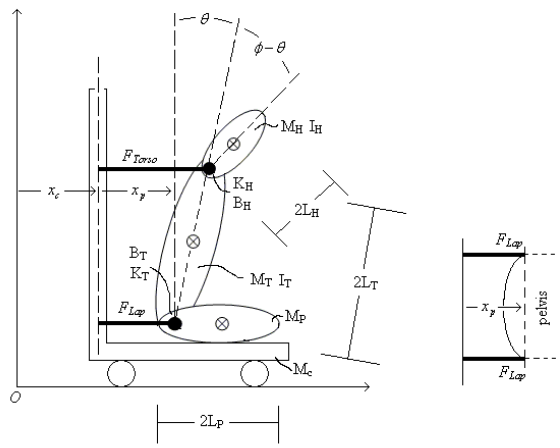

Figure 2: Three-bodied occupant model restrained by a torso and lap belt, dimensioned as shown. Figure on right shows geometry of lap belt on pelvis.

The general approach taken by these authors is as given, and more information can be found by referring to work by Paulitz.[20, 21] First a dynamic model of a 50th percentile ATD approximation was done using the model above with a function-based mathematical seatbelt model. This model was piecewise continuous in force-displacement space. While the belt is elongating, the force-displacement relation is given by a multivariable polynomial cubic in elongation and linear in rate of elongation. While the belt is relaxing the relation is given by a quadratic in elongation only. This model has been shown to closely match the behavior seen in testing.[13, 14] The coefficients for the elongating portion were determined by parametrically altering the values until good correlation was seen between the model seatbelt force and body accelerations and the data obtained from NHTSA. The coefficients for the relaxing portion were determined by assumptions of continuity and some inherent belt properties. This resulted in a functional model of the viscoelastic seatbelt that is believed to be accurate, at least in relation to the assumptions and cases studied.

These results were then compared to those when the viscoelastic seatbelt is replaced by a constant force seatbelt. This was modeled as a constant with a linear ramp done over a period of 10 ms, although any non-zero time can be used. The magnitude of the CFR is determined by parametrically varying the values until the occupant excursion is maximized while interior collisions are prevented. The CFR makes no assumptions about how the force is created or maintained, so the force can be prescribed simply as a function of time rather than finding some relation in the state variables. This model was used to illustrate the possible benefits of a truly constant force restraint that could be provided by some kind of controllable and adaptive system.

The last part of the study involved combining the two models. The viscoelastic restraint model (as earlier determined) was used but is attached to some actuator rather than connected directly to the vehicle. The response of the actuator is then determined such that the restraint force exerted on the occupant is that of the CFR (as earlier determined). This provides insight into the behavior and control issues of such a model, as well some physical characteristics that may need to be considered of the seatbelt and actuator systems.

Results

The simulation results showed that the collision model with a viscoelastic seatbelt is accurate and correctly simulates the system dynamics. The results were found by comparing the simulation to the available test data from similar crash tests. The results shown are for the model verified against a test of a 1998 Chevy Malibu. The vehicle deceleration model was adjusted to the test data and values of k = 1.2, t_f = 105 ms, and V_0 = 15.728 m s^{-1} were found. The acceleration profile of the vehicle is shown in Figure 5; note the data plotted does not contain every data point from the test. Briefly, the vehicle deceleration model is a parabolic equation whose coefficients are determined by the placement of the zeros, $t=0,t_f$, and the total change in velocity kV_0 where k is assumed to be between 1 and 2.[20, 21] The simulation results for the 1998 Chevy Malibu were obtained with the seatbelt properties given in Tables 4-5 and Figure 4.

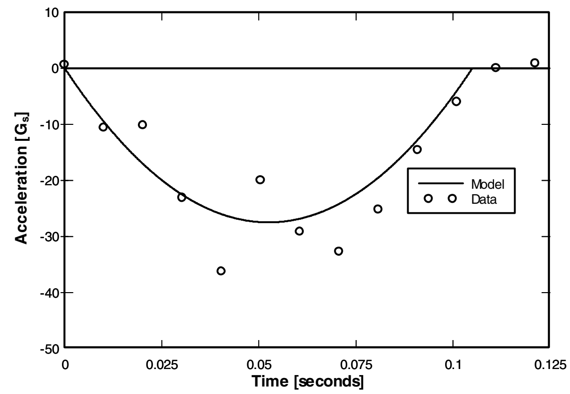

Figure 3: Vehicle A(t) as compared to accelerometer data of a 1998 Chevy Malibu (O) in a 35 mph frontal crash.

Table 4:

Viscoelastic seatbelt coefficients for a 1998 Chevy Malibu

	Cubic [kN m^{-3}]	Quadratic [kN m^{-2}]	Linear [N m^{-1}]	Damping [N s m^{-1}]
Lap	300	200	30	600
Torso	200	115	20	600

Table 5:

Constant force seatbelt parameters for a 1998 Chevy Malibu

	Force F [N]	Tension Time T_t [ms]
Lap	3800	11
Torso	4200	10

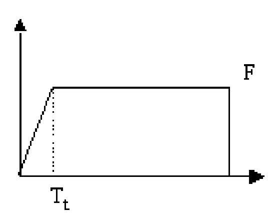

Figure 4: CFR profile, as determined by magnitude F and rise time T_t.

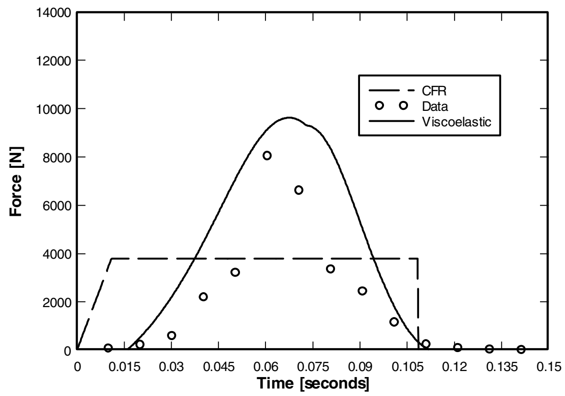

Figure 5: Lap belt F(t) comparisons between viscoelastic and constant force seatbelt models, as compared to data of a 1998 Chevy Malibu (O).

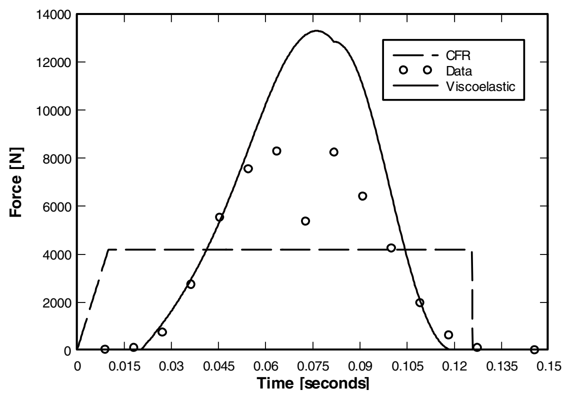

Figure 6: Torso belt F(t) comparisons between viscoelastic and constant force seatbelt models, as compared to data of a 1998 Chevy Malibu (O).

The seatbelt force and body acceleration results for both restraint models are shown in Figures 5-9. The dashed line denotes the results for the constant force restraint (CFR), and the solid line denotes the results for the viscoelastic restraint. The open circles denote the corresponding measured data for the crash test. In general, the results for the viscoelastic model of the lap belt are conservative, where higher belt forces and accelerations are usually predicted. It is clear that the constant force seatbelt can result in lower restraint forces exerted on the occupant, and drastic reductions in pelvis, chest, and head accelerations. A more thorough discussion is given in [20, 21].

Some additional comments, the peak value of the simulated viscoelastic torso belt force in Figure 6 is highly conservative. The shape is similar during increasing load, until 50 ms into the crash when it is presumed the stitch-tearing load limiting is occurring. In this vehicle, load limiting is achieved through tearing of stitches in the seatbelt. This could be because the restraint model stretches and yields in a continuous fashion, while in real-life this occurs discretely. The continuous model was used for simplicity and because the conservative belt forces did not appear to result in large deviances in the accelerations, possibly because while the simulated torso belt force is over estimated, it is accounting for restraint that would be offered by the air bag.

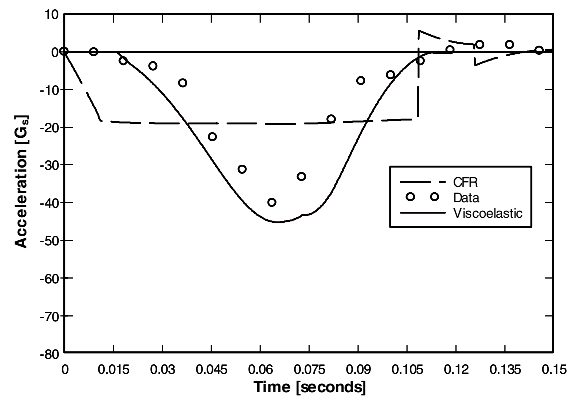

Figure 7: Pelvis A(t) comparisons between viscoelastic and constant force seatbelt models, as compared to data of a 1998 Chevy Malibu (O).

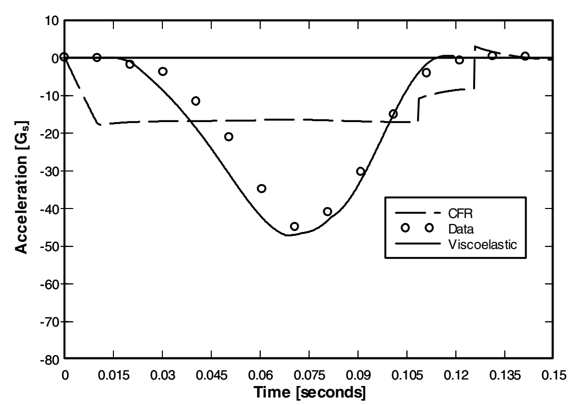

Figure 8: Chest A(t) comparisons between viscoelastic and constant force seatbelt models, as compared to data of a 1998 Chevy Malibu (O).

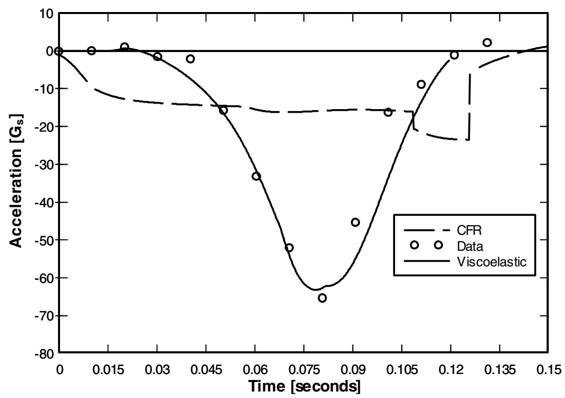

Figure 9: Head A(t) comparisons between viscoelastic and constant force seatbelt models, as compared to data of a 1998 Chevy Malibu (O).

The only negative aspects found of the CFR results are the discontinuities in acceleration, as seen in Figures 7-9. The largest discontinuity seen is roughly 24 G in magnitude, but is not critical because the pelvis is less prone to injury due to large changes in acceleration than the head and neck. Although the sudden change in acceleration subjected to the head and neck may prove injurious. Mathematically, the discontinuities in the accelerations result directly from the discontinuities in the constant force profiles for the torso and lap belts. Piecewise continuity in the constant force profile will remove the discontinuities and yield smoother results. This could be accomplished by giving the $F_{CFR}(t)$ a trapezoidal shape rather than a step shape at the end time. For this reason, the affects of the discontinuities will be for the most part ignored since later simulations could be corrected to reduce their effect.

Perhaps one of the greatest benefits of CFR systems is the possible reduction in head injuries. The general response of the head acceleration for both restraint models differs from those of the pelvis and chest because it is free to respond and is not directly coupled to the vehicle through the restraint. The CFR results in a 62.8 % reduction of head acceleration, even when considering the spike of 23 G. This large reduction could greatly reduce the risk of head injury. The HIC value was not calculated, but it is clear in comparing the shapes of the head accelerations resulting from two seatbelt models that there is a large difference in the area under each curve, inferring large HIC reductions.

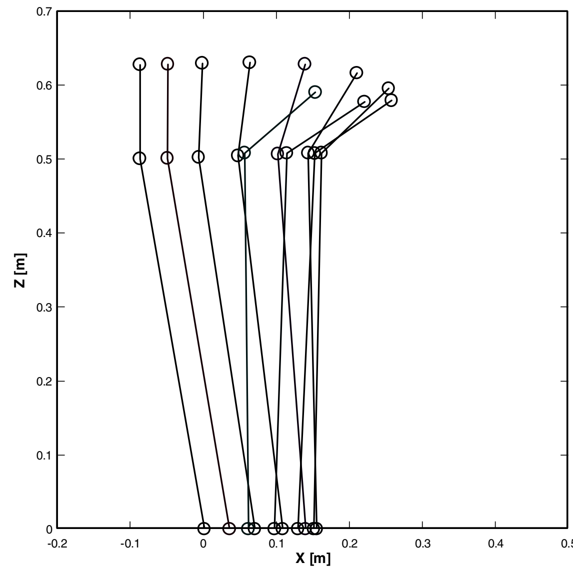

Figure 10: Relative motion of pelvis, torso and head within 1998 Chevy Malibu using a viscoelastic seatbelt model.

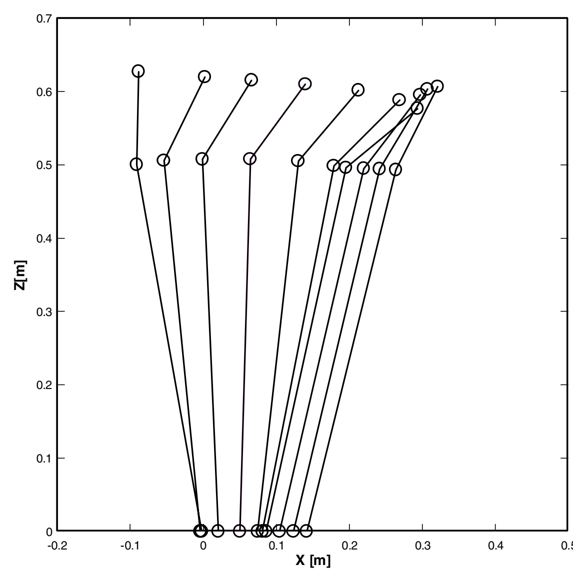

Figure 11: Relative motion of pelvis, torso and head within 1998 Chevy Malibu using a constant force seatbelt model.

The physical motion of the occupant with-respect-to to the Malibu is shown in Figures 10-11. The lines define the centerlines of the torso and head. The markers define the motion of the hip, shoulder, and head/neck cg. Figure 10 illustrates the motion under the viscoelastic seatbelt model, and seems to match with the observed motion during the collision. The body typically translates and rotates forward to a near vertical position. The head rotates forward and the largest relative rotation is seen after the maximum chest excursion. The curve the head Cg follows in space is similar in shape to that shown in the literature [6, 9]. Figure 11 illustrates the motion of the body under the CFR model, where occupant excursion is increased substantially. This result was anticipated as it was assumed that the constant force restraints would increase excursion, making use of more of the available space in the compartment. The relative motion of the head is improved as well, with a dip in the motion of the head as the occupant nears maximum excursion, and is likely caused by the discontinuity in the constant force seatbelt model.

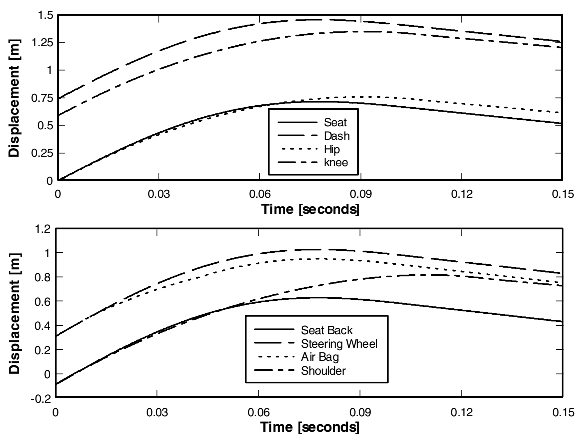

Figure 12: Global motion of pelvis and torso with 1998 Chevy Malibu using a viscoelastic seatbelt model. Top figure is thigh motion with seat and dashboard; bottom figure is shoulder motion with seat, steering wheel, and air bag.

Figure 12 illustrates the global motion of the body. The lines denoting the motion of the occupant denote motion of the centerlines, and do not account for chest depth. One benefit of the CFR model is that rebound of the occupant after maximum excursion is limited, as shown in Figure 12 where the knee and chest approach but do not cross the lines denoting the dashboard and air bag. The values for the CFR were found such that internal collisions were prevented and the rebound rates of the vehicle and occupant matched on a global level. Large rebound rates could result in neck injuries, and suggests that more than required restraint force was applied and could therefore be reduced.

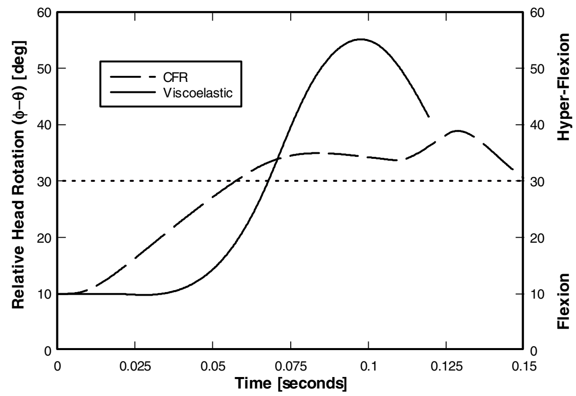

Figure 13: Relative head rotation $\phi(t)-\theta(t)$ comparisons between viscoelastic and constant force seatbelt models of a 1998 Chevy Malibu.

The relative rotation of the head about the torso is given in Figure 13 for both restraint models. The maximum rotation for the viscoelastic restraint model is 55.14 degrees, and for the CFR model it is 38.9 degrees, a 29.5 % reduction. The shape and magnitude are similar to data published of expected head rotations for the human body [6]. One conclusive benefit of the CFR system is that it reduces peak relative head rotation; it also broadens the amount of time the rotation occurs. In the viscoelastic restraint model, the head remains fixed relative to the torso until about 40 ms at which time the head begins to rotate forward at a high rate. For the CFR model the head rotation begins to increase noticeably at about 10 ms, at a much gradual rate. The large reduction in relative rotation into the region of hyper-flexion should greatly reduce injury, as shown on the right hand side of Figure 13.

"Black Box" Controller

Having established a viscoelastic and a constant force seatbelt model, it was then possible to combine them to create a basic control concept for an adaptive constant force restraint. The restraint control was not the focus of this work, rather the simulation of the restraint, so a large amount of effort was not put forth into creating a full-scale feedback-control system to realize a truly adaptive seatbelt.

The general concept is that it is possible to apply a predetermined force to the occupant, without concern to how it was achieved. With this approach, it is possible to obtain some general guidelines of what physical characteristics such a system should have and what kind of response would be expected from the control system. The basic control was found by combining the constant force profile to the viscoelastic seatbelt model that is a function of belt elongation. The device providing the control will be coupled to the body through the seatbelt. So it is believable that an adaptive control scheme would include both models. The controller would act like a pretensioner that could pull and release the belt such that the response to the viscoelastic restraint in a collision is transformed into the response seen under a CFR.

In theory, an adaptive restraint system might look like that shown in Figure 14. The viscoelastic belt models are rewritten so that rather then being only functions of occupant displacement with respect to the vehicle, they are functions of belt elongation *d*. Here $d=x-y$ where x is the occupant motion with respect to the seat, and y is the motion provided by the actuator which is considered controllable; no constraints were placed on the magnitude of y or $\dot{y}$ at this time. To determine the $y(t)$ output for the controller, the constant force seatbelt model is equated to the viscoelastic seatbelt model with the coefficients determined previously, as shown by equation 2. Equation 2 can be included in to the CFR simulation, and $y(t)$ can be found where $F_{CFR}(t)$ and $x(t)$ are determined through simulation as discussed previously.

$$F_{CFR}(t) = F_{VISCO}(x-y) \qquad (2).$$

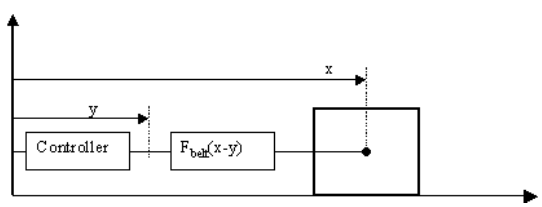

Figure 14: Basic adaptive seatbelt controller schematic. *X* denotes the displacement of the occupant; *y* denotes the displacement done by the controller such that $F_{belt}(x-y)$ is constant.

Using this method, the response of the controller for an adaptive constant force restraint was found. The response for both belts is shown in Figure 15 and 16. Here, the response of the control will be referred to as the Active Force Response, where the pretensioning device actively controls $y(t)$ to maintain a constant force while the body is moving. This device and the seatbelt system as whole will be referred to as an Adaptive Restraint System, where some *a priori* decision is made as to the magnitude of the force required, and the AFR responds accordingly to provide that force based on simulation. Later, feedback would be included that would allow the AFR to respond to changes that occur during the crash. This could include a secondary collision in a multi-vehicle crash, or an over- or under-prediction of occupant motion. In any case the AFR would respond to increase the load or decrease the load to the necessary value to prevent collisions and reduce injury. Such a system should replace current pretensioning and load limiting systems.

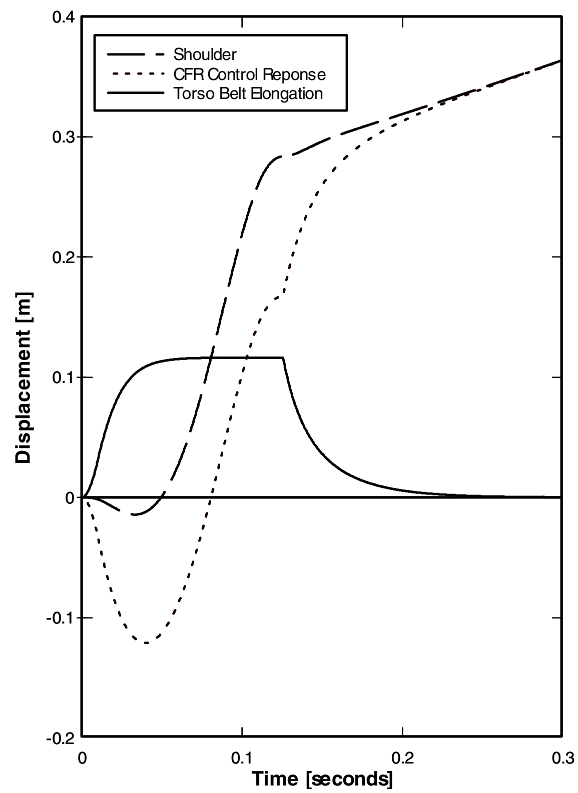

Figure 15: Active Force Response of torso belt for 1998 Chevy Malibu to provide a constant force from a viscoelastic seatbelt model.

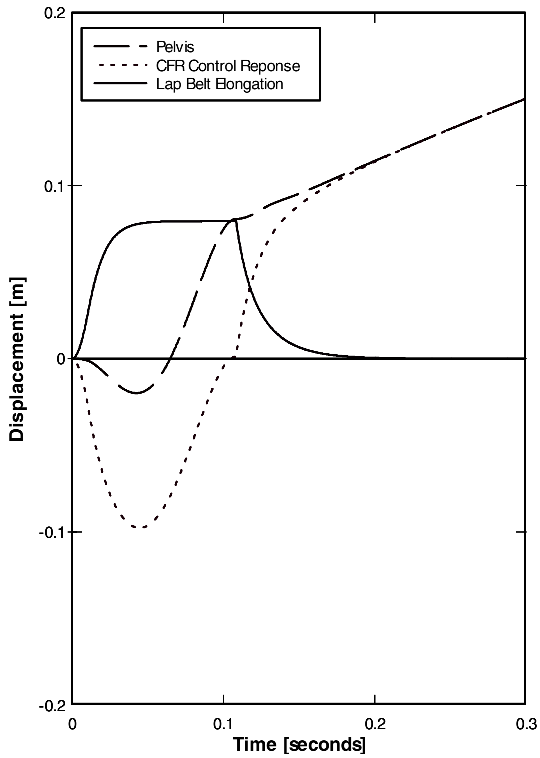

Figure 16: Active Force Response of lap belt for 1998 Chevy Malibu to provide a constant force from a viscoelastic seatbelt model.

The response for the torso belt is shown in Figure 15, and the response for the lap belt is shown in Figure 16. The solid lines denote the belt elongation. It is worth noting that these curves match those expected for the creep and recovery of a viscoelastic material[3]. The portion with negative concavity denotes the loading of the material, and the portion with positive concavity denotes the response once the load has been removed. The extension phase also matches expected behavior of a Voigt-Kelvin material model to a unit force.[3] This suggests that although the mechanical Kelvin/Voigt model was not used for the restraint system, the polynomial function chosen does respond similarly.

The dashed lines denote the motion of the body, where in the torso plot it is the motion of the shoulder, and in the lap plot it is the motion of the pelvis. As the device begins to pull the belt out away from the body to increase the load, the body does move backwards some amount. While this amount is small, it suggests that at least a simple model for a seat back should be included at some future point.

The dotted lines denote the response *y(t)* of the AFR. In general, the shape tends to follow that of the motion of the occupant, although with larger magnitudes. The device would have to be capable of retracting about 5 inches on the torso belt and about 4 inches on the lap belt. An important note here is that no slack was considered for these models, so the retraction capabilities would be this value plus some estimate for a slack value based on occupant position and size. These models also require an elongation of the torso belt on the order of 4.5-5 inches and an elongation of the lap belt on the order of 3 inches. However, this is for the 50^{th} percentile person, so for a smaller occupant less would be necessary, and the bounds for this would have to be established by the largest expected occupant, 95^{th} or higher percentile.

This model assumed that the seatbelt itself is performing all the restraint, and may be limited by the performance of the webbing in maximum allowable elongation. The allowable elongation will depend on the length of the seatbelt itself, which was not considered. If good estimates of belt length and size were available along with the test data used to obtain this model, then a better idea of the limitations for the load control could be found. This may be found to be limited on either the amount of webbing the retractor can actually reel in or the amount of strain the belt can withstand. Although, in comparing the amounts of elongation between the viscoelastic and adaptive models, this does not appear to be an issue at present. If the limiting factor is found to be a material property of the belt, it may be possible that realizing a restraint of this design could require new and advanced polymers for use in seatbelts specifically tailored for use in adaptive seatbelts.

Another solution could be the use of the air bag to provide supplemental restraint. If the seatbelt 'knew' how much restraint the air bag would offer for a particular occupant and crash, it could tailor the seatbelt force with this in mind such that together they offer the restraint necessary for preventing injury rather than relying solely on the seatbelt itself. These cases will all have to be accounted for in the design of the controller, since there will be certain bounds based on physical and mechanical limitations of the system.

Conclusions

The simulation considering a viscoelastic restraint can reproduce test data from certain vehicles with acceptable accuracy. Good correlation was seen between the pelvis, chest, and head accelerations and satisfactory correlation was seen between the lap and torso belt forces. Drastic reductions in seatbelt force and occupant accelerations were achieved through the use of a controlled constant restraint force profile. This restraint was defined to closely relate to modern load limiting and pretensioning technology but do so in an improved fashion. These improvements warrant the further investigation of controlled and adaptive constant force seatbelts.

The adaptive restraint system outlined here controls the belt elongation to maintain the restraint force at a constant level as the occupant moves through the vehicle. Such a system would adapt to the crash environment by making control decisions on restraint force based upon occupant size, weight, and location, as well as crash severity. This forms the basic theory for an adaptive restraint system. An adaptive restraint system would consist of a controllable seatbelt and air bag that are in constant communication with one-another and sensor data to make control decisions that will drastically reduce occupant injury. This work has shown that an adaptive seatbelt will provide sufficient restraint and could greatly reduce risk of injury, especially HIC, in the cases studied. These systems suggest the potential for great benefits over conventional and constant force restraints achieved through non-adaptive and mechanical load limiting. It also has been shown that a means to control the seatbelt force is attainable, and a basic control strategy and concept has been proposed. While this is shown in the case of a mid-sized male occupant, it is believed that these benefits will apply to other occupant types. This will hold especially for those that are more susceptible to injury from high belt forces and occupant accelerations, such as children, women[26], and the elderly.

Recommendations

For larger occupants and in high severity crashes it is unlikely that the seatbelt alone will be able to protect the occupant from injuries. For these cases the air bag will still play an important role in injury prevention and reduction, so to extend the analysis of adaptive and constant force seatbelts to higher crash severities, it will be necessary to incorporate a model of the air bag, at least to some degree. Future analysis should also include the air bag to determine the possibility of timing the air bag inflation to provide additional head restraint later in the crash and help reduce the peak acceleration experienced by the head, and also the development of adaptive restraint systems that would include active seatbelts and air bags.

This work only considered frontal impacts of 35 mph, and it would be valuable to see the results for offset frontal, rear-end, and side-impact crashes using the restraint system proposed here. In these cases, it is not expected that the CFR will have as drastic improvements on injury, however no modeling of these cases was done here nor does there appear to be much done the literature in regards to constant force restraints in non-frontal collisions. Creating individual simulations of these cases could prove useful as well in determining improved restraint systems since it may be true that the best force with which to restrain an occupant in each crash orientation, i.e. frontal, rear, side, etc. may be different. However one physical device may be capable of creating the necessary restraint force. For a frontal collision, the adaptive restraint system would apply a constant force, but for a rear- or side-impact it could apply the necessary restraint to greatly reduce injury, as determined through simulations of these other cases.

With the purpose and goal of the simulations and models used and created here in mind, it is the authors' opinion that it is important that the models remain as simple as possible to be of most use. The recommendations pertain to the assumptions used to create these simplistic models. The direct influence of the assumptions made here should be better understood to ensure the soundness and robustness of the model as it exists and to find simple additions to the simulation that could improve its accuracy and usefulness. Once a continuously adaptive restraint system has been created and made available, it is also necessary to perform physical testing to verify these models. In the case that the test data does not correlate well to the predicted response, suitable corrections should be made while attempting to preserve the simplicity that is desired for controller design and feedback.

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DEVELOPMENT OF RESTRAINT SYSTEMS WITH CONSIDERATIONS FOR EQUALITY OF INJURY RISK

Stephen A. Ridella, Matthew J. McCann, Roy C. Turnbull
TRW Automotive, Occupant Safety Systems
Gregory S. Bayley
Automotive Consultant
United States
Paper Number 05-0403

ABSTRACT

Occupant restraint system development continues to evolve as new regulations and consumer demand drive more complex solutions. Traditional seat belt and airbag designs are giving way to more intelligent systems that respond to crash and occupant conditions. In regulated vehicle compliance safety tests, occupant performance is usually judged against injury criteria that differ with respect to occupant size. While for a given test, two different occupant sizes may give results that pass the criteria, their probabilities of injury for a given body region may not be equal. It may be possible to change restraint configurations that not only demonstrate compliance to recognized injury criteria for a given occupant, but additionally demonstrate that for a given crash mode, an equal probability of injury exists for all body regions of a range of adult occupant sizes. This paper will discuss a computer modeling approach devised to analyze a particular vehicle environment and range of occupant sizes. A design of experiments was carried out that adjusted parameters of the restraint system including seat belt pretensioners, load limits, and various airbag components. For each analysis, the probability of injury by body region and occupant were compared to find the set of components that comprise a system to give equal probability of injury for each body region for each occupant. Results of the design of experiments, statistical analysis and impact on restraint system development will be discussed. This paper documents a new approach to restraint system development as it looks beyond specific injury criteria to injury risk comparisons.

INTRODUCTION

Previous Studies on Adaptive Restraints

Adomeit quotes in a previous report "The more loads differ within the range of injury criteria under different test conditions or under real world accident conditions – or even exceed injury criteria in certain circumstances – the more we need active restraint system adjustments related to input parameters: in other words, adaptation of restraint system" (1). These words have motivated a number of studies to explore the adaptability of restraint systems to the occupant and vehicle crash environment. Bendjellal et al(2) described a "programmed restraint system" that incorporated airbag pressure and seatbelt force limiters to reduce occupant injury criteria relative to standard belt/bag systems. Their aim was to reduce thoracic loads induced in occupants for different crash modes. Foret-Bruno et al (3) determined occupant thoracic injury risk by age based on analysis of crashes of vehicles equipped with this programmed restraint system. A 4kN shoulder belt load limit was recommended for all occupants based on this analysis, but made no mention of occupant size. Miller and Maripudi (4) performed a computer modeling study to determine restraint parameters required for 5th percentile female, 50th percentile male, and 95th percentile male dummy models. By adjusting belt load limit and airbag venting properties for these 3 occupants in normally seated positions, they could determine the optimal requirements for those restraint parameters that resulted in the lowest injury criteria for each dummy size. That study, however, did not make any adjustments to the inflator performance during the simulation.

Happee et al (5) showed that by varying occupant size through scaling techniques, outside the standard dummy model sizes, large variations in injury criteria could occur as a result of different seating positions for the same restraint systems. Cuerden et al (6) proposed that a 25-45% reduction of AIS 2 and 3 injuries could be achieved with adaptive restraint systems compared to belted only occupants. His analysis relied on a hypothetical injury reduction matrix applied to set of field injuries with known severity for a given occupant type. Breed (7) hypothesized that airbag inflation rate as well as gas discharge from the airbag could be controlled relative to occupant position and morphology if the ability to determine that position and morphology existed. This follows the Happee study, but no test or model data is given. These early studies suggested the need to have a restraint system that adjusted to the occupant size for a large range of crash conditions and occupant size.

Dummy performance by size

In its efforts to provide regulations aimed at more of the adult population, NHTSA added the 5^{th} percentile female to the passive safety requirement and has proposed adding the same dummy to the belted, 35mph (56kph) NCAP barrier crash that will also phase into the passive safety requirement (8). In its own testing of 18 vehicles with non-adaptive restraint systems, NHTSA has found 6 vehicles that exceeded injury value limits for 5^{th} percentile female drivers in the areas of head, chest, and/or neck regions while the 50^{th} percentile male driver did not exceed criteria. In the other 12 vehicles, it was found that "the overall injury values for the 5^{th} percentile adult female driver dummies in [the tested] vehicles were somewhat higher than the values for the 50^{th} percentile driver dummies tested in the same vehicle (9)." The neck area was usually the highest value difference.

In a more detailed study (10), NHTSA reported results from 5 paired vehicle crash tests where either the 5^{th} female (full forward) or 50^{th} male (mid-track) was the driver and passenger occupant. The results showed that 5^{th} female driver and passengers typically had higher chest acceleration and neck injury criteria (Nij) values than the 50^{th} male driver and passenger in the same vehicle. HIC values did not differ significantly between dummy sizes. Maltese et al (11) ran 35 vehicle tests with mostly unbelted 5^{th} and 50^{th} percentile dummies and saw similar increase neck injury criteria for the 5^{th} percentile female dummy regardless of vehicle type or crash severity.

An Analysis of NCAP Results for 5^{th}/50^{th}

In an effort to better understand the differences between 5^{th} and 50^{th} percentile dummy responses, data from three different driver and seven different passenger NCAP tests or mathematical models was collected relative to 5^{th} or 50^{th} dummy response in the same vehicle sled test or model. NCAP star ratings based on HIC and chest acceleration (Gs) were compared for the same test or model and are shown in Figure 1. In every case, the 5^{th} percentile female chest G's increased relative to the 50^{th} percentile male while HIC exhibited little difference, or in some cases, slightly improved. Chest deflection in the 5^{th} percentile female showed increases in 5 of the passenger and 2 of the driver tests or models, however all values were below the FMVSS 208 injury criteria value for chest deflection (Figure 2).

Taking the analysis further, the injury probabilities for an AIS 3+ chest injury using these chest deflections were compared between 5^{th} and 50^{th} dummy sizes (Figure 3). The scale factors from published data by Mertz et al (12) were used to calculate the probabilities. The comparisons of injury probabilities reveal that the likelihood of an AIS 3+ chest injury (e.g.: multiple rib fractures) was significantly higher for the 5^{th} female in all the cases where the injury criteria was higher. In one case, the risk of chest injury was 3 times higher for a 5^{th} female even though the injury criteria increased by 30% compared to the 50^{th} male occupant.

It is the response to the crash loads among different occupant sizes in a given crash configuration that may need to be addressed with an adaptive restraint system. As the issues of addressing the restraint requirements for the smaller occupant arose as a result of the airbag-induced injury, NHTSA added the small female crash dummy to its passive restraint certification requirements for passenger vehicles. It may not be enough, however, to accept the fact that an injury criteria for a 50^{th} %ile male may translate into a 15% probability of injury, while a 5^{th} %ile female is subject to a 30% probability of injury for the same crash configuration and restraint system.

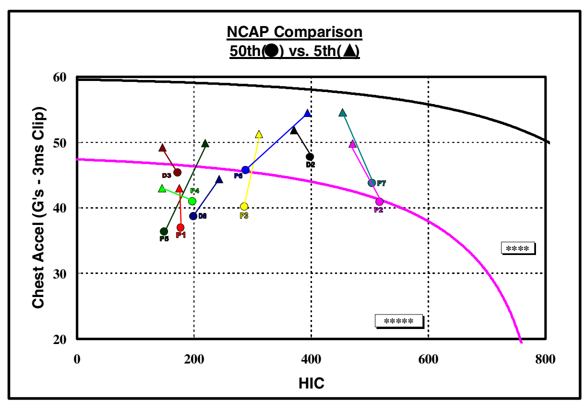

Figure 1. Driver and/or passenger occupant response for 5^{th} %ile female (triangle) and 50^{th} %ile male (circle) for various driver and passenger vehicle restraint systems.

The possibility to equalize the injury probability for the two occupant sizes by body region in a given crash configuration forms the basis for the current study. Identifying restraint system parameters that can be adjusted to the occupant while maintaining a balanced or equal probability of injury and complying with existing injury criteria can only be solved using computer techniques building on the biomechanics data existing in the literature.

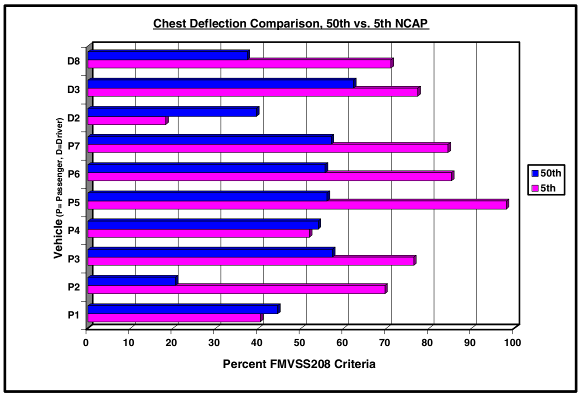

Figure 2. Driver and/or passenger occupant chest deflection response (5th %ile female in pink and 50th %ile male in blue) for various driver and passenger vehicle restraint systems.

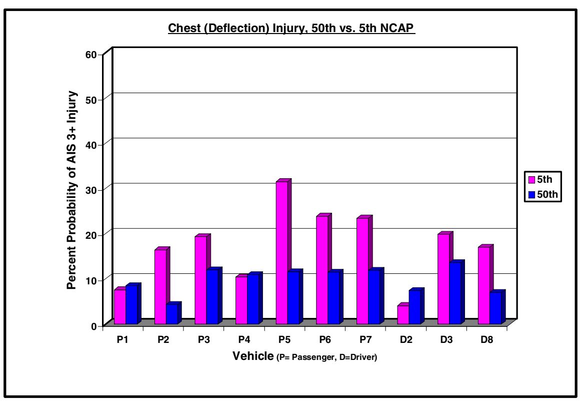

Figure 3. Driver and/or passenger occupant chest injury probability (based on deflection response) for 5th %ile female and 50th %ile male) for various driver and passenger vehicle restraint systems.

METHODS

The basic premise for the analysis was a full-factorial Design of Experiments (DOE) on 5 restraint system parameters. The restraint parameters are shown in Table 1. Pretensioners A and B are single pretensioners while C and D are dual pretensioner seat belt systems.

Table 1.
Restraint Parameter Levels Used in Analysis

Variable	Levels
Seat Belt Pretensioner	Types A,B,C,D
Seat Belt Load Limiter	Low, Medium, High
Seat Belt Payout	Low, Medium, High
Inflatable Knee Bolster	On/Off
Active Airbag Vent	On/Off

Four MADYMO (13) base models were created for the purpose of this study using a sport utility vehicle configuration. The first was modified from an existing 50th passenger NCAP model by adding pretensioner Type A and by adding replaceable parameters for turning pretensioners and active venting on or off based on parameters in the matrix. The first file also called out the proper load limiter functions based on peak and payout of the load limiter (9 combinations). The second file for the 50th male has an added inflatable knee bolster. The third and fourth input files were created from the first two files by repositioning the seat and replacing the 50th male dummy with a 5th female dummy. The iSight (14) program was used to generate a 72 run matrix with the remaining input parameters (load limit peak and payout, active vent, and pretensioner configuration). It was set up to make preliminary calculations to get the required replaceable parameters for each run, make the proper substitutions in all four input files, submit the jobs to the MADYMO solver in parallel (up to 3 jobs could be run simultaneously), extract desired data from the output files after completion, perform calculations of injury probabilities from the output, perform combined calculations after all four runs for each iteration finished, then start over with the next line of the matrix and continue until all 72 lines of the matrix were done. When the runs were complete, a complete results file was generated from all 288 runs (72 parameter combinations times 4 input files) to use for analysis with the input parameters, the results, and the calculations.

Probabilities for AIS 3 and greater head, chest and neck and AIS 2 (and greater) lower extremity injury were derived from published charts by NHTSA (15,16) and Mertz et al (12,17,18). HIC was used as the head injury measure, while absolute chest compression and neck tension were used as injury measures for the chest and neck respectively.

The peak injury values taken from the MADYMO output file and compiled in the results file database of the 288 runs were compared to the published injury probability functions for an AIS 3+ injury. An RMS (root mean squared) value was calculated from the head, chest, and neck injury probabilities (square root of the sum of the squares). The rationale for using the RMS value will be discussed later. Each run was ranked in terms of its RMS value and the associated restraint parameters. The MiniTab Statistical software was used to process the data to obtain relevant statistical measures, and provide main effects plots, and plot the data for each run with respect to injury probability and various restraint parameters.

RESULTS

A plot of all 288 runs for the SUV model demonstrated the ability of the analysis to show differences (Figure 4). It can be seen immediately from the figure that the probability for injury of the various body region is low for this model. Neck injury

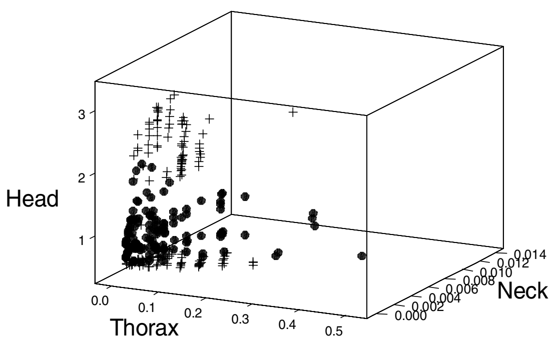

Figure 4. Percent probability of AIS 3+ head, neck or thorax injury for 5th percentile female (+) or 50th percentile male (•) for each of parameter run of the DOE matrix for the SUV model.

shows the lowest probability followed by thorax and head with increased probabilities respectively. The 50th percentile male dummy shows a tight single cluster of results with a small distribution of outlier results. The 5th female dummy shows two clusters of results with the second cluster showing higher head injury risk than the first cluster. Further examination of the second cluster of results indicates that all of those cases did not have the active venting feature in the airbag module.

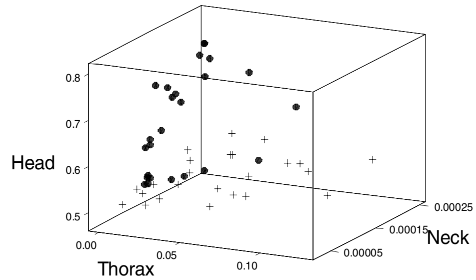

Figure 5. Top 25 restraint systems in terms of percent probability of AIS 3+ head, neck or thorax injury for 5th percentile female (+) or 50th percentile male (•) for each of parameter run of the DOE matrix for the SUV model.

Rejecting those cases, the top 25 systems for both 5th percentile male and 50th percentile male are shown in Figure 5. A tabulation of those cases was made from lowest RMS score to highest RMS score. The top 5 systems for each occupant are shown in Table 2 in terms of the combined injury risk defined as the RMS value for the three injury criteria. The table shows the system components for those top 5 systems for each occupant. As previously stated, the active venting was present in all systems for the 5th as well as the 50th. All systems included the lowest load limiter used in the analysis, however, all the 5th percentile dummy systems used the high payout option. A mix of pretensioners is also present with the 50th systems dominated by the more complex pretensioner types. No 50th system in the top 5 required a knee bag.

Table 2.
Restraint System Definition for Top 5 Scoring Systems According to Occupant Size. (PRET=Pretensioner, LL=Load Limit, PAY=Webbing Payout, AV= Active Vent, KB= Knee Bag)

OCC	RMS	PRET	LL	PAY	AV	KB
5th	.489	A	LOW	HIGH	Y	N
	.499	B	LOW	HIGH	Y	N
	.507	D	LOW	HIGH	Y	Y
	.513	B	LOW	HIGH	Y	Y
	.515	A	LOW	HIGH	Y	Y
50th	.563	C	LOW	LOW	Y	N
	.567	D	LOW	LOW	Y	N
	.572	D	LOW	MED	Y	N
	.576	A	LOW	MED	Y	N
	.577	C	LOW	MED	Y	N

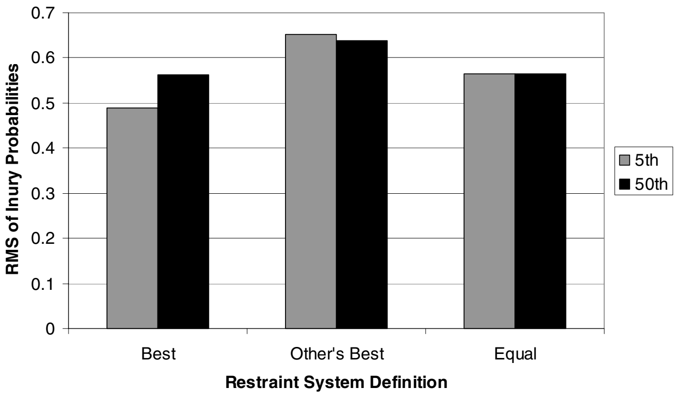

Figure 6. RMS comparison for 5th and 50th in terms of best system for itself, the other dummy's best system, and system for equal probability.

Table 3.
Restraint System Definition for Equal RMS Probability of Injury for 5th Percentile Female and 50th Percentile Male Dummy.
(PRET=Pretensioner, LL=Load Limit, PAY=Webbing Payout, AV= Active Vent, KB= Knee Bag)

OCC	RMS	PRET	LL	PAY	AV	KB
5th	.565	A	MED	HIGH	Y	N
50th	.565	C	LOW	LOW	Y	N

The 5th percentile dummy's best system was the 13th best system for the 50th (out of 144), while the 50th's best system was the 38th best system for the 5th.

When comparing the result of using the other dummy's best system in the simulation, i.e., using the 50th's best system in the 5th's model and vice versa, the result is shown in Figure 6. Both dummies RMS probabilities increase relative to its best system. In terms of actual injury criteria, the HIC and chest compressions can increase by as much as 30% for these simulations. By picking the systems that result in equal probability for both dummy models, there is no degradation for the 50th percentile dummy (RMS changed 0.002), but a more substantial increase for the 5th percentile dummy (from 0.489 to 0.565). When looking at the injury criteria, this result translates into a 35 point increase in HIC and 3mm increase in chest compression. Both systems for equal probability favor no knee bag and the presence of active venting while none of the seat belt characteristics are same in either system.

DISCUSSION

The efforts to define adaptive restraint systems have been discussed in both the media and scientific publications (1,7). It is generally acknowledged that these systems would have a beneficial effect on occupant response as the components of the restraint system could be adjusted to the occupant size, position, crash configuration, etc (6,19,20). It becomes prohibitive, in terms of cost, to test all possible combinations of test and restraint system conditions, thus leading to computer methods to analyze the system. Iyota and Ishikawa (21) demonstrated a modeling method to assess injury risk for 5th, 50th and 95th percentile dummies based on load limiting at the seat belt retractor and airbag vent hole size. Using the NHTSA derived combined head (HIC) and chest injury (chest G) injury probability calculation, they defined the parameters of the two variables that would give a similar injury probability for all three occupant sizes.

The current study uses a similar modeling approach, but uses three injury parameters (HIC, chest compression, and neck tension) and more restraint system components to define the restraint system that results in equal probability of injury risk by body region for the 5th percentile female and 50th percentile male dummies. Defining injury risk is not a new issue as both governmental (US-NCAP) and consumer testing agencies (IIHS and EuroNCAP) express their injury criteria and levels of performance based on risk of injury to various body regions (16,22,23). However, the probabilities for ratings are not balanced. For example, the IIHS criteria for an acceptable-marginal vehicle rating based on head (HIC), Chest (chest compression) and Neck (neck tension) injury criteria would give an unequal probability of AIS3+ injury for head (5.6%), neck (4.5%) and chest (45%) for a 50th percentile male dummy. The approach described in this paper selects restraint system parameters that result in an equal probability of injury for each dummy body region as well as for each dummy size. In this manner, the overall system design can be achieved that satisfies the equal probability goal. The system for the 5th female and 50th male that gave the best result for each dummy would not have been the best system for the other dummy. By defining an equal probability, it was possible to find the appropriate system components. In the current simulation, the HIC, chest compression, and neck tension probabilities remained equal as the RMS number indicates. Also, it is assumed that the injury severities considered for each body region were equal as determined by their AIS value. That is, an AIS 3 head injury carried the same severity as an AIS 3 chest injury. While NHTSA sums the head and chest injury probabilities in their NCAP star rating, this report calculated an RMS value for head, neck and chest that provided a method for ranking the various systems analyzed.

There may be challenges in achieving this goal of equal injury probability as the restraint system parameters are adjusted. System designs may not be possible based on the components selected in the analysis. In its response to the NHTSA NPRM on addition of 5th female to NCAP test conditions, General Motors cited that the performance of the 5th percentile female dummy "improved with higher output/more aggressive airbags"(24). This can have negative consequences on other test conditions such as unbelted occupants and out-of-position occupants. This was discussed by Trosseille et al (25) who analyzed the out-of-position risk of an optimized thorax restraint system comprised of a pretensioner, load limiter and airbag system.

The current analysis did not take into account airbag inflator output, airbag shape, or vent hole size, all of which may have an effect on the occupant response. The active venting feature used in the analysis, provides for a controlled release of airbag gas that was shown to have a positive effect on the occupant response when used. It is the process in this study that needs to be highlighted rather than the results since an analysis comprised of thousands of simulations is possible as the number of parameters increases. Regardless of parameters used, all results will lead to an equalization of injury probability by occupant size and occupant body region rather than just considering the basic injury reference values. This analysis does not consider effects of age on likelihood of injury (26) nor does it consider that the system definition to achieve equal probability from one vehicle may be different than that of another vehicle. On a higher level of any injury risk to any occupant, Kullgren et al (27) demonstrated that the injury risk functions differ from vehicle to vehicle for a given crash severity. As the future development of restraint systems continues, this new technique of establishing equal injury probability for all occupant sizes, while maintaining margins for acceptable injury criteria, may lead to further improvements in vehicle safety.

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OPTIMISED PRETENSIONING OF THE BELT SYSTEM: A RATING CRITERION AND THE BENEFIT IN CONSUMER TESTS

Harald Zellmer
Christoph Kahler
Burkhard Eickhoff
Autoliv GmbH, Elmshorn,
Germany

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ABSTRACT

A common understanding is that in a frontal crash an early coupling of the occupant to the vehicle deceleration is required. This is provided by pretensioning of the belt system. The objective of our study was to set up a rating criterion for pretensioner performance, to benchmark current pretensioner systems, to define requirements for an optimal pretensioning, and to quantify the benefits in both US- and EuroNCAP testing.

A generic test environment was developed and sled tests with different pretensioners and combinations of pretensioners were conducted. As a result, systems with either both retractor and anchor plate pretensioning, or buckle and anchor plate pretensioning gave direct reduction of the dummy chest acceleration values. Additional to the reduction in dummy loading, a reduction in dummy forward displacement occurred. Using this additional space by reduction of the load limiter level of the seat belt resulted in a further reduction in occupant loading, especially in chest deflection. For the determination of the appropriate load limiter level, MADYMO simulation was used. In a further step, a rating criterion for pretensioners was defined. It rates the energy difference of the dummy compared to the vehicle during the crash as percentage of the vehicle energy, i.e. a low figure indicates a good coupling.

As a result, with double pretensioning and respectively tuned load limiter level, chest deflection and acceleration in both EuroNCAP and US-NCAP can be reduced by about 20% - 25% compared to single pretensioning. A low energy difference in the pretensioning rating criterion showed a good correlation to the dummy readings.

With the outcome of the study, requirements to an optimal pretensioning are discussed in respect to a good coupling and to possible injuries induced by aggressive pretensioning.

1. INTRODUCTION

Consumer tests world-wide are posing constantly increasing demands on the vehicle structure and the occupant protection system. As a consequence of the offset-crash, the rigid vehicle structure results in increased loads on the occupants, in particular in frontal crash tests with 100 % overlap. Thus an optimised restraint system consisting of a safety belt with pretensioner, load limitation and airbag has to reflect these demands. Here particular attention has to be paid to the belt system, since it is exclusively the belt system which is responsible for the restraint of the occupant during the first phase of the crash.

The effect of the belt system can be classified into two phases:

1) the belt pretensioning following the crash only by a few milliseconds and creating the optimum pre-requisites for the restraint of the occupant;
2) the load limitation which keeps the force at the shoulder belt to a pre-defined level during the forward displacement of the occupant thus leading to an optimum utilisation of the space available in the interior.

1.1 Belt pretensioning and load limitation

Too much slack in the belt system results in a deterioration of the occupant loading in frontal crashes and may favour submarining. For instance, 80% of car drivers have a slack varying between 40mm and 90mm in the summer and 40mm to 120mm in the winter /1/. The pretensioner is intended to minimise this slack even before an occupant forward displacement in a crash. So to say, the belt system is fine tuned for an optimum starting position in the first milliseconds following a crash.

The load limiter in the 3-point belt is intended to limit the forces exerted by the belt and thus the values for the thoracic load. Already in the early 1970ies load limiters were applied in serial production, at that time, of course, without airbag. Their benefit has been demonstrated by accident analyses /2/. Today load limiters are mostly applied in combination with an airbag to achieve an optimum alignment of the restraint system.

Even if the impact of the load limitation is of importance for the occupant load /3,4/, in the following we intend to focus on the influence of the pretensioner.

1.2 Pretensioning approaches for front seats

Figure 1 depicts various approaches for pretensioning of the belt system. It is differentiated between the following pretensioner systems each positioned at the respective fixation points:

- retractor pretensioning
- buckle pretensioning
- anchor plate pretensioning
- any combination of the above three methods.

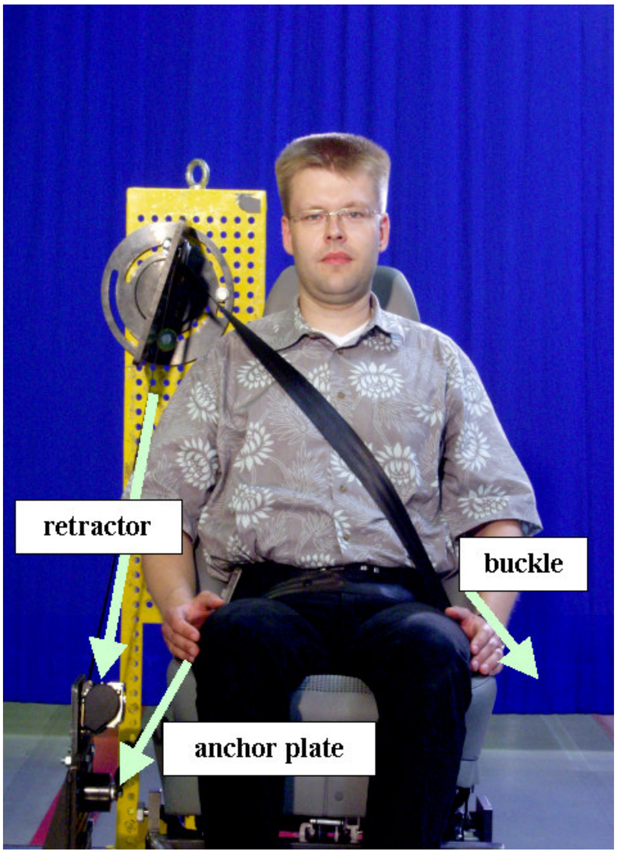

Figure 1. Pretensioning of the Belt System

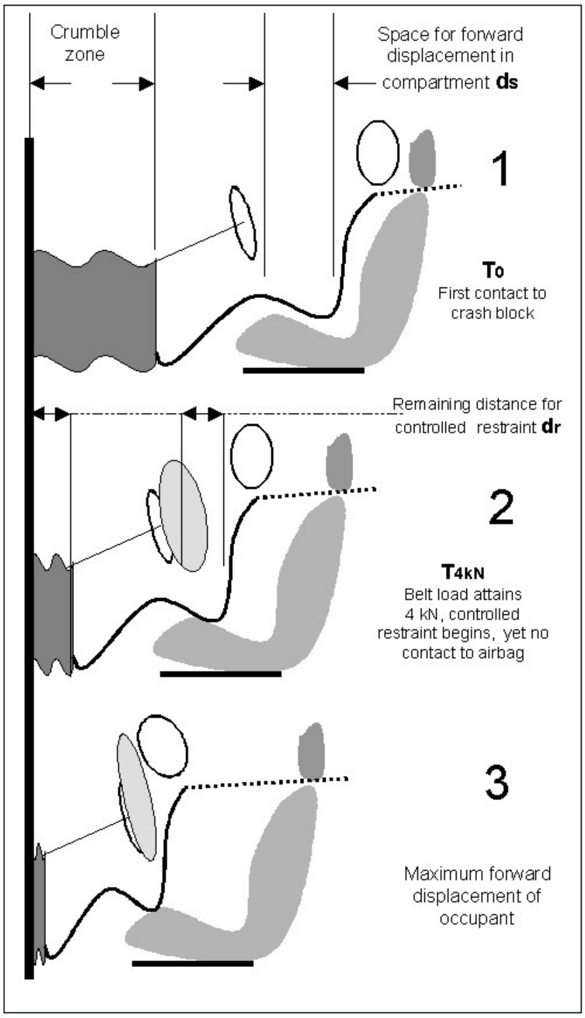

Figure 2. Car and occupant behaviour in frontal impact

1.3 Criterion for occupant coupling

In frontal impact events, one of the main functions of the belt system is an early coupling of the occupant to the vehicle deceleration. In order to evaluate this, a coupling criterion was defined /5/ on the basis of the "Ride Down Effect" (RDE) /6/. The RDE gives a percentage value of the remaining crumple zone, when the dummy retardation starts. The criterion we defined evaluates the path remaining for the passenger when a load of 4kN is reached on the shoulder belt. The point in time when this happens is defined as $\mathbf{T_{4kN}}$. In doing so, we assume that this is the moment when a controlled restraint starts. The forward displacement of the dummy in the area of the upper thoracic vertebra T1 at $\mathbf{T_{4kN}}$ is defined as $\mathbf{d_{4kN}}$ and the remaining distance for the retardation of the occupant $\mathbf{d_r}$ is computed as follows (please refer to figure 2):

$$\mathbf{d_r = d_c + d_S - d_{4kN}}$$

with:

$\mathbf{d_c}$ = remaining length of crumple zone at $\mathbf{T_{4kN}}$
$\mathbf{d_S}$ = space for forward displacement at $\mathbf{T_0}$

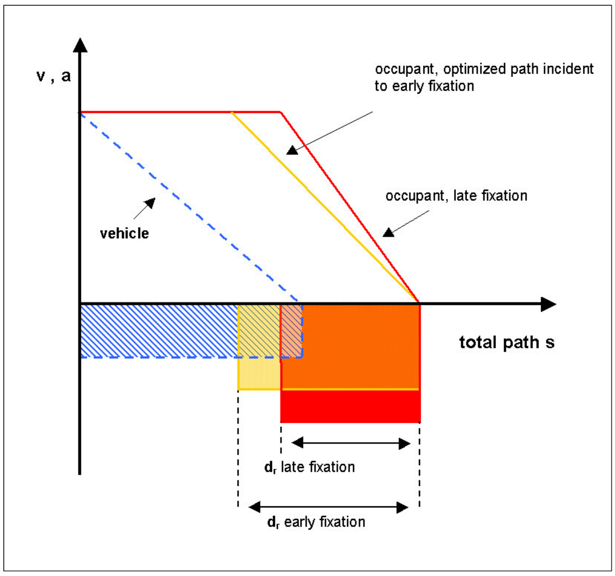

Figure 3. Speed v and deceleration a vs. distance at idealistic square shaped decelerations

Figure 3 shows that a high value for $\mathbf{d_r}$ results in a marked reduction of the occupant acceleration. This has been confirmed in tests and simulations /5/.

1.4 Limitations posed by single pretensioning

From the technical point of view, increasing the pretension load aiming at achieving 4kN on the shoulder belt already during the pretensioning phase would be feasible. This would result in an optimum occupant coupling. However, this achievement would fire back on the occupants. For biomechanical reasons, the force exerted on the shoulder should not exceed 1.5kN – 2.0kN, see fig. 4. This holds true in particular since a pretensioning does not only make sense for frontal impact situations. In this case, it would be reasonable to assume that shoulder forces of this type are reached during the crash incident anyhow. In the cases of rear impact or roll-over we do not assume a priori that forces of this magnitude are reached. These loadcases would then require an additional pretensioner with lower performance or a retractor pretensioner with variable tension performance.

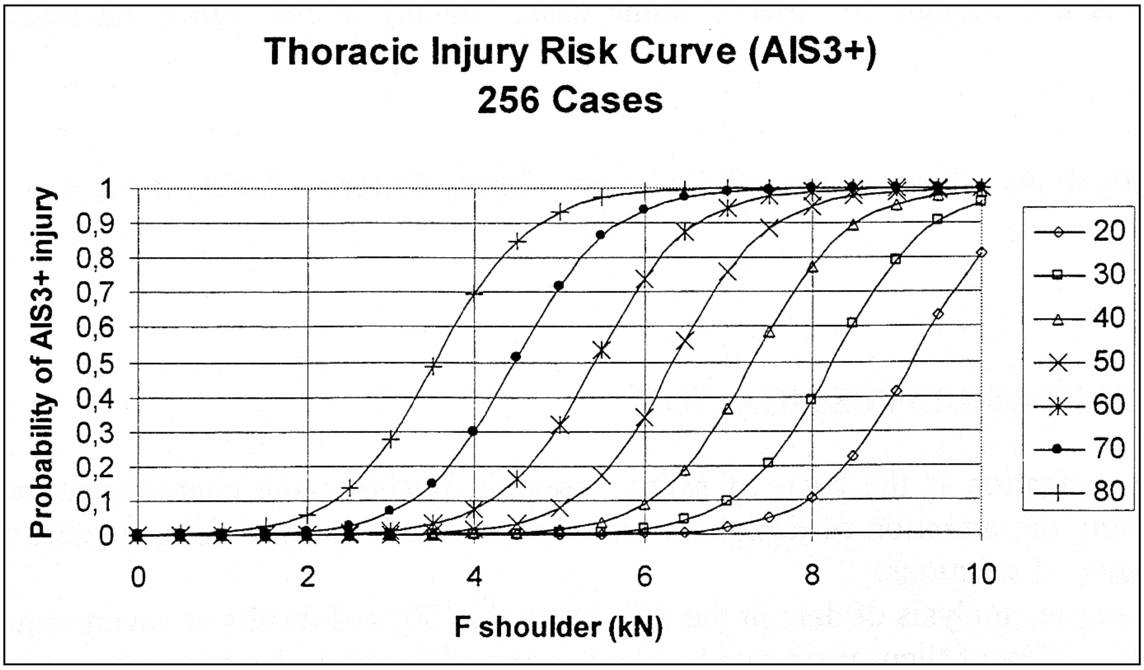

Figure 4. Probability of severe thoracic injuries (AIS3 or more severe) depending on the shoulder belt force and the occupant´s age /7/

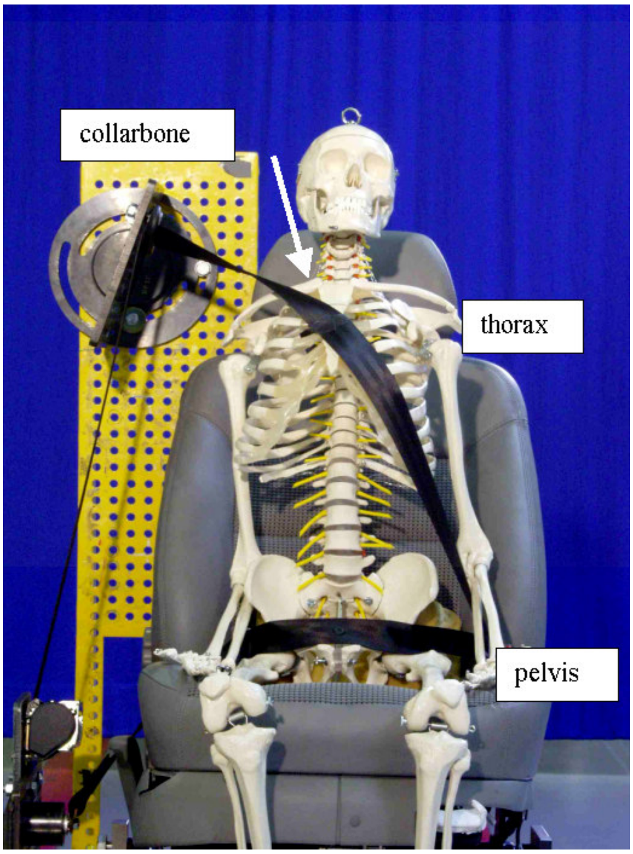

Figure 5. Loading to the body by the belt

1.5 Optimisation of the pretensioner system

Now, which combination of pretensioners is the most suitable choice to achieve an optimum occupant fixation? The following pre-requisites have to be met:

1. Strong fixation of the occupant as early as possible;
2. Limitation of the shoulder force to 1.5kN – 2.0kN;
3. Set-up of a suitable force on the retractor to minimise the film spool effect.

In order to meet these pre-requisites, it is necessary to guarantee that a high load is applied in the pelvis area by means of an anchor plate pretensioner. From the biomechanical point of view, this area can withstand higher loads than the shoulder area. In our opinion, even a value of 4kN would not pose any problem. As a consequence, we need a first pretensioning at the buckle or retractor to take out the belt slack observing conditions 2 and 3, and the second pretensioner to be fired at the anchor plate. Moreover, a high application of load in the pelvis area makes sure that an essential factor of the restraint effect is performed there, see figure 5.

1.6 Scope

The objective of our study is to quantify the benefits of the above outlined double pretensioning strategy. The rating criterion for occupant coupling described in section 1.3 is no longer valid for this kind of pretensioning as it only rates the force at the shoulder belt and not at the lap belt in which we want to impose the stronger part of pretensioning. Furthermore, there are other ways to couple the occupant to the car deceleration in the very first part of the frontal impact, such as the knee airbag or the pelvis restraint cushion, an inflatable seat ramp which is in production in some cars. A new rating criterion should take care about these as well. In order to develop such a criterion,

1. a generic test environment was developed,
2. sled tests with different pretensioners and combinations of pretensioners were conducted, simulating both US- and EuroNCAP crash pulses,
3. the load limiter level of the seat belt was tuned to use the gained space for forward displacement to its optimum,
4. the benefit in terms of occupant loading in the tuned configuration was determined,
5. a new coupling criterion was developed and compared with the outcome of the sled tests.

2. TESTS: SET-UP AND RESULTS

The generic test environment developed should represent a European mid-size car (cf. figure 6.) The seat has a stiff structure but a seat cushion with a deformable sheet metal below from a serial car in order to simulate the pelvis seat interaction. The steering wheel with airbag is fixed to a stiff bar. The dummy used was a Hybrid III 50th percentile. Two different pulses, one representing an US-NCAP, the other presenting an EuroNCAP pulse were selected (cf. figure 7.)

The following pretensioners were tested:

1. baseline: without pretensioner,
2. buckle pretensioner only,
3. retractor pretensioner only,
4. retractor and anchor plate pretensioner,
5. buckle and anchor plate pretensioner,
6. retractor and buckle pret. (EuroNCAP only.)

In the double pretensioner configurations the pretensioner named first was fired first, the second one with a time delay of 5ms to 11ms in order to avoid interactions between both.

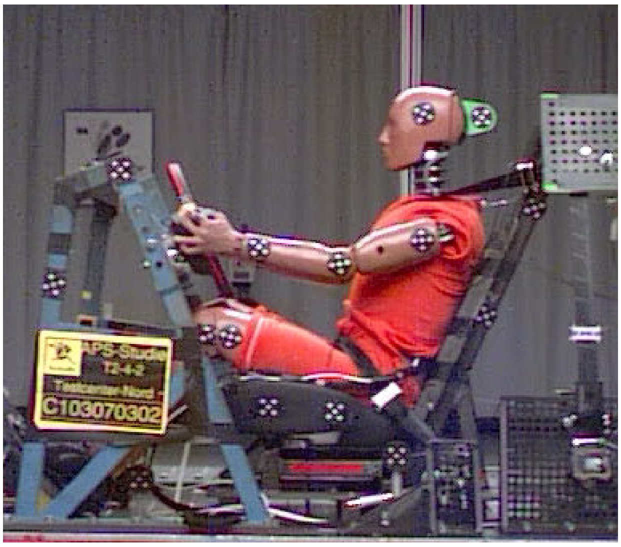

Figure 6. Generic test set-up

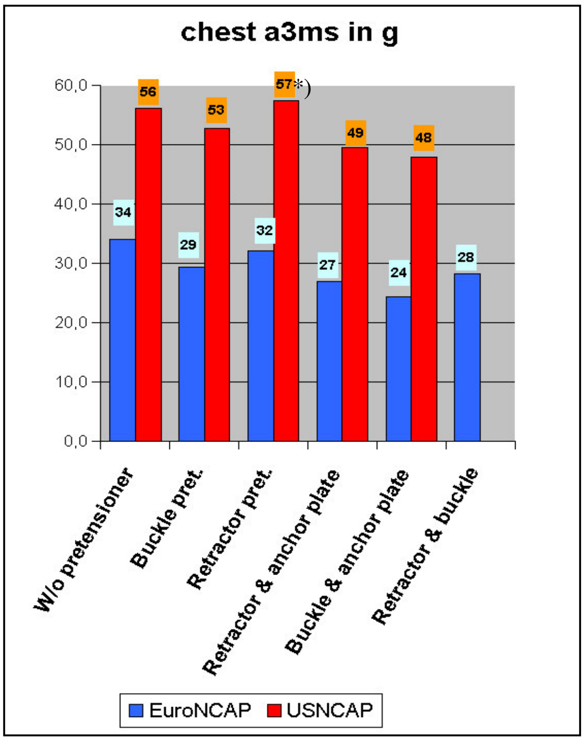

Figure 8. Chest acceleration a3ms

*) this value is higher than expected and caused by contact of the dummy pelvis to stiff substructures in the seat, see text

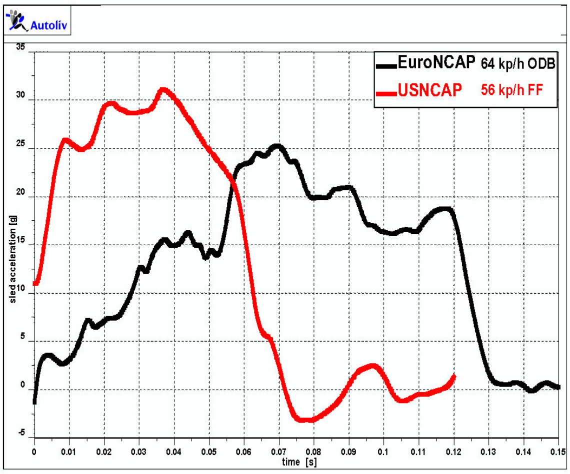

Figure 7. Crash pulses simulated in sled tests

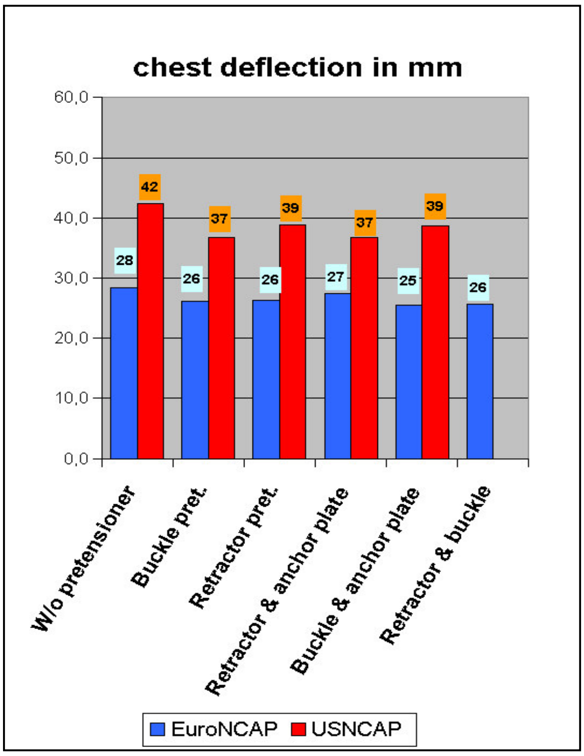

Figure 9. Chest deflection

In the further description of the test results we will focus on chest loading, i.e. chest acceleration a3ms and chest deflection, as these are the parameters that predominantly are influenced by the belt system. Figures 8 and 9 show the results. For each test configuration two tests were performed, the figures listed are the mean values. Figure 8 shows that pretensioning in general gives a benefit in chest acceleration with the one exception of the retractor pretensioning in the US-NCAP configuration. This one is to be considered being an artefact of the generic test set-up: the dummy pelvis had contact to the stiff seat substructure resulting in a pelvis z-acceleration which gave rise to a chest z-acceleration. Thus, the resultant chest acceleration given in the figure rose as well. In terms of chest acceleration, double pretensioning again reduced the figures.

Figure 9 shows the results for chest deflection. It can be seen that pretensioning in general reduces chest deflection, but at this stage no advantage of double pretensioning can be detected.

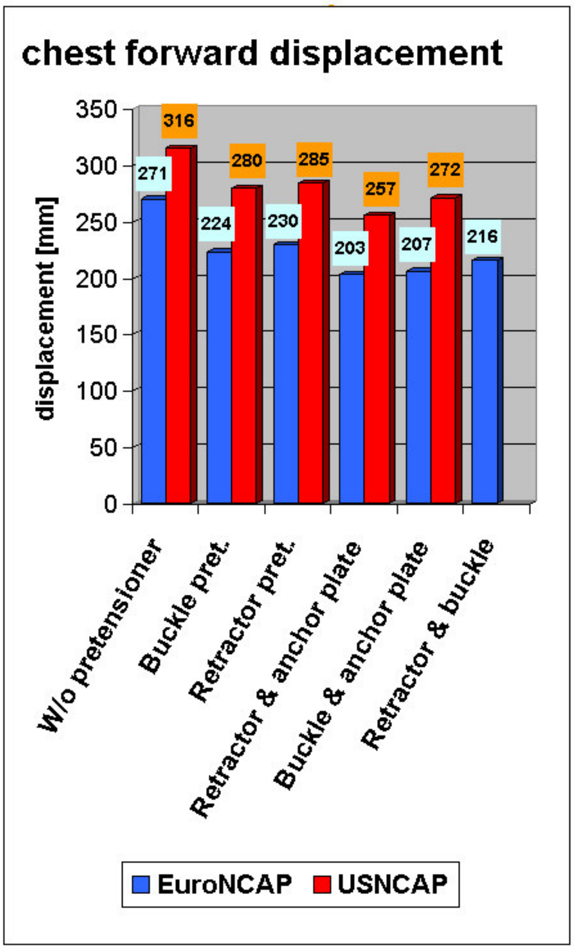

Figure 10. Maximum chest forward displacement at lower steering wheel level

Figure 11. Double pretensioning reduces chest forward displacement

Dummy chest forward displacement at the level of the lower steering wheel rim is reduced by single pretensioning by about 10% compared to the tests without pretensioner and again by around 10% when comparing double with single pretensioning (cf. figures 10 and 11.) It shows that the combinations retractor, resp. buckle and anchor plate pretensioner show significant bigger reductions than the combination of retractor and buckle pretensioner. Therefore, in the following discussion the latter combination is disregarded.

Taking the forward displacement of single pretensioning as baseline, the question is which benefit would provide double pretensioning when the load limiter force is reduced to a figure that forward displacement is equal to the one of single pretensioning. In order to answer this question, a MADYMO model was set up and validated. As a result, lowering the shoulder belt force, which was about 5000N in the baseline set-ups, by 500N to 1000N reduced the chest deflection by 2mm to 4mm in both US- and EuroNCAP configurations.

In order to validate the findings of the MADYMO investigation, a new test series (series 2) was set up. The configuration with buckle pretensioning was chosen as baseline because it gave the lowest chest forward displacement in the first test series. As it is well known that especially chest deflection is very dependant on the individual dummy, two repeat tests of this configuration were performed. The mean values of the two repeat tests now serve as baseline for the further comparison. The figures are listed in table 1 for the US-NCAP test condition and in table 2 for the EuroNCAP one. Compared to the tests of the first series, the chest deflection of the new baseline tests is 3mm to 5mm higher and the forward displacement in the EuroNCAP set-up 18mm lower. Additional to the above mentioned dummy problem, the latter one might be the result of slight deviation of the EuroNCAP test pulses between the two series.

Table 1.
Chest deflection and acceleration in relation to pretensioning and load limiter level, US-NCAP test condition, test series 2

	Shoulder belt force [N]	Chest forward displacement [mm]	Chest deflection [mm]	Chest acceleration a3ms [g]
Baseline buckle pretensioner only	5000	279	40	48
Buckle and anchor plate pretensioner	4500	276	34	44
Retractor and anchor plate pretensioner	4000	279	35	44

Table 2.
Chest deflection and acceleration in relation to pretensioning and load limiter level, EuroNCAP test condition, test series 2

	Shoulder belt force [N]	Chest forward displacement [mm]	Chest deflection [mm]	Chest acceleration a3ms [g]
Baseline buckle pretensioner only	5000	206	31	30
Buckle and anchor plate pretensioner	4500	204	25	27
Retractor and anchor plate pretensioner	4500	199	26	28
Retractor and anchor plate pretensioner	4000	218	23	29

The shoulder belt forces listed in tables 1 and 2 are to be regarded displaying the order of magnitude. The shoulder belt force in general is dependant on the diameter of the torsion bar in the spindle of the retractor, the number of turns of the torsion bar by pay out of webbing, the amount of webbing on the spool, and the friction in the pillar loop. With this, the load limiter level at the shoulder does not remain constant over the crash and can deviate from the given values by about 200N.

All load limiter levels are tuned to show the same forward displacement as the respective baseline test. The only exception is the case with retractor and anchor plate pretensioning in the EuroNCAP loadcase and 4000N shoulder belt force. This test had 12mm forward displacement more than the baseline. As this configuration showed in the US-NCAP loadcase the same forward displacement as the respective US-NCAP baseline, it can be assumed that this can be a valid configuration for both loadcases.

For each configuration two tests were performed. The mean deviation was ±0,61mm (1,89%) in chest deflection and ±0,19g (0,61%) in chest acceleration a3ms for an overall of 14 Euro- and US-NCAP sled tests. Thus, the figures can be regarded being quite reliable.

As a result, with double pretensioning, chest deflection was reduced by 16% to 26% in the EuroNCAP set-up and by 13% to 15% in the US-NCAP set-up.

3. COUPLING CRITERION

As outlined in the introduction, an early coupling of the occupant to the car deceleration is required. In section 1.3 a coupling criterion is described. This criterion is related to the force in the shoulder belt. As shown in the previous chapter, a strong coupling in the pelvis area is beneficial for the dummy loading, but this does not show in the shoulder belt force. Thus, there is a need for a new coupling criterion which reflects this. The goal is to define a calculation method to compare different pretensioning systems in the same test environment in respect to occupant coupling. The following conditions were defined:

- direct calculation of dummy data, no use of indirect forces (like belt forces);
- separate calculation for chest and pelvis, to pay respect to pretensioning at the pelvis.

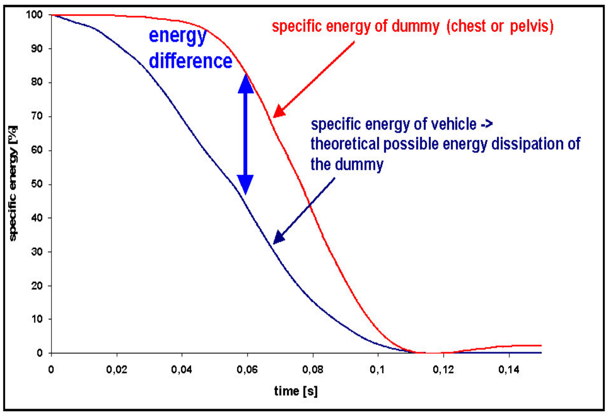

Figure 12. Energy loss of car (blue line) and occupant (red line) in a specific sled test

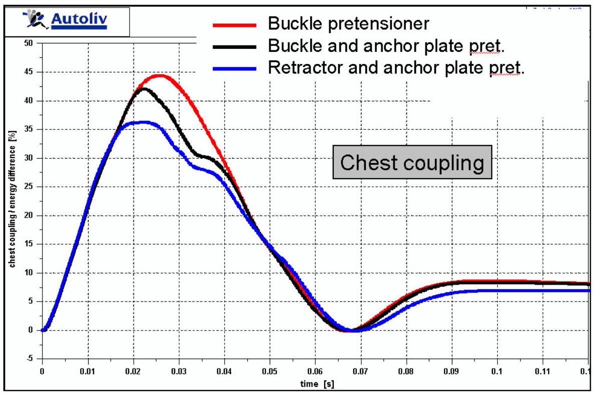

Figure 13. Relative energy loss detected from the thorax accelerometer of the dummy, US-NCAP test pulse

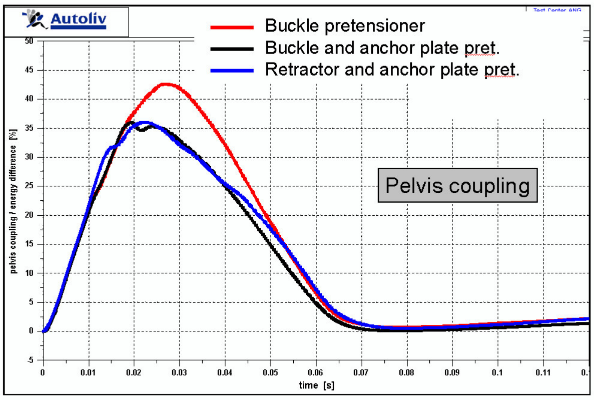

Figure 14. Relative energy loss detected from the pelvis accelerometer of the dummy, US-NCAP test pulse

The approach chosen is a comparison of theoretically possible and real energy reduction of the dummy during deceleration. Figure 12 depicts the relative energy reduction of the car, resp. sled and the dummy during a frontal crash. The energy at the beginning of the crash is taken as 100%.

The relative energy **e** is calculated from the car acceleration **a** by integration:

$$e = \frac{E}{m} = \frac{1}{2}\left(v_0 - \int a\, dt\right)^2,$$

with $\mathbf{v_0}$ being the velocity at the beginning of the crash.

The same energy calculation is done for the dummy, for practical reason independently for chest and pelvis. In the theoretical case that the dummy is optimally coupled to the car deceleration, its energy path will follow that of the car. Therefore, a deviation from this is a loss in coupling. Lower energy difference means better coupling.

Figures 13 and 14 depict the relative energy loss of chest and pelvis in the US-NCAP test condition. The lowest energy loss is detected for the double pretensioning set-ups. In the pelvis coupling both double pretensioner configurations show almost the same coupling. For the chest the combination of retractor and anchor plate pretensioner show the lowest energy loss.

It has to be mentioned that the maximum of energy difference is reached before airbag contact and before activation of the load limiter. Tests with a pelvis restraint cushion as supplementary restraint system to a retractor pretensioner were evaluated in the same manner. They showed that the coupling criterion here was as well a valid rating criterion.

Figure 15 depicts the coupling figures for the EuroNCAP and the US-NCAP loadcases from the first test series. It shows that the relative energy loss in general is bigger in the US-NCAP loadcase. In both loadcases it is reduced significantly by pretensioning and especially double pretensioning. Figure 16 shows as an example the relative energy difference of the dummy plotted vs. chest acceleration for the EuroNCAP loadcase. They show to be well correlated.

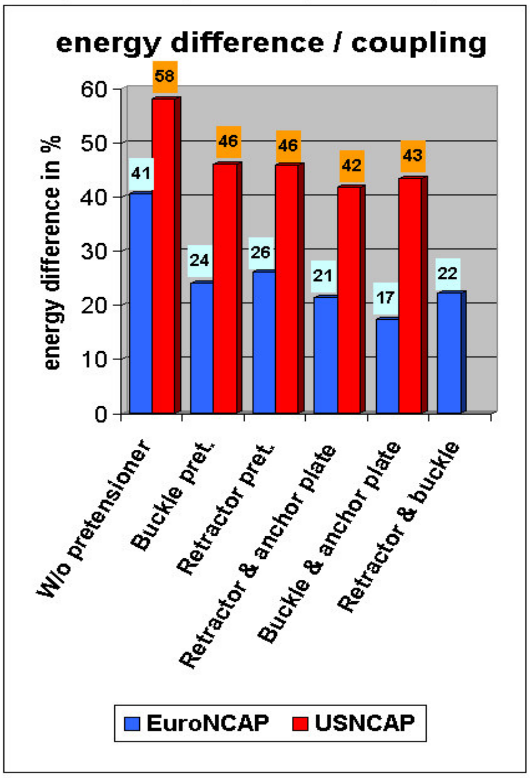

Figure 15. Maximum relative energy difference of the dummy compared to the sled from the first test series. The figures listed are the mean of chest and pelvis figures.

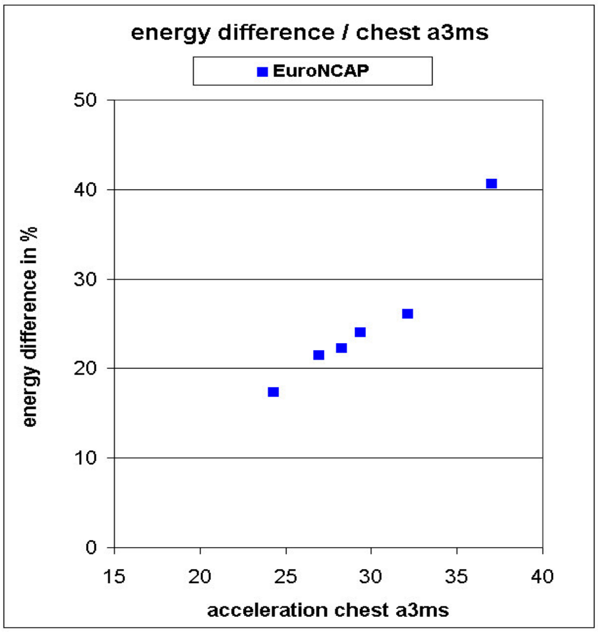

Figure 16. Maximum relative energy difference of the dummy plotted vs. chest acceleration, EuroNCAP loadcase, test series 1

4. DISCUSSION

In the following the benefit of double pretensioning in US-NCAP and EuroNCAP rating shall be discussed. For this we focus on test series 2, the results of which are listed in table 1 and 2. The outcome for the US-NCAP is shown in figure 17. For the generic test set-up it would mean an improvement from 3 to 4-star rating. As in our test series the airbag performance was left unchanged, improvements in terms of HIC might be achievable.

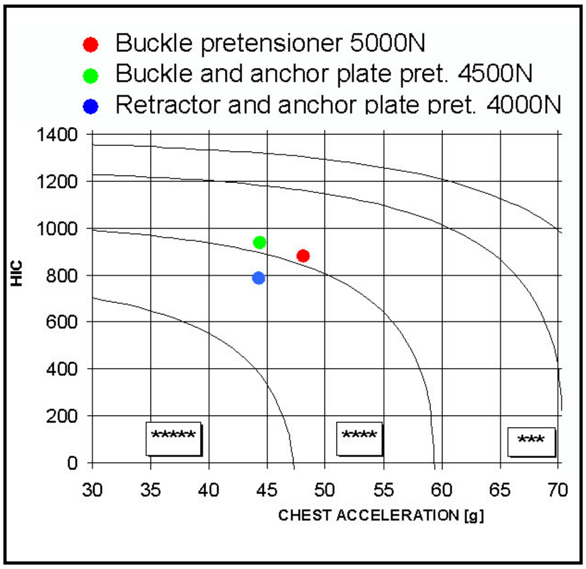

Figure 17. US-NCAP rating for test series 2

In the EuroNCAP loadcase, the chest deflection was reduced from 31mm to 23mm. This would mean in terms of points for the chest body region in the EuroNCAP rating an improvement from 2.77 to 3.80 points, i.e. one full point more in the total rating. Furthermore, the pelvis forward displacement is reduced by 44mm with double pretensioning. This can help in avoiding knee contact to the dash board and thus avoiding knee modifiers.

It has to be mentioned that by keeping the load limiter level at 5000N, pelvis forward displacement can be reduced by up to 75mm. Depending on whether the focus is on reduction of chest deflection or avoiding knee contact, the belt system can be adjusted accordingly.

The question arising is which additional benefit could be expected by further increasing pretensioner strength. This was investigated with the MADYMO model mentioned above. A double pretensioning set-up was chosen for the EuroNCAP loadcase. Both anchor plate and retractor pretensioners were fired without any time delay at T_0 (t = 0ms.) The anchor plate pretensioner was tuned to yield 4000N pre-load at the outer lap belt, the retractor pretensioner was tuned to yield 2000N pre-load at the shoulder. Both pretensioning forces being values considered to be close to biomechanical maximum values (cf. chapter 1.4). The load limiter in the shoulder belt was adjusted to 2000N, i.e. to the pretensioner level. As a result, chest deflection could be reduced by less than 10% compared to the benchmark of the best real system. This shows, that current double pretensioning systems are very close to the optimum. A further reduction in especially chest deflection can only be achieved by improved load limiter characteristics /4,5,8/.

The introduced coupling criterion (chapter 3) is a good tool to rate the coupling of the occupant to the car. As it only rates the beginning of the crash, it can be used in early stage of car development for improving coupling separately without being influenced by load limiter or airbag performance. As well it is a good tool in developing new pretensioner systems.

5. CONCLUSIONS

In US- and EuroNCAP test conditions, double pretensioning directly reduces chest acceleration by 10% - 20% compared to single pretensioning. Preliminary tests and simulations show that this is valid as well for the 5th percentile female Hybrid III dummy. Double pretensioning and reduction of the load limiter level, in order to use the full space for dummy forward displacement gained by better occupant coupling, reduces chest deflection by about 15% - 25%.

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IMPROVED UNDERSTANDING OF PASSENGER BEHAVIOUR DURING PRE-IMPACT EVENTS TO AID SMART RESTRAINT DEVELOPMENT

Richard Morris
Gabrielle Cross
MIRA Ltd.
United Kingdom
Paper Number 05-0320

ABSTRACT

The PRISM project is a European Commission funded 5th Framework project that is intended to determine appropriate smart restraint technologies for Europe.

This paper describes a volunteer study undertaken as part of the PRISM project. The purpose of the study was to gain an understanding of how passengers "brace" and react during pre-impact vehicle manoeuvres (emergency braking, rapid lane changing etc.). This information, linked to real world occupant photographic studies, gives indications of real world postures at impact that can be considered for smart restraint systems.

A total of 49 volunteers were driven in an instrumented test car and were subjected to fierce pre-impact manoeuvres without warning. Each volunteer undertook 3 tests over a period of time either from their own normal postures, from pre-defined postures, or whilst undertaking various tasks.

Project staff, aware of the tests and in control of the severity and the frequency of the tests, undertook higher risk tests including unbelted and extreme out of position tests. Also 6 crash test ATDs of different sizes were subjected to the same vehicle manoeuvres, so that their inertial behaviour could be compared with human behaviour. In all, 230 tests were undertaken, with each test being filmed from 5 on-board cameras.

The development of the test methodology is described and the drawbacks of the earlier concepts are explained, together with the improvements made. The strengths and limitations of the tests and results are also explained.

Following a discussion of the results, a number of conclusions have been drawn, regarding both human behaviour and the strengths and limitations of using crash ATDs for pre-impact work. These conclusions have implications for managing occupant postures at the commencement of impact events.

INTRODUCTION

Many of the occupant restraint systems fitted to European road vehicles only react when there is a crash and mitigate injuries in a fixed or limited manner. Some of the more modern systems have improved functionality and can "tune" their response to suit a range of variables. These may include: impact severity, occupant weight and occupant fore/aft position. Such adaptive restraints are sometimes known as "smart restraints" and most are developed to meet the US requirements of FMVSS208, in the absence of any European equivalent. Vehicle manufacturers may have their own standards in addition, but are generally considered to be based upon FMVSS208.

Restraint systems are developed around certain recognised occupant sizes, these being 5th%ile female, 50th%ile male and 95th%ile male and sometimes child ATD's. ATDs and computer models exist that facilitate this work. However, consideration should be given to the proportions of the population outside these sizes.

Restraint systems are often checked to ensure that the occupant (ATD) is not injured if the restraint system is deployed whilst the occupant is Out Of Position (OOP). A very wide range of "OOP" tests are used by the industry but little information is readily available regarding the incidence of such postures in general driving and in accidents, so prioritisation of such tests can be difficult. Anecdotal evidence and casual observation have shown that some occupants can and do adopt particularly extreme postures, such as passengers with their feet on the facia and children standing in front of the front seat passenger (Bingley et al 2005). These cases are rarely considered by the manufacturers. Accident data can give good indications of injuries sustained in specific cases, eg. CCIS (Ref. 2) and GIDAS (Ref. 3), however, it is unusual that the pre-impact posture is known or can be determined.

Typically in development programmes, there is little consideration given to pre-impact vehicle manoeuvres (such as pre-impact braking) and the resulting occupant motion from their "normal" seating position. Although some research has been undertaken by TRW (Ref. 4) on a range of volunteer drivers and by Autoliv (Ref. 5) on a single volunteer passenger, understanding in this area is still quite limited.

It is generally considered that ATDs are not good indicators of human behaviour during the pre-impact phase, as they do not respond to stimuli and do not adopt "bracing" responses.

The first work package of the PRISM project provides new and extended data in this field to assist in the development of smart restraint systems.

Photographic Studies

The initial stage of the occupant posture work was a photographic study, as detailed in the written paper "Determination of Real World Occupant Postures by Photo Studies to Aid Smart Restraint Development" (Ref. 6, paper 05-0319, Bingley). The objective of this study was to determine how occupants sit in vehicles on the roads of Europe. A total of over 5000 samples were taken from 6 test sites across Europe. These samples were analysed to determine occupant longitudinal, lateral and upper limb locations. Other potentially useful data, (child occupancy, luggage location etc) were also collected. The results from this work provided statistical information on real postures that may be considered as "pre-event" start positions – inputs for this study (Table 1).

Passenger Response Studies - Overview

A total of 49 volunteers, 4 MIRA project staff and 6 ATDs undertook a range of tests, totalling 230 in number. A range of pre-impact manoeuvres events were undertaken and the occupants were encouraged to adopt various postures before the events took place. Most of the volunteers were unaware of the nature of the tests to ensure realistic responses. The ethical issues were also considered and the ethical guidelines of the British Psychological Society were followed. One of the results of this was to ensure that the higher risk tests (including unbelted) were only undertaken by project staff in strictly controlled safety conditions. The test vehicle was instrumented and carried a range of on-board video cameras to allow later assessment of the passenger behaviour.

Table 1.
Test Postures (Selected From Photo Studies)

Volunteer Posture (Not Aware of Event)
Normal (Own) Position
FMVSS 208 ATD Equivalent Position
Looking in Vanity Mirror
Dash Control Adjustment (Radio)
Arm on Waist Rail
Arm on Arm Rest
Holding Roof Grab Handle
Arm Out of Window
Holding Head Restraint (both hands)
Holding Magazine, Legs Crossed, On Phone
MIRA Staff Posture (Aware of Event)
Reaching into Footwell
Adjusting Seatbelt
Drinking / Eating
Sitting on Foot / Feet
Turning to Talk to Rear Seat Passengers
Unbelted
ATD Posture
HIII 95th %ile Male - normal
HIII 50th %ile Male - normal
HIII 5th %ile Female - normal
HIII 6 Year – normal (No child seat)
HIII 3Year – Held Standing Between Passenger Legs
CRABI – Held in Passengers Arms

The test programme matrix was determined from the postures selected from the photographic studies (Table 1) and with a range of vehicle motions (Table 2).
Risk assessments for the tests were also undertaken.

Table 2.
Vehicle Manoeuvres (Simulating Pre-Impact)

Straight line emergency braking
Rapid lane change (sudden right, then left, as if to avoid an oncoming vehicle)
Rapid lane change, then emergency braking
Rapid direction change followed by a lift-off over steer (resulting in a spin or a partial spin, as if out of control before sliding into a tree)
Rapid direction change followed by an opposite direction change (as if driving fast down a sweeping road)

METHODOLOGY

Test Rationale

In this work, the basic rationale was that under extreme stress or perceived danger, basic human survival instinct would dominate and in general, passengers would react in the same way under similar test conditions.

The primary assumptions were that the passengers would react to the vehicle motion, the sudden braking etc. Although the tests would be carried out on a proving ground, it was considered that the tests would be sufficiently realistic to obtain valid occupant reactions. In the event however, other factors proved to be dominant and additional controls had to be put in place to obtain acceptable results.

Test Facilities

The tests were undertaken at MIRA Ltd. in Warwickshire, UK. The test track selected was the handling circuit. The circuit is a closed, single user facility with a number of potential routes and the direction of travel is totally free. The surface is a very high grip material called "Delugrip" which allows for extremely high deceleration levels and cornering speeds. The circuit has an office close by for briefing and de-briefing of volunteers.

In general, the test vehicle was driven around the circuit in a clockwise direction and various vehicle manoeuvres were undertaken at suitable points around the track.

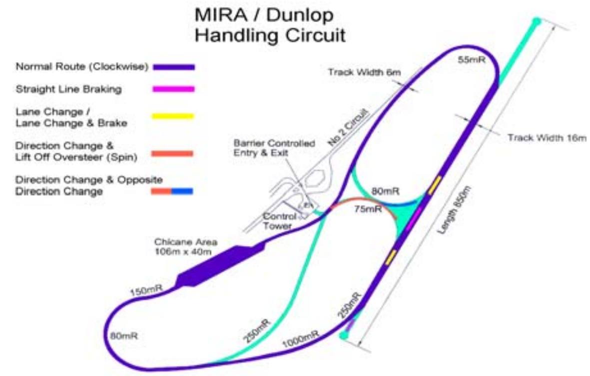

Figure 1. The MIRA Handling Circuit

The Test Vehicle

The test vehicle was a RHD 5-door Ford Focus (Figure 2). It was selected since it represents the medium hatchback size popular across Europe and it was available at MIRA during the scheduled test period. The airbag system was disabled for safety, in case it should deploy when the volunteers were in close proximity.

The vehicle was fitted with 5 cameras, longitudinal and lateral accelerometers and a data logging system. The ethical constraints meant that it was necessary to declare to our volunteers that they may be filmed, but it was not intended that the cameras affect the passenger behaviour, so they were hidden as much as possible. Two of the cameras could not be hidden, but were placed out of the passenger's line of sight. As a result, the semi-concealed cameras were rarely noticed and the novelty of the testing and the environment and the deliberate distractions ensured that the volunteers quickly forgot that they were being filmed.

Figure 2. The Ford Focus Test Vehicle

The five cameras fitted were arranged to provide the optimum views of the volunteer passenger. Miniature cameras were installed forward of the passenger's head whilst the larger cameras were positioned behind.

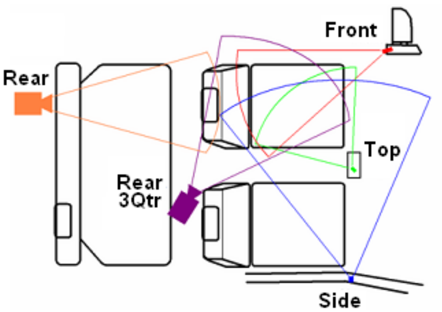

Figure 3. Diagram showing positions of the 5 cameras and their fields of view

The front miniature camera was mounted in the passenger door, in the panel surrounding the door mirror adjuster giving a frontal view of the volunteer. (Figure 4.)

Figure 4. Showing the front camera location and a sample of the camera output.

The top camera was mounted in the roof, concealed in the overhead lamp. This gave an overhead view showing position of the hands and giving information about the foot position (Figure 5).

Figure 5. Showing top camera location in the map lamp and a sample of the camera output.

The side camera was mounted in the opposite A-pillar trim and provided a lateral view, giving a clear indication of forward motion of the occupant and proximity to the airbag module (Figure 6).

Figure 6. Showing the side camera location on the A-pillar and a sample of the camera output.

A pair of video cameras were also fitted to the vehicle, one behind the passenger, giving a rear view, showing head lateral position and one mounted off the rear of the driver's head restraint, giving a rear ¾ view (Figure 7). These were supported on rigid brackets.

The miniature cameras were lower resolution and had limited dynamic range, so the quality of some of the images was not ideal. The larger cameras with image stabilisation and audio data, provided further insight into the passenger's behaviour.

In addition to the cameras, and the accelerometers, a brake pedal force transducer was also fitted, together with a twin display, showing longitudinal acceleration and brake pressure for the driver. These are just visible in Figure 6.

Figure 7. Showing the rear and rear ¾ cameras, mounted on their brackets and sample views from each.

Test Procedure Outline

Most of the postures were considered to be low risk and so were safe enough for the volunteers to undertake without any type of warning. Some postures were considered too hazardous for the volunteers, but were acceptable if undertaken by project staff who understood the risks and controlled the severity of the test by instructing the driver. Some tests were considered to be too dangerous to be undertaken at all. These included: feet on fascia, drinking from a glass bottle and sleeping fully reclined whilst belted (strangulation hazard).

It was suggested that only 3 tests could be given to each volunteer before they began to suspect the reason for the test and then possibly change their behaviour. It was intended to have 50 volunteers, giving 150 possible tests. The intention was that each test would be performed several times with different volunteers to show consistency, so the total range of tests had to be limited.

Volunteers who had completed the tests were isolated from those that had not, to ensure that no "pre-warning" of events was given. A MIRA researcher was present in the rear of the vehicle to run the data-logger and to advise the volunteer of the postures required. In the early tests, a number of settling in laps were undertaken to relax the passenger so that they were less prepared for the violent pre-impact manoeuvre. Also there were a number of laps between each test for the same reason. All events were undertaken from a test speed of 50mph (80.5kph)

Although each volunteer was asked to adopt a posture, their interpretation of the posture varied, and in some cases, the posture was actually impossible (especially large male occupants who could not cross their legs above the knee). Where the volunteer adopted an unexpected posture, or misunderstood, they were not corrected (unless they asked if it was correct). This allowed the posture to be as natural as possible for the volunteer.

Methodology Development

Initial test procedure - The volunteer passengers were made aware that the driver was a MIRA professional driver. Events were undertaken in random order, but within the following schedule:

1) Briefing session
2) Lap 1 = Warm – up / settle passenger
3) Lap 2 = Warm – up / settle passenger
4) Lap 3 = Event 1
5) Lap 4 = No activity – to settle passenger
6) Lap 5 = Event 2
7) Lap 6 = No activity – to settle passenger
8) Lap 7 = Event 3
9) Return for debrief session

The methodology was altered during the testing when the early results became apparent. Some of the assumptions made regarding occupant behaviour proved to be incorrect. In particular, the volunteer's responses were affected by the many safety measures that were evident. These measures included:

- The knowledge that the driver was a professional test driver.
- The necessary process of explaining the safety aspects risk assessments and obtaining signed consent during the briefing session.
- The knowledge that the test track was a safe, test environment with wide run-off areas.

This lead to many of the volunteers assuming an un-naturally relaxed attitude, happily and confidently accepting the vehicle sliding and spinning around and treating the experience in a similar manner to a fairground ride.

Since the second and third points were difficult or impossible to work around, it was decided to modify the volunteer passengers perception of the driver.

Final test procedure A fake "driver volunteer programme" was conceived and the professional driver acted as though he was one of several volunteer drivers on this project, though this run "was his first time here at MIRA" The driver also engaged the passenger in conversation about his (bogus) job as a plumber and how he was not used to driving automatic transmission vehicles. This, together with very detailed instructions on how to drive around the track convinced all the volunteers, although some started to suspect the truth after some of the tests. Events were undertaken in the following specified sequence:

1) Briefing session
2) Lap 1 = Warm – up / settle passenger, “accidentally” over-shoot end of main straight and undertake emergency Straight Line Braking.
3) Lap 2 = Gentle Lane Change or Lane Change & Brake on main straight.
4) Lap 3 = Violent Lane Change or Lane Change & Brake on main straight.
5) Lap 3 = Direction Change or Direction Change and Lift-Off Oversteer (Spin) at end of lap.
6) Lap 4 = A half lap to return to the briefing room
7) Debriefing session

Clearly, the level of sophistication of the subterfuge was important, so a more detailed explanation of the final methodology is given below:

Lap 1 The driver was instructed to drive around the circuit at “the speed at which he felt comfortable” – which was actually gradually built up to achieve 50mph along the main straight. The passenger was asked to adopt their first posture, or not, depending if their natural posture was required.

At the start of the main straight, the confusing instruction was given to “turn right at the end of the straight”. The actual turning was just before the end of the straight so the driver deliberately overshot it and had to brake hard (straight line braking) to avoid the concrete barrier at the extreme end of the straight. The reactions from the passenger were marked – appearing to believe that the driver might crash into the barrier.

One of the volunteers with automotive industry experience realised that the braking was too good – the driver did not lock the wheels and stopped the vehicle impressively quickly and this raised some doubts. Most passengers accepted the story, many suggesting that he should be careful and not to worry about his “mistake”. The driver then continued round to start the second lap.

Lap 2 At the start of the main straight, the passenger was asked to adopt their second posture and the driver was asked to “weave gently from side to side”, sometimes with the instruction to come to rest gently afterwards. This is what the driver did and then carried on round to start lap 3.

Lap 3 At the start of the main straight, with the passenger still in their second posture, the driver was asked to repeat the weave from the previous lap “just a little more vigorously”. In fact, the driver undertook the weave and, if required, the braking, very violently, at the limit of adhesion of the vehicle. (Lane Change or Lane Change and Brake) After “recovering” from this, the driver apologised to the passenger, explaining that the power steering and the brakes were much more sensitive than he was used to! The volunteer passenger was then asked to adopt the third (and final) posture as the driver started round to start the fourth lap.

Lap 4 Since the main test manoeuvres had taken place on the broad main straight, the volunteer was expecting the driver to continue around the main outer circuit again as instructed. However, on the entry to the start of the fourth lap, the driver swerved right without warning to enter the centre section of the track, followed by either a swerve to the left or by putting the vehicle into a spin (Lift Off Over-steer). This surprised all volunteers and confused most, though some (almost exclusively the male volunteers) initially realised the truth. The majority of female volunteers still believed the false credentials of the professional driver up until the experimental debriefing.

The debriefing explained the testing and its purpose within the project and the volunteers were monitored for any signs of ill health or sickness. In fact no volunteers reported feeling unwell after the tests.

RESULTS

General Notes

The work produced large amounts of data in various forms, especially video. To date analysis of the results has been limited to identify trends and concepts to assist in the selection of critical scenarios within next stage of the PRISM project.
In total, 230 tests were undertaken, usually with 5 video clips per test. The video clips collected consist of a 3 second period before the initiation of the event, through to a “steady state” conclusion, when the event can most definitely be considered to be over. Typically, each video clip duration was between 10 and 15 seconds.

The results are split into 3 basic sections:
- Bracing incidence, using volunteer and some staff data.
- Higher risk “Out of Position” tests.
- ATD tests.

For the ATD tests simple comparisons with similar occupant tests have been made. The types of tests and the distribution of postures and vehicle motions are shown in the following figures.

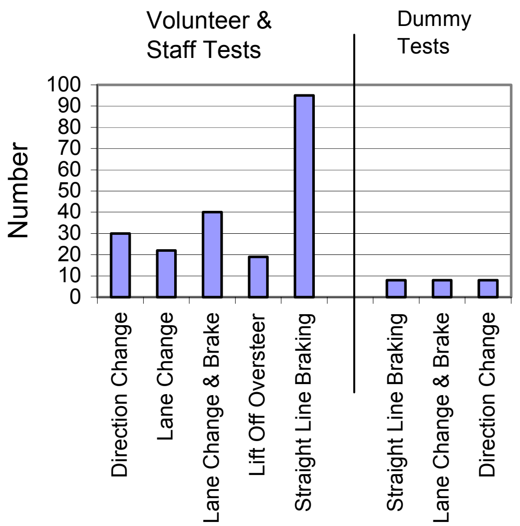

Figure 8. Vehicle Test Manoeuvre Distribution

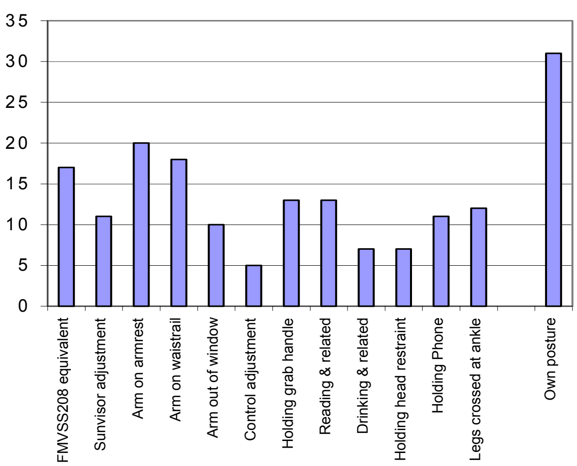

Figure 9. Volunteer Postures

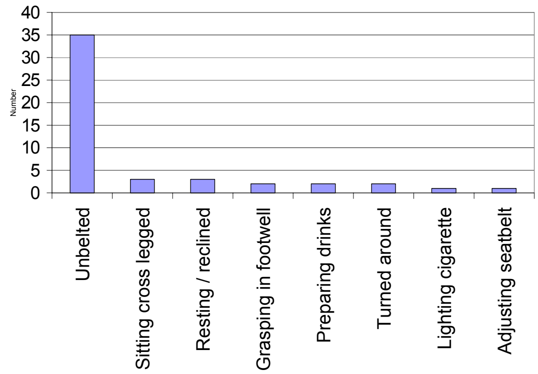

Figure 10. High Risk Postures

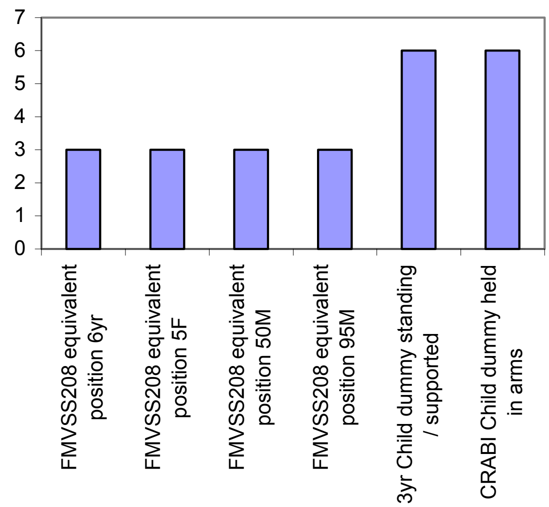

Figure 11. ATD Postures

The vehicle motions were determined by accelerometers fitted in the centre of the vehicle. The data was not corrected for vehicle pitch & roll.

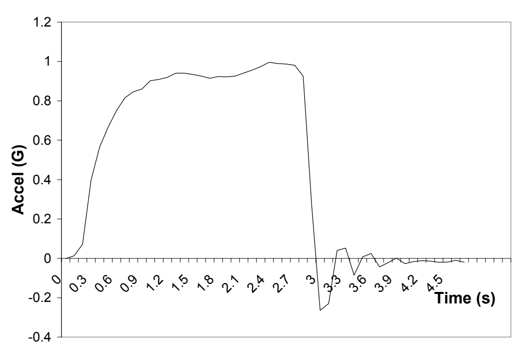

Figure 12. Straight Line Braking – Typical Vehicle Acceleration

OVERVIEW OF RESULTS & OBSERVATIONS

Bracing Incidence

The general trends discussed next are taken from the volunteers and the MIRA staff tests, with some test induced exceptions (unbelted tests etc.) where these clearly distorted the trends. The perceived levels of the validity of the tests varied depending on the confidence of the test subject. The data shown in the next section was taken only from clearly valid tests. Results were analysed by viewing the video clips and identifying reactions and limb motions. The wide range of potential limb locations were simplified for statistical purposes, concentrating on "bracing" behaviour.

Arm (and hand) locations were considered as :

- Full bracing : Hand holding on to firm structure
- Part bracing : Arm resting against firm structure or hand holding seat cushion
- Task occupied : Hand is holding an object or undertaking a non-bracing task
- Other : Generally hand on lap
- Aborted bracing : Clear case of a bracing action started, but aborted – hand remains in space.

The tables below summarise the results by vehicle manoeuvre.

Table 3.
Arm Location - Straight Line Braking

Left Arms			Right Arms		
Initial Position	**Final Position**	**Qty**	**Initial Position**	**Final Position**	**Qty**
Full Bracing	Full Bracing	7	Full Bracing	Full Bracing	
Part Bracing	Full Bracing		Part Bracing	Full Bracing	
	Part Bracing	2		Part Bracing	1
	Aborted Bracing			Aborted Bracing	
Task Occupied	Full Bracing	1	Task Occupied	Full Bracing	
	Part Bracing			Part Bracing	
	Task Occ.			Task Occ.	1
	Other			Other	2
Other	Full Bracing	3	Other	Full Bracing	1
	Part Bracing			Part Bracing	3
	Aborted Bracing	2		Aborted Bracing	2
	Other	3		Other	8
Some Bracing Effect = 72%			**Some Bracing Effect = 28%**		
Increased Bracing Effect = 22%			**Increased Bracing Effect = 22%**		

Table 4.
Arm Location – Lane Changing

Left Arms			Right Arms		
Initial Position	**Final Position**	**Qty**	**Initial Position**	**Final Position**	**Qty**
Full Bracing	Full Bracing	4	Full Bracing	Full Bracing	
Part Bracing	Full Bracing	1	Part Bracing	Full Bracing	
Task Occupied	Full Bracing	1	Task Occupied	Full Bracing	
	Task Occupied	1		Task Occupied	2
Other	Full Bracing	3	Other	Full Bracing	
	Part Bracing			Part Bracing	4
	Other			Other	4
Some Bracing Effect = 90%			**Some Bracing Effect = 40%**		
Increased Bracing Effect = 50%			**Increased Bracing Effect = 40%**		

Table 5.
Arm Location – Lane Changing & Braking

Left Arms			Right Arms		
Initial Position	**Final Position**	**Qty**	**Initial Position**	**Final Position**	**Qty**
Full Bracing	Full Bracing	5	Full Bracing	Full Bracing	
Part Bracing	Full Bracing	1	Part Bracing	Full Bracing	
Task Occupied	Full Bracing	2	Task Occupied	Full Bracing	
	Part Bracing	2		Part Bracing	1
	Task Occupied	3		Task Occupied	7
	Other			Other	1
Other	Full Bracing	2	Other	Full Bracing	1
	Part Bracing			Part Bracing	4
	Other	2		Other	3
Some Bracing Effect = 71%			**Some Bracing Effect = 35%**		
Increased Bracing Effect = 41%			**Increased Bracing Effect = 35%**		

Table 6.

Arm Location – Direction Change & Lift Off Over Steer

Left Arms			Right Arms		
Initial Position	Final Position	Qty	Initial Position	Final Position	Qty
Full Bracing	Full Bracing	2	Full Bracing	Full Bracing	
Part Bracing	Full Bracing	3	Part Bracing	Full Bracing	
	Part Bracing			Part Bracing	1
Task Occupied	Full Bracing	1	Task Occupied	Full Bracing	
	Task Occupied			Task Occupied	3
Other	Full Bracing	1	Other	Full Bracing	
	Part Bracing	2		Part Bracing	6
	Aborted Bracing			Aborted Bracing	1
	Other	0		Other	
Some Bracing Effect = 82%			Some Bracing Effect = 64%		
Increased Bracing Effect = 64%			Increased Bracing Effect = 55%		

Leg and foot locations were considered as

- Rearwards : Tibia to femur angle <= 90 degrees
- Mid : Foot on floor, tibia to femur > 90 degrees
- Forwards : Foot on toe-board / leg near straight.

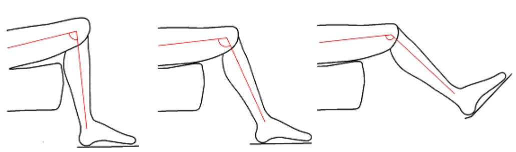

Figure 13. Leg / Foot Location Options

Table 7.

Leg Location – Straight Line Braking

Left Legs			Right Legs		
Initial Position	Final Position	Qty	Initial Position	Final Position	Qty
Rear	Rear	27	Rear	Rear	21
	Mid	5		Mid	6
Mid	Mid	26	Mid	Mid	30
	Forward	1		Forward	1
Crossed	Crossed	1	Crossed	Crossed	1
Leg Brace Movement Forward = 10%			Leg Brace Movement Forward = 13%		

Table 8.

Leg Location – Lane Changing

Left Legs			Right Legs		
Initial Position	Final Position	Qty	Initial Position	Final Position	Qty
Rear	Rear	2	Rear	Rear	2
	Mid	3		Mid	2
Mid	Mid	4	Mid	Mid	5
	Forward	0		Forward	0
Forward	Forward	1	Forward	Forward	0
Crossed	Crossed	0	Crossed	Mid	1
Leg Brace Movement Forward = 30%			Leg Brace Movement Forward = 30%		

Table 9.

Leg Location – Lane Changing & Braking

Left Legs			Right Legs		
Initial Position	Final Position	Qty	Initial Position	Final Position	Qty
Rear	Rear	7	Rear	Rear	6
	Mid	1		Mid	2
Mid	Mid	8	Mid	Mid	6
	Forward	1		Forward	0
Crossed	Mid	0	Crossed	Mid	2
	Crossed	0		Crossed	1
Leg Brace Movement Forward = 12%			Leg Brace Movement Forward = 24%		

Table 10.

Leg Location – Direction Change & Lift Off Over Steer

Left Legs			Right Legs		
Initial Position	Final Position	Qty	Initial Position	Final Position	Qty
Rear	Rear	6	Rear	Rear	5
	Mid	0		Mid	1
Mid	Mid	3	Mid	Mid	3
Forward	Forward	0	Forward	Forward	1
Crossed	Mid	1	Crossed	Mid	0
	Crossed	1		Crossed	1
Leg Brace Movement Forward = 9%			Leg Brace Movement Forward = 9%		

Extreme Out Of Position Tests

Holding Objects - The stability or motion of an object held appears to depend on 3 factors:

- The mass of the object.
- The strength of the passenger.
- The degree of extension of the shoulder and elbow joints.

The first and second of these factors were expected unlike the final point. It was quite possible for a small female passenger to hold the CRABI ATD against her chest during braking, (Figure 14) but was almost impossible for a large male to hold a full water bottle whilst drinking – moved away from the mouth and towards the airbag and a possible projectile hazard in the event of airbag firing (Figure 15). Similarly with the standing 3 year H3 ATD – whose head hit the dash panel, (Figure 16).

Figure 14. Holding CRABI ATD

Figure 15. Holding Water Bottle

Figure 16. Holding 3Year H3 ATD

Reaching Into Footwell - If extreme braking is undertaken whilst the passenger is in this position the natural reaction is to raise the head up to see the problem, (Figure 17). If the reaching activity is incomplete, the passenger may keep their hand(s) locked in whatever position it is in. Alternatively, the passenger may put a hand against the facia to push back towards the seat.

Whichever action occurs, the "peep" over the facia exposes the head and, in particular, the neck, to increased risk of injury from the passenger airbag.

Figure 17. Reaching into Footwell

Lying in Fully Reclined Seat - The passenger is unlikely to be aware of any impending vehicle manoeuvre. Severe submarining under the lap belt occurs with little restraint. The diagonal belt is in minimal contact (Figure 18). Virtually all occupant restraint is obtained by heavy knee or lower leg contact to the dash panel or glove box lid. Upon impact with the glovebox the passenger is stable under braking acceleration loads.

Figure 18. Fully Reclined

Turning Around To Rear Seat Passengers - The front seat passenger is restrained by the diagonal belt around the neck. The occupant does not tend to react other than to "freeze" in position. The occupant trajectory is unlikely to cause any problematic airbag interaction in itself but the belt loads on the neck could be considerable and painful under extreme braking (Figure 19). The additional loads caused by pre-tensioners and then by the crash deceleration could be a significant risk in this case.

Figure 19. Turning Around

<u>ATD / Volunteer Comparison Tests</u> - A series of ATDs were evaluated in similar test conditions to the volunteers. The ATDs used were : H3 5th%ile female, H3 50%ile male and 95th%ile male.

Figure 20. Comparison Of 5th Female H3 ATD With Small Female Passenger

Summary

Figure 20 shows that ATDs and human volunteers behave differently in similar straight line braking conditions. Points to note include:

- ATD torso has limited motion: buttocks remain very close to start position and upper torso rotates forward slightly.
- Human torso has more motion: buttocks slide forward more and upper body motion is exaggerated by more rotation around the diagonal belt (especially in this case with hand bracing)
- ATD head flops forward, rotating head and neck downwards; hence the gap under the chin to chest decreases with forward motion.
- Volunteer head is held upright, eyes remain level, to retain forward vision; hence the gap under the chin to chest increases with forward motion.
- The feet of the ATD did not slide forward under braking and in this respect, showed some similarity to the human volunteers.

Figure 21. Comparison of 95th Male H3 ATD With Large Male Passenger

Figure 21 shows that ATDs and human volunteers also behave differently in lateral accelerations, in this case similar violent lane change manoeuvres.
The centre images show the maximum lateral displacement during a swerve to the right. The lower images show the locations of the ATD and volunteer immediately after the swerve back to the left. The displacements are now totally different, but this represents the maximum lateral displacement point of the ATD. Points to note include:

- The ATD has particularly broad shoulders, limiting lateral motion compared to the volunteer.
- In the first manoeuvre, the ATD torso remains linear and the whole torso rotates about the buttocks.
- The volunteer spine describes a curve, so whilst the body weight is transferred to the

outer buttock like the ATD, the shoulders remain relatively level.

- The ATD head moves outboard but there is no noticeable neck bending.
- The volunteer head is held inboard, following the lateral curve of the spine, maintaining the eyes near level and retaining the field of vision.
- During the second manoeuvre, the ATD torso swings across, the buttocks remain in approximately the same location but weight is transferred to the inner buttock. Obviously there is no bracing action. The seat belt fell from the shoulder.
- The volunteer lower torso appears to have far greater lateral motion inwards, with the spine curving the opposite way but to a lesser extent. As before, the shoulders remain relatively level.
- The ATD head moves inboard and again, there is no noticeable neck bending.
- The volunteer head is held near seat centreline, following the lateral curve of the spine, maintaining the eyes near level and again, retaining the field of vision.

The ATD and volunteer postures are now different.

DISCUSSION

General Observations

Based on the results of this study front seat passengers do not tend to move their legs during pre-impact events. However, in instances where this did arise it was generally found that they move one leg forward (and occasionally outwards in lateral events).

In shorter duration events, such as Straight Line Braking, the level of leg bracing motion is low (~10%).

In the longer duration events or events where the acceleration direction changes, there is some more leg bracing motion (~20%).

In the long duration lateral loading events with no acceleration reversal, the proportion of leg bracing motion is low (~10%).

Since the test car was RHD, the passenger left hand was used more often for bracing against the door fitments.

Of the 31 "own posture" tests, only one had any bracing, which was partial with one arm. Many of the "requested" postures involved some sort of bracing with left arm/hand, resulting in a distortion of the bracing figures upwards.

There appeared to be a lower incidence of bracing increase in the shorter duration pre-impact events (Straight Line Braking).

Longitudinal Stability - A very clear and important observation from the work is that longitudinal stability for a belted occupant is heavily dependent on leg and foot location.

Bracing using arms and hands seems limited unless the passenger is already holding onto some firm structure (seat, roof grab handle, arm rest, etc.) in which case the grip tightens.

In some cases, if a firm structure is a short distance from the hand, timescales permit and the individual is sufficiently motivated, they may reach for this but sometimes the reaching motion is aborted if the diagonal belt halts the torso motion.

If both feet are forward, slightly splayed and the knees near locking point, the stability provided to the pelvis is very high. This reduces as one or both legs are brought rearwards. It would appear that one leg well braced is generally sufficient for severe emergency braking but two are better. It would also appear that one leg well braced forward is generally better than two partially braced, though more work is needed to confirm this hypothesis.

Lateral Stability - Lateral stability for a belted occupant also appears to be affected by foot position. A wide placement provides a degree of pelvis restraint, and once again, this appears better if the knees are near locked. There is very little control for the upper body however. All bracing effects are far less pronounced than for frontal decelerations. A narrow or rear foot position provides virtually no lateral motion control to the pelvis.

There is generally insufficient time to react with hands unless they are already holding the seat, door, grab handle or some other structure, so upper body motion control is almost non-existent. There is also minimal seatbelt influence with the belt type fitted to this test vehicle.

Generally inertial behaviour dominates the occupant motions in violent lateral movement, so occupant response is largely unimportant. If several cycles of reversing lateral acceleration allow sufficient cumulative time the passenger may find a suitable structure against which they may brace themselves. This was observed in one case but the motions and interactions were extremely complicated and it would appear that further work in this area would be of limited value.

Pre-Impact Braking – The Four Primary Cases

From the observations of the volunteer and the project staff tests four primary cases of importance have been identified for passenger trajectories during pre-impact braking:

Belted Occupant With Legs Braced - The occupant's braced legs prevent or limit pelvis motion significantly. This may be influenced by seat design to some degree, although it is difficult to ascertain from this project. The occupant loads the diagonal part of the seatbelt and "hangs" against it after a limited amount of upper torso motion, (Figure 22). Generally, no hand bracing is required, so if the occupant is holding an object etc., he continues to do so. If one or other hand is already bracing or is so close to bracing that contact is made this may reduce forward displacement of the upper torso slightly. However, the amount is not great compared to the effect of the diagonal belt.

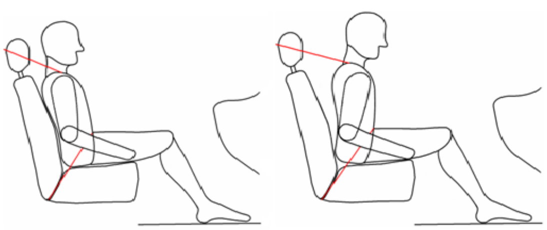

Figure 22. General Motion Of Belted Occupant With Braced Legs

Belted Occupant, Legs Not Braced - The occupant's un-braced legs do not appear to prevent or limit pelvis motion to any significant extent.

The occupant loads both the diagonal and lap parts of the belt and "hangs" against both (Figure 23). A more equal loading (than the legs braced condition) means that the torso remains more upright. Again, no hand bracing is required, so if the occupant is holding objects etc, he is likely to continue to do so. Again, if one hand is already bracing this is likely to reduce forward displacement of the upper torso slightly.

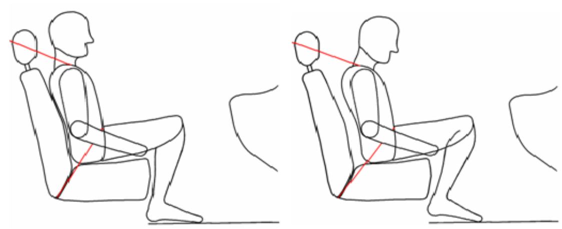

Figure 23. General Motion Of Belted Occupant With Legs Not Braced

Unbelted Occupant, Legs Braced - If the unbelted passenger is subjected to moderate to severe braking forces (up to about 5m/s/s) his braced legs may prevent pelvis motion, but are unlikely to prevent it at higher deceleration levels (above 5m/s/s) with typical seat / clothing friction. Knee impacts are unlikely except from high speeds with high decelerations or very close initial positions. Substantial upper body motion occurs and the occupant will have a definite tendency to put out hands to the dash to brace for impact at higher deceleration levels (above approx 4m/s/s). If one or other hand is already bracing this is likely to reduce forward displacement of the upper torso, possibly with yaw rotation. No unbelted trials were made with hands already occupied so that the conflict between hand bracing and continuing to hold the object was not investigated.

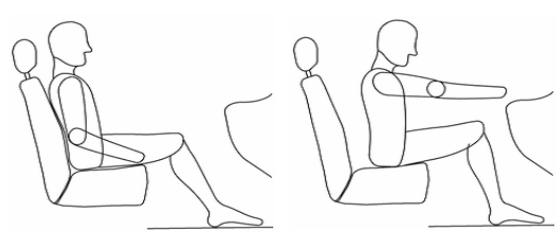

Figure 24. General Motion Of Unbelted Occupant With Legs Braced

Unbelted Occupant, Legs Not Braced - The unbraced legs of an unbelted passenger do not prevent rapid pelvis motion, with knee impact to the lower dash occurring relatively early, even at lower deceleration rates. As the femurs tend to point upwards and forwards slightly, and given the centre of mass of the body is near the pelvis, this impact condition can be quite stable, not requiring any hand bracing for stability, even though the event is so rapid that there is probably insufficient time to move the hands to the correct position, (Figure 25).

At higher decelerations a second motion begins to occur with the whole upper body rotating forwards and upwards about the knee impact point. The head and face can rapidly approach the header rail and the upper windscreen. By now there may have been sufficient time to deploy the hands to brace against the facia. This case could be important for upper torso, head, neck and hand injuries caused by a deploying passenger airbag. Ejection or partial ejection through the windscreen may also be a risk. Knee bolster airbag deployment during knee contact may also cause additional injury or promote further occupant trajectory problems.

Figure 25. General Motion Of Unbelted Occupant With Legs Not Braced

Influencing Factors on Behaviour

The test work has shown that a wide variety of factors can affect the posture of the passenger. Some of these factors we envisaged before commencement of the test work, others were not. The influence of some of the other factors was seen to be problematic and attempts were made to control these. However, this was not possible for all of the factors identified.
In an attempt to explain the various factors a schematic plan has been developed within the project. This is explained in more detail in the full report on the PRISM website (Ref 7.). In summary, the posture at impact can be considered to be the result of three phases: The first phase is the "General Posture" of the passenger, possibly modified by some pre-event activity, to give a second phase "Instantaneous Pre-Event Posture". This in turn may be modified event inertial or reaction effects to give the final phase "Instantaneous Pre-Impact Posture". These phases and the factors that affect them are described in more detail in the full report.

CONCLUSIONS

At the start of the testing the test methodology was not particularly realistic. This improved as the tests progressed. The final volunteer tests appeared more believable. It is believed that none of the volunteers acted for the cameras but their state of mind regarding their personal safety played a larger role than expected.
The following points have been determined as the most significant:

Pre-Impact Braking

Occupant trajectory during pre-impact braking is most heavily influenced by 2 pre-event factors:
- Seat belt use (or non-use)
- Foot location, especially of the most forward foot.

Bracing effects may be summarised:
- The longer the duration of the pre-impact event the more likely any bracing effect is to be undertaken.
- Changing of leg positions occurs in a minority of cases and then only one leg, always forwards.
- Bracing with arms and hands occurs in a minority of cases when belted and is most likely if already holding or close to holding a firm structure, such as the door.
- If the passenger is holding an object or is engaged in some task they tend to remain "frozen" mid task until the event is over.
- If holding an object of significant weight or not close to the body, the object's inertia will carry it forward towards the dashboard.
- The influence of bracing is greater if no seat belt is worn, when trajectories differ, especially at higher deceleration levels.

Pre-Impact Direction / Lane Change

- Occupant trajectory in violent lateral accelerations is almost entirely inertial in the initial phase.
- The head and neck tend to be maintained upright and level allowing field of vision to be maintained.
- During the first phase arm/hand bracing often occurs to prepare for the reverse acceleration.
- After the first phase some lateral leg bracing may occur if some further lateral motion is expected.

Extreme Out Of Position Tests

Each of the scenarios considered has its own hazards and problems. Each scenario should be considered based on likely incidence (from the photographic studies, or similar), risk of injury (from modelling work) and from likely cost of applying a suitable mitigating technology.

ATD Tests

The ATDs appeared to bear little similarity to the human volunteers. The adult ATDs have very stiff spines that limit motion in high acceleration cases and the lack of neck muscles frequently put the head in the wrong location and attitude. The lack of bracing means that the similarities with human volunteers reduce as the pre-impact event time increases.

Other Observations & Conclusions

It was also noted that a very wide range of variables affect a passenger's posture before and during a "pre-impact" event. The test methodology was modified to minimise the effects of unwanted variables. A general overview of all the factors and variables is given in the main report to assist in understanding the scope of the subject for further work.

ACKNOWLEDGEMENTS

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The support and assistance of the volunteers is also acknowledged. Further details are given in the report from which this paper is derived, together with other related reports on the PRISM web site at : http://www.prismproject.com.

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DEVELOPMENT AND ASSESSMENT OF A BONE SCANNING DEVICE TO ENHANCE RESTRAINT PERFORMANCE

R.N. Hardy
J.W. Watson
Cranfield Impact Centre
R. Cook
P. Zioupos
Cranfield University
B. Forrester
Autoliv
R. Frampton
M. Page
VSRC
A. Kennedy
Nissan Technical Centre Europe
S. Peach
McCue
P. Sproston
TRW Automotive
UK
Paper Number 05 –0245

ABSTRACT

The objective of the BOSCOS (BOne SCanning for Occupant Safety) project was the development of a system that can make an assessment of the bone characteristics of each vehicle occupant in order to estimate their skeletal strengths. The seatbelt and airbag characteristics can then be adjusted to deliver optimum levels of protection specifically for each occupant. A system introduced into every vehicle has the capacity to save lives and reduce injury levels across the whole spectrum of vehicle occupants. This paper describes the contributions from academic and industrial partners to this UK Department for Transport funded project.

Commercial pressure focuses restraint design on meeting legal requirements for vehicle approval, but legal requirements use dummies which do not represent the range of car occupant shapes, sizes, and driving positions. A person with lower skeletal characteristics may not be able to withstand the current fixed levels of restraint without sustaining injuries. Conversely, a person with greater skeletal characteristics may be capable of withstanding greater levels of restraint.

Possible technologies that are available have been assessed for their suitability for an in-vehicle monitoring system. Accident studies have been conducted to create a baseline of statistics in terms of casualties and their injuries. Initial bone scanning studies have utilised different types of equipment and a new prototype scanner has been developed for use in a vehicle environment using ultrasound technology.

Computer based occupant mathematical modelling has been used to establish the potential gains from a working system and also the requirements needed of the restraint systems to achieve these gains. In addition, bone scanning has been conducted, to determine a method to read across from scan values to skeletal condition to provide data for the optimisation of the restraint system.

BOSCOS OBJECTIVES

Background

Over the last decade the quest to improve the levels of vehicle safety has intensified dramatically and is now used as a sales feature by marketing departments. But as the criteria for vehicle crashworthiness have changed from vehicle deformations and decelerations to occupant related parameters (body accelerations, forces, deflections, etc.) a recognition of the implications of human diversity has been slow. This is illustrated by the fact that whilst there are child and adult anthropometric devices (dummies) available for use in vehicle testing, in the case of vehicle type approval (certification) test requirements are defined solely for a 50th percentile adult male driver representation. Consequently, it is easy to perceive that the safety systems in motor vehicles are developed, tested and approved for optimum use by a narrow band of the driver population whose physical characteristics are not representative of the whole of the driver population.

With the mass of sensors that are now beginning to appear in motor vehicles, the ability to determine information about the driver, e.g. an indication of their mass, the position of the seat and the position of the driver on the seat, is much greater. However, even those parameters that can now be quantified give only limited information that can be used to extend the narrow optimum occupant protection band to a greater proportion of drivers. To successfully extend this band we need to have more information about the individual occupants of each car if they are also to be better protected. The type of information that is needed concerns the physical injury tolerance limits of each individual so that the restraint systems can be 'tuned' by on-board processing to deliver the optimum protection for a specific crash/impact event. This means that the maximum levels of protection can be delivered for each vehicle occupant improving the likelihood, not merely of survival, but of minimal injuries.

A preliminary assessment of technologies such as a "smart personal card" or a button transponder reveals considerable opportunity for misuse and inappropriate settings of the restraint system. The BOSCOS project (BOne SCanning for Occupant Safety) focuses on the development of a fully passive system which will ideally operate without the positive action of the seat user. The BOSCOS project is a Foresight Vehicle Project funded by the UK Department for Transport (DfT).

The intention of the Foresight programme is to bring together UK resources and expertise to create components and systems for the vehicles of the future. Within this programme, the specific aim of the BOSCOS project is to initiate development of a new product that will improve vehicle occupant safety (reducing fatalities and lowering the severity of injuries) and also have a direct influence on UK Health and Societal costs (hospital costs, rehabilitation costs, pain and suffering and industry costs associated with loss of personnel).

Overview of Phase 1

In the first Phase, the possible technologies that are available were assessed for their suitability to an in-vehicle monitoring system. Accident studies were conducted to create a baseline of statistics in terms of casualties and their injuries, followed by an extrapolation of this data, taking into account the effect of technologies already in vehicles but not yet providing sufficient statistics to quantify their effectiveness. Initial bone scanning studies began to build a database for use in later tasks. Further studies established the correlation between the scanning value and bone properties and the correlation between the bone properties and bone strength.

Overview of Phase 2

In the second Phase the technology was reviewed for its use in an in-vehicle application and the actions needed to achieve this were identified and followed through to establish the methods of accomplishing the objective. Computer based occupant mathematical modelling established the potential gains from a working system but also the requirements needed of the restraint systems to achieve these gains - these will serve as part of the specification for a successful system. Further bone scanning was conducted, leading to the specification of the most suitable car occupant bone(s) that can be scanned in a vehicle environment to provide data of the best quality to the electronic control unit (ECU) for optimisation of the restraint system.

SKELETAL PROPERTIES

Existing biomechanical data relating to human bone, has shown that with old age, there are statistically significant reductions in load carrying capability, when compared with youth [1]. Yamada showed that bones were only able to resist 78% of the mechanical forces applied to them by the age of 70-79, in comparison to their peak at 20-29.

This reduction in biomechanical competence is supported by data from cadaver crash tests, which show that increasing age leads to greater probability of injury in the thorax and abdomen [2, 3, 4].

The reason for this reduction in the mechanical properties is due to a multitude of factors combining a reduction of the overall density, and structural competence (See Figure 1), combined with changes in the biochemical makeup of the bone.

The easiest parameter for assessment of bone status is the reduction in density. This is the parameter used in the clinical environment for the diagnosis of low bone density and osteoporosis. There are different systems clinically available for the measurement of the bone density, the technique considered to be the gold standard is dual energy X-ray absorptiometry (DXA). Others are available such as radiographic absorptiometry (RA), single photon absorptiometry (SPA), dual-photon absorptiometry (DPA), single X-ray absorptiometry (SXA), quantitative ultrasound (QUS), magnetic resonance imaging (MRI) and quantitative computed tomography (QCT).

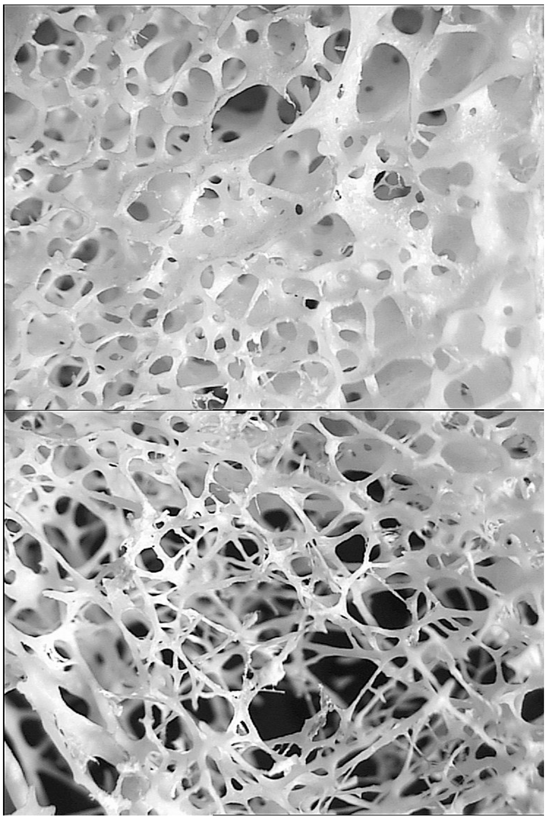


Figure 1. Bone structure of 54 year old female (top) and a 74 year old female (below), spongy bone from the hip, showing the degeneration of both the structure and density.

The ultimate aim was to establish specific algorithms and relationships between one of the clinically available techniques so as to accurately predict the condition of the bone based solely on non-invasively acquired data. Existing bio-mechanical data was used as a reference point; however it was anticipated that this was not related quantitatively to the measurements gained from the selected technologies. The aim was to complete collecting data and material after two winter and one summer seasons followed by the material studies on the collected tissue, and correlation of the material properties with the clinical work.

Non-invasive Bone Assessment

To ensure the accurate measurement of bone quality, the subjects bone needs to be assessed directly. Of the techniques mentioned previously, DXA, SXA, SPA, DPA, RA and QCT are either out-dated, too inaccurate or require the use of X-rays, and therefore contribute too great a risk to the health of the subjects. The remaining two techniques are quantitative ultrasound (QUS) and magnetic resonance imaging (MRI). The practicality of placing a MRI machine into a motor vehicle renders it unsuitable for use. The system best suited to the BOSCOS design is quantitative ultrasound (QUS).

Health Concerns

According to popular belief ultrasound is relatively risk free. However, ultrasound waves are a form of energy, and in order for the wave to be absorbed, and the amplitude reduced, this energy has to be dissipated.

The two problems arising from this are heating and cavitation. Despite mineralised bone having the highest absorption coefficient (10dB/cm.MHz) [5] the intensity level of the ultrasound used in the assessment of bone is below the levels outlined by the Food and Drug Administration as being safe from heating effects. The mechanical index indicates the risk of cavitation; the higher the mechanical index the greater the probability of a biological effect. The values published for the ultrasound of bone are between 0.22-0.28, with values below 1 considered to be safe [6].

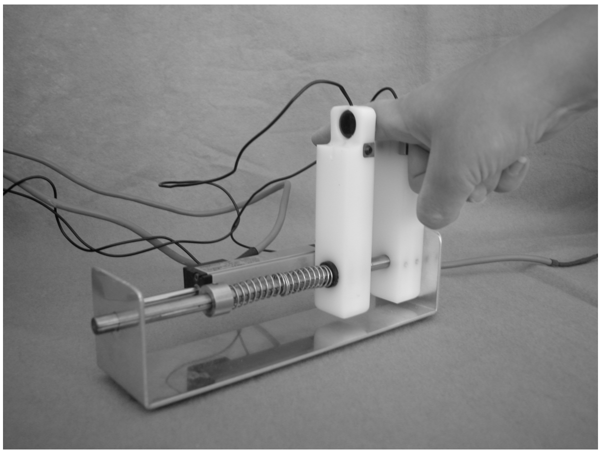
Ease of Use

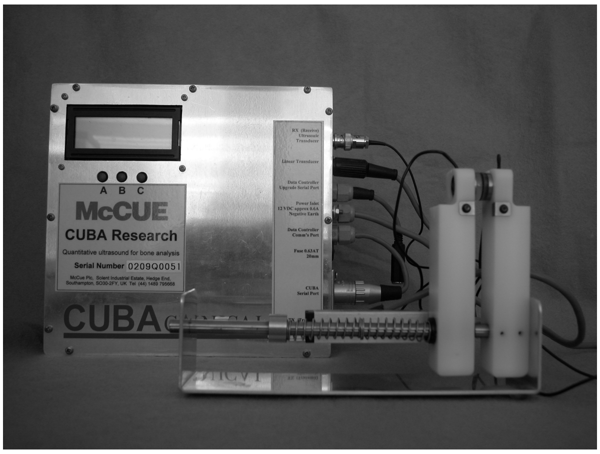
In order to ensure that occupants use the system it must cause minimal inconvenience to the driver. For this reason the BOSCOS scan needs to be preformed on a readily accessible bone site, that is generally free from both clothing and jewelry. The finger, and in particular the proximal phalanx bones, are used in clinical tests as a means of assessing a patient's bone status, and have been shown to have an ability to predict fracture risk [7, 8, 9, 10].

The BOSCOS Device

The ultrasound system has been developed by McCue plc. using technology from their commercially available CUBA Clinical™ system. The BOSCOS system is designed to measure the proximal phalanx of the index finger on the non-dominant side of the subject (See Figures 2 and 3). The system works by positioning two ultrasound transducers either side of the finger and an ultrasound pulse is transmitted between the two transducers through the finger. The system takes a measurement of the separation of the transducers and the time taken for the ultrasound pulse to travel this distance. From this information, the speed of sound can be calculated. The speed of the ultrasound pulse is affected by the quality of the bone it passes through, with good quality bone enabling the pulse to travel faster.

The BOSCOS system compared the newly measured speed to a reference database, allowing for a quantitative evaluation of the subject's bone status in comparison to an expected normal. When the result indicated the subject's measured bone speed of sound was below normal, the subject is deemed to have low bone quality and was therefore at higher risk of sustaining a fracture.





Figures 2 and 3. The BOSCOS Ultrasound Device.

Initial Results

The best we can aim for is for the prototype to perform as well as the commercially available portable QUS scanners. We have therefore conducted extensive studies on the precision/accuracy and the sensitivity and specificity of two commercially available QUS scanners, the Sunlight Omnisense and CUBA Clinical along with the BOSCOS prototype.

The precision error of a bone scanning technique refers to its ability to produce the same result, when no change, apart from re-assessment, has occurred. [11] For the BOSCOS system the precision error needs to be minimal to ensure repeat measurements do not cause different restraints reaction scenarios. The perfect technique would present a precision error of 0% to show that measurements had no difference between them. Assessment of precision error showed the commercially available finger scanner was capable of a precision error of 0.55%, in comparison to the other techniques that ranged from 0.29-2.88%.

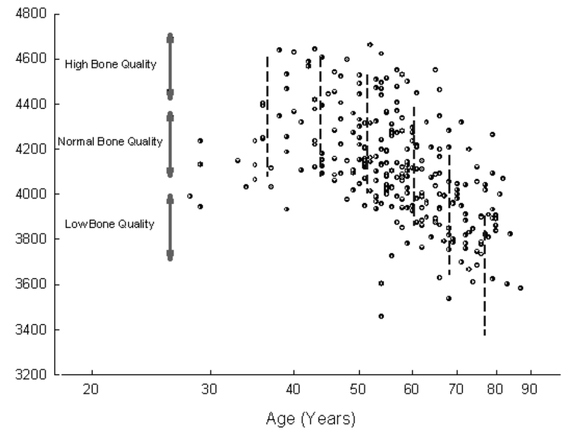


Figure 4. The speed of sound (SOS) measurement values from a commercially available finger scanner versus age for 295 volunteers, showing how the system could be used to sub-classify the population into at least three groups.

Using data obtained from 295 subjects, the finger showed the highest correlation with age 0.533 (p value < 0.001) (See Figure 4). The p-value is the level of statistical significance; a value below 0.05 (95% confidence) is considered to be of statistical significance.

The prototype system was used along side DXA assessment of the total hip and lumbar spine (Hologic QDR-4500C; Hologic Inc. Bedford, MA, USA); QUS assessment of the calcaneal (heel) bone (CUBA Clinical; McCue plc. Winchester, UK), proximal phalanx and distal radius (Sunlight Omnisence; Sunlight Medical, Rehovot, Israel), in a study on a group of 102 subjects (7 males, 95 females) aged between 24 and 85 years of age (mean: 57 years). The correlation between the new phalangeal assessments and age gave a correlation of r = -0.597 (p value < 0.001), and regression analysis (See Figure 5) gave the relationship:

Phalanx SOS = 4604.66 – 9.15609 age
$R^2 = 35.7\%$

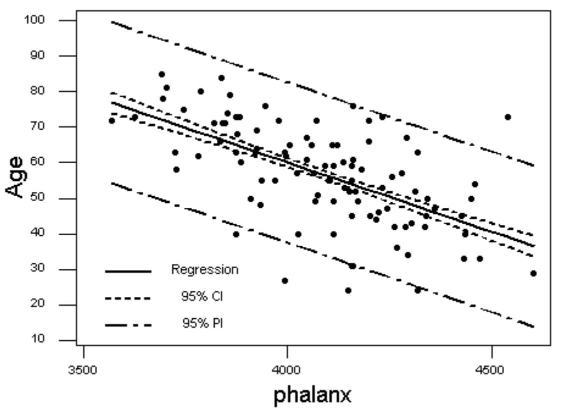


Figure 5. Regression plot of age vs phalangeal SOS

Not knowing the actual condition of the bone the performance of the prototype was assessed against a 'composite' parameter by combining the average of scaled values of the CUBA clinical, Sunlight Omnisense and DXA.

For each individual patient the proximal phalanx of the index finger was assessed using the BOSCOS system, and 10 waveforms with results were saved (See Figure 6).

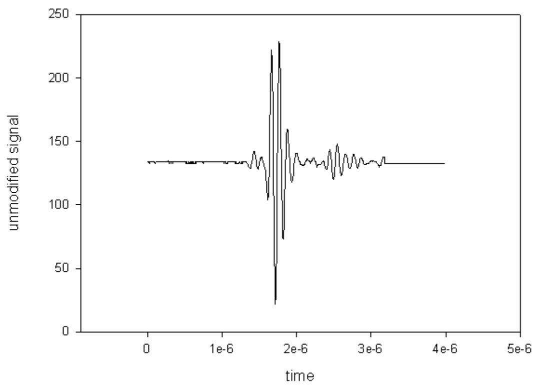


Figure 6. A representative ultrasound pulse after transmission through bone.

The pulses were converted to absolute values in relation to the baseline and the time and amplitude of the four greatest peaks was noted (See Figure 7). The ultrasound pulse was analysed by retrieving information about:

- The time incident of the first of the four greatest peaks assessed.
- The time and amplitude difference between the first and second peaks.
- The time between the first and fourth greatest peak
- The area under the waveform.
- The amplitude of the biggest of the four peaks. (The maximum amplitude)

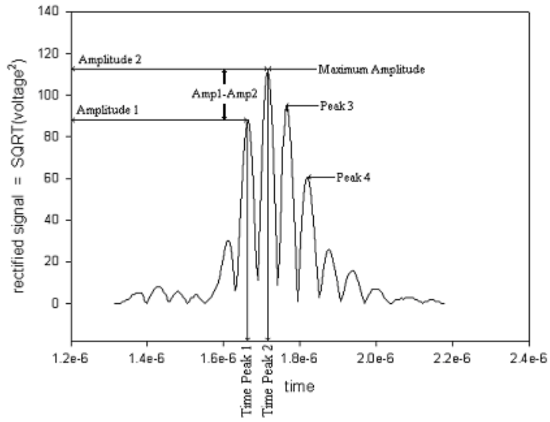


Figure 7. The 'extrapolated' parameters produced from the positive waveform.

These parameters were used alone and in combination with weight and age for the assessment of the composite measure.

The results showed that a combination of the ultrasound parameters with weight and age enabled the BOSCOS system to predict the status of a persons bone with an R^2 of not better than 50% (the R^2 represents the coefficient of determination, which is a measure of how well the regression model defines the data). However, by making use of available superior technology, the predictive ability of the system may well be improved, which could enable the differentiation of individuals into groupings according to their bone status. Further work is required to enable an understanding of what the measurement value (taken from the phalanx) means, in terms of actual bone properties, with respect to the rest of the skeleton.

REAL-WORLD INJURY ASSESSMENTS

Real-world data

The primary reasons for the use of accident data in BOSCOS were to identify the types of crashes and occupants who would most likely benefit from the system in order to address conditions where real people in real crashes were being injured. The accident data was to provide a basis for the modelling of the current injury situation. This baseline was the starting point from which to assess the potential effectiveness of modifying restraint system performance parameters based upon an estimate of occupant skeletal strength. The real-world data used within the BOSCOS project was collected by the UK Co-Operative Crash Injury Study (CCIS), which samples accidents based on vehicle age, vehicle damage and injury outcome. To be included in the database, the accident must have included at least one car that was at most seven years old at the time of the crash, was towed away from the accident scene and in which an occupant of the car was injured. The data are also collected within a stratified sample which is biased towards 'fatal' and 'serious' injury outcome crashes. Of all crashes occurring in the geographical sampling regions, approximately 80% of all fatal and serious accidents, and 10-15% of slight injury crashes were investigated.

Because of the bias within the CCIS data towards serious and fatal injury outcomes, it was necessary to weight the data so that it was representative of the whole population of injury, tow-away accidents. To do this, weighting factors were calculated which correct the under-representation of slight injury accidents.

Problem Definition

Impact Type

During the initial stages of the project it was intended to examine as many different impact types as possible. 73% of belted front seat occupants who sustained an AIS 2+ injury were involved in either frontal or side impacts. Impacts such as rollovers and under-runs were not considered, since not only is occurrence of these impact types low, but mathematical modelling of such impacts is very difficult due to their inherent variability.

Although side impacts make up around 23% of injured (MAIS 2+) occupants, it was decided not to attempt to apply the BOSCOS system to side impacts at this stage for the following reasons:

- Side restraint systems have far less time in which to operate, hence the extent to which their deployment can be adjusted to differing scenarios and occupant types is limited.
- Due to it's retrospective nature, the CCIS data contained relatively few crashes with cars fitted with side airbags, hence making an assessment of the effectiveness of the BOSCOS device compared to current technology is difficult.

Therefore, at this stage it was decided to restrict the investigation to frontal crashes only, which still covered 57% of the occupants in the database. However, it was anticipated that the application of BOSCOS to side impacts could provide the basis for further development work in the future.

Body Regions and Types of Injury

The next stage of problem definition was to identify the body regions and types of injury that were most likely to be mitigated with the introduction of a BOSCOS system. Since the basis of such a device was to adapt the restraint system according to the skeletal strength of the occupant, it follows that skeletal injuries are those most likely to be reduced. Obviously a reduction in skeletal injury resulting from "softer" restraints is also likely to be accompanied by a reduction on the occurrence of soft tissue injuries, although the exact influence on these types of injuries will be harder to determine.

Figure 8 shows the location of skeletal injuries for belted drivers with airbags. It is clear that the body regions of concern in this context were the chest and upper and lower extremities. Since injuries to the chest are likely to pose a higher threat to life than those to the extremities, chest injuries provided the focus for the initial development of BOSCOS.

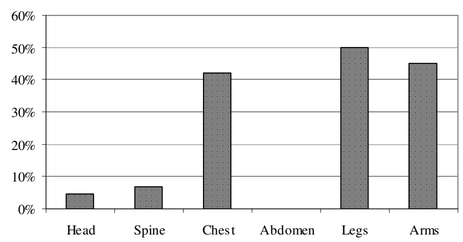


Figure 8. Location of skeletal injury for belted drivers with airbags.

71% of all serious (AIS 2+) chest injuries for belted drivers were fractures to the ribs or sternum. Of these skeletal injuries, 66% were considered to have been caused by the restraint system (either belt or airbag), whilst 53% of all AIS 2+ chest injuries were attributed to the restraint system. In crashes where the crash severity is known, as determined by an ETS calculation, 75% of injuries occurred at speeds lower than 56km/h, the current basis for legislative testing. ETS is the vehicle delta v, calculated on the assumption that deformation was caused by impact with a fixed rigid barrier [12]. Since 96% of these cases below

56km/h sustained little or no facia intrusion (<4cm) it is clear that there is the potential for an adaptive restraint system to provide significant benefit to chest injury risk.

Occupant Types

It is widely accepted that human bone strength decreases with age, and as such it is expected that the benefits of a BOSCOS system will be of greater magnitude to the elderly. With the aging population of the UK, the societal benefit as a whole will increase as more and more older drivers and passengers become exposed to the increased risk of injury attributable to a decrease in bone strength.

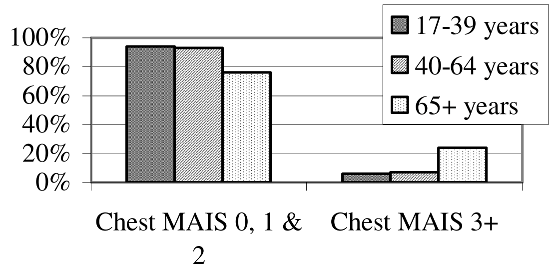


Figure 9. Distribution of maximum chest AIS of belted drivers by age group.

Figure 9 shows the distribution of maximum chest AIS for belted drivers of varying age groups. It is apparent that injury risk remains constant for the 17-39 and 40-64 age groups, but that there is a clear shift towards more AIS 3+ injuries for the 65+ age group However, it is expected that a BOSCOS system will also be of benefit to younger occupants.

Although risk of chest injury in AIS terms is similar for ages 17 to 64, a number of clinical studies [13, 14] show that morbidity from rib fractures can increase from a much younger age, possibly as young as 40 onwards. As such, although the risk of specific injuries may not increase in the 40-64 age group, the risk of complications and associated increased costs of treatment (and ultimately cost to society) can increase.

The ability of the BOSCOS system to measure bone strength means that sufferers of conditions such as osteoporosis will be detected and the restraint system tailored to them as much as is practicable.

Development of Accident Matrix

Analysis of the real-world data presents an obvious target group, for which a BOSCOS system should provide an improvement in occupant protection. This group was broadly defined as belted drivers and front seat passengers in vehicles fitted with pre-tensioners and who sustained an injury attributed to the restraint system. Whilst it is likely that others outside this target group would also benefit from BOSCOS, this group was the most appropriate on which to base the next stage of the work – development of a matrix of accident scenarios.

One of the limitations of mathematical modelling is that models have to be validated by full-scale crash tests to ensure that the results produced are valid. Since the motor industry has a need to optimise performance for legislative and consumer tests, there is no guarantee that extrapolating the models outside these types of impact will produce valid results. For this reason, the BOSCOS target group was categorised into the following impact types:

- Full overlap – This type of model will be used to represent all the real-world impacts with an overlap greater than 85%. The ETS selected for this group were 25km/h and 45km/h, since these were the 25% percentile and 75% percentile respectively of the real-world full overlap crashes.

- Offset – Since an offset test is designed to test the crash performance assuming that one longitudinal member absorbs the majority of the impact energy, this type of model will represent all real-world impacts with an overlap up to 55%. However, 90% of the real-world offset crashes fell between 23km/h and 33km/h and therefore a median speed of 28km/h was chosen to represent this group. Impacts to poles and trees only represented 4% of the BOSCOS group. The data was insufficient to develop a scenario for modelling. Improving safety in a small offset impact should, however, also address some of these narrow object impacts.

- The remaining group consists of crashes where only one of the vehicle's longitudinals was directly loaded, but a significant proportion of the energy was absorbed by loading of the engine block.

In effect, a wide overlap impact but directly impacting only one longitudinal. An overlap of 75% and ETS of 40km/h was deemed suitable to model this group of crashes.

BASIS OF COST BENEFIT STUDY

Background to Cost Benefit Study

In order to assess the potential benefits of BOSCOS, it was necessary to evaluate changes in injury risk and their associated costs. In this way, any benefits can be shown clearly as monetary values, which are directly comparable to costs incurred by proposed BOSCOS systems.

'Willingness to Pay' Approach

Several cost benefit scales were considered including the HARM concept developed in the US by Malliaris et al in the early 1980s [15] and Miller et al, 1991 [16]. HARM was considered inappropriate for use in BOSCOS because injury costs in Europe do not exist in a form usable by HARM. For this reason, it was decided to consider the 'Willingness to Pay' approach, which was developed by the UK Department for Transport (DfT) to calculate costs of injury in the UK.

The Willingness to Pay approach to injury costing was first used in 1988 by DfT to value the cost of road accident fatalities. The concept behind it is to consider what people would be prepared to pay in order to reduce the risk of being killed in a road accident. According to TRL Report 163 [17] this approach is 'consistent with cost benefit analysis, in that decisions reflect the preferences and attitude to risk of people who are likely to be affected by them.' In 1993 the same method was used to revise the values for non-fatal road accidents and in 1994 other accident costs were also derived. There are two areas of costs which have been defined; casualty related costs which include lost output, human costs and medical and support costs and accident related costs which encompass property damage, insurance administration and police costs.

Severity of an accident is defined as fatal, serious or slight. A serious injury is defined in TRL Report 163 as covering a wide range 'from a fractured finger, to those resulting in severe permanent disability, or death more than 30 days after the accident.'

Serious injuries were divided into sub-groups according to treatment length, extent and duration of pain and recovery time.

Table 1.
Injury State Descriptors, Hopkirk & Simpson, 1995

Injury Code	Injury State
F	Recover 3-4 months (Out-patient)
W	Recover 3-4 months (In-patient)
X	Recover 1-3 years
V	Mild permanent disability (Out-patient)
S	Mild permanent disability (In-patient)
R	Some permanent disability with scarring
N	Paraplegia/Quadriplegia
L	Severe head injuries

The Willingness to Pay approach was implemented to determine the 'human cost' of an accident. A Standard Gamble questionnaire was used to carry out a survey of 450 people, asking them how much they would be willing to pay to reduce the risk of injury, relative to the cost of a fatality.

The respondents ranked the injury states and placed each one on a scale from 0-100. The majority regarded injury state L as being as bad as or worse than death and injury state N as only slightly better than death. The respondents were also asked to specify the level of risk at which they would opt for treatment of an injury. It was then possible to convert the survey results into values relative to the value of death and as a percentage value of death. Therefore the human cost of each injury state can be expressed as a percentage of the human cost of a fatality. The cost for a slight injury, including whiplash, has also been determined.

New injury costing method – VSRC

Medical researchers at the VSRC have mapped 300-400 trauma injuries from the CCIS database from the AIS level (AIS 1990 revision), [18] to the injury states defined by Hopkirk and Simpson in TRL Report 163. This enables the calculation of the human cost of a trauma injury according to its AIS code. In TRL Report 163, complete lists are given for slight and serious injury costs as a percentage of the overall value of a fatality in 1994. The 2003 figure for a fatal casualty is given in Road Casualties Great Britain 2003: Annual Report and therefore all 2003 human costs for an injury can be calculated [19].

Cost benefit calculations

Using the injury costs defined by the VSRC, it is possible to give a monetary value to reductions in injury risk achieved by the BOSCOS project.

For example, if simulations are performed using increasing load limiter settings, on a strong and a weak occupant (in terms of skeletal strength), then the different chest injury risks can be assessed for each occupant using appropriate risk curves. The risk of head injury with the differing load limiter values can also be simulated. The costs can be derived for each type of injury, according to occupant strength and load limiter. The optimum load settings can then be determined for each type of occupant depending on skeletal strength. Using the proposed BOSCOS system, it would be possible to adjust the level of the load limiter as required, depending on what is most beneficial in terms of occupant injury, therefore reducing potential injury costs.

NEW TECHNOLOGIES OF RESTRAINT SYSTEMS

The present protection system on front seats features a belt system, incorporating a pretensioner, a load limiter, and an airbag. This protection system is not capable of changing its performance characteristics during a crash event. The ability of an occupant protection system to adapt itself to dominant crash condition parameters, such as impact speed and type, occupant size and mass, bone characteristics offers a great improvement in occupant protection for a wider range of crash conditions, as well as occupants.

New technologies are being rolled out to address these issues. These technologies will require new sensors in order to detect certain parameters e.g. the BOSCOS scanner and new actuators in order to protect the occupant.

The car occupant restraint industry has so far mainly focused on “In-crash systems” aimed at mitigating the consequences of an accident. However, for example, Autoliv’s Total Safety System concept has widened the scope of safety enhancing areas to include both “pre-crash systems” and “post-crash systems”. The pre-crash systems are often active systems that are aimed at preparing the safety systems for an imminent crash or, preferably, avoiding the crash altogether. Post-crash systems are devised to increase the occupant’s chances of surviving after a serious accident.

Components and sub-systems must therefore be designed to interact with each other as one system. Seat belt pretensioner and frontal airbags, for instance, are tuned to complement each other via the same electronic control unit to give the best possible protective effect. In addition, the deployment of the frontal airbags should be adjusted depending on crash severity, seat belt use and occupant characteristics.

Future restraint systems should provide protection for all kinds of occupants in various seat positions with or without seat belts (infants, elderly people, petite females, and large males).

In real life, crashes are almost never "head-on" frontal collisions into a rigid unmoveable object at one specific speed (as in most crash tests required by the government regulators). Consequently, future safety systems should be able to do more than just determine if an accident is a frontal crash, a side impact, a rear-end collision or a rollover.

An ideal system should be able to identify and provide protection to car occupants in collisions with various types of vehicles and objects (car-to-car, car-to-truck, etc.) up to a collision speed where there is still a survivable space in the vehicle’s compartment. New technologies may include the concepts described below.

Smart Seat Belt

In a crash, a smart belt starts by tightening the belt, using a pyrotechnic pre-tensioner. This eliminates slack and makes it possible to release some webbing at a later stage, if the load on the occupant becomes too high. The airbag is instead used to absorb more load.

In a traditional system, the loads to the occupant from the seat belt and the airbag are added to each other, when the airbag also starts to restrain the occupant. But in the smart belt, the system just shifts into the second lighter gear so that the load on the occupant’s body can be maintained at a relatively constant level.

Equally important is the fact that the force of the combined systems – and thus the load on the occupant – can be tuned to the severity of each crash. Many future vehicles will have advanced occupant weight sensing systems. In those vehicles, a smart belt could be tuned to each occupant individually. This will be particularly important for occupants who are more susceptible to high chest loads.

Pre-Pretensioning

The pre-pretensioner will give a more gentle load distribution on the occupants chest in the event of a car crash. The device will tighten the seat belt as early as one tenth of a second before a likely crash, using a fast electrical motor.

The elimination of slack in the belt system can therefore start earlier, even before a crash and the system can be made reversible. Consequently, it is possible to "strap in" the occupant more gently. It also makes it possible to tighten the belt, as a precaution when it is difficult to predict whether there will be a crash or not. The new system will be especially effective in preventing occupants from being thrown forward during severe braking.

Pre-Crash Sensing

In a few tenths of a second before a crash, radar sensors are capable of identifying the relative speed towards an object and the estimated time of impact. This will allow better discrimination of the crash severity and events identified in the BOSCOS accident studies.

Secondly, this will enhance the detection capability and timing of existing safety systems, particularly for relatively small, narrow objects, such as a corner of another vehicle, or pole or lamppost. The pre-crash sensing system will be especially useful in combination with pre-pre-tensioning.

Even if this pre-crash system gives just a few more milliseconds to inflate the airbags, it could open the possibility to make the airbags “softer” during deployment without compromising their protection capability.

PARAMETRIC MODELLING

In phase 2 of the project a series of mathematical modelling parametric studies were conducted to investigate different accident scenarios. The different scenarios were generated from the accident analysis performed by VSRC and have identified crash configurations where there are AIS 2+ chest injuries attributed to the seat belt. These injuries are in the form of broken bones as well as other soft tissue injuries. Dummy models were used to develop a generic seating and interior design to enable comparisons between different models to be evaluated.

The dummy models are able to predict the levels of acceleration, belt loads and trajectories of certain body parts. For each different configuration these criteria indicate the severity of the crash pulses. Parameters such as the seat belt tension and pre-tensioners were incorporated into the model to represent the range of safety restraint systems which are currently available. An airbag was included in the model, as they play an important role in the protection of vehicle occupants.

Initial simulation results with the selected accident scenarios predict injury indices below those allowed in the higher speed legal or EuroNCAP tests.

CONCLUSIONS

The BOSCOS project to-date has set out to identify the best means of calculating the bone strengths of vehicle occupants. The ultrasound technology has been selected as the most effective and safe tool to use and highlighted its benefits through scans of human subjects. Different ultrasound devices have been evaluated and a new prototype devise has been built which could be adapted for in-car use. Real world vehicle accident data has been assessed to determine which accidents are causing rib fractures. New restraint technologies have been identified which could be enhanced with the addition of BOSCOS type technology. A number of accident scenarios have been selected and they have been used in the initial mathematical modelling.

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FUTURE WORK

In the last phase of the BOSCOS project the technical issues that need to be addressed in the use of the bone scanning technology in a vehicle will be investigated to provide input to the development of the system. During the course of this Phase this process will be reviewed as other tasks define particular aspects of the technology. Final bone scanning will be completed leading to the definition of the bone property ranges that can be successfully identified by the scanning techniques chosen. A study will establish the sensitivity of the scanner device in a vehicle environment as influenced by factors such as the bone selected for scanning, the possible locations of the device in the vehicle, ambient conditions in the vehicle and occupant diversity. Mathematical modelling will predict occupant injury indices with the new technologies. A cost benefit study will utilise these results to deliver an indication of change in injury risk and the potential gains from a BOSCOS system.

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Protection of Rear Seat Occupants in Frontal Crashes

Suzanne Tylko
Dainius Dalmotas
Crashworthiness Division,
Transport Canada
Canada

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ABSTRACT

Regulations and vehicle design optimization have traditionally focused on the occupants of front seats. Stringent requirements exist for the driver and front right passenger but there are no dynamic crash test requirements for rear seat occupants. The introduction of frontal airbags and the concurrent increased incidence of child fatalities in low speed frontal collisions brought urgency to the public health message advising parents to place children 12 years and under in the rear seat.

Anthropometric test dummies representative of a 5th percentile adult female or 12-year-old child were used together with the recently introduced Hybrid III 10-year-old dummy and the Hybrid III 6-year-old dummies to evaluate rear seat occupant protection in full frontal rigid barrier tests and frontal offset deformable barrier tests. The 6-year-old dummy was restrained with a belt-positioning booster while the 10-year-old was restrained with either a belt positioning booster or the vehicle 3-point seatbelt. Dummy responses were examined as a function of seat position and in the case of child dummies, booster seat type.

Successful restraint of the chest was associated with high belt loads and pronounced chest deflections while slippage of the belt from the shoulder led to extreme flexion of the torso, head strikes and elevated neck loads. Booster seats had no effect on shoulder belt translation during the dynamic event but were observed to maintain the abdominal portion of the belt in place, over the pelvis. Opportunities for rear seat occupant protection and child dummy enhancements are discussed.

INTRODUCTION

In 2003 Transport Canada began adding dummy occupants to the rear seating positions of vehicles being tested in Full Frontal Rigid Barrier (FFRB) compliance tests (CMVSS 208) and Offset Deformable Barrier (ODB) research tests to evaluate rear seat restraint performance. Testing began with an evaluation of booster seat performance for the Hybrid III 6-year-old dummy and evolved to include a comparison of the Hybrid III 5th female dummy seated in the front and rear seats of different vehicle models. The Hybrid III 10-year-old dummy, being a relative newcomer to the family of frontal dummies, was included in a small number of tests to try and gain a better understanding of this dummy's attributes. Furthermore, because this dummy represents an older child it provided an opportunity to compare the response of a dummy restrained with the lap/shoulder belt with and without of a booster seat.

The paper will first describe the results of the small female as this will quantify the differences observed between the front and rear seats in the context of a dummy that is familiar to most and highlight certain key measurement parameters. Trends and responses observed with the Hybrid III 6-year-old will be presented and compared to the responses of the 6-year-old in two modified restraint configurations. The results of the paired comparison for the 10-year-old with and without a booster seat will complete the analysis and illustrate the importance of including a suite of measurement parameters in the dynamic testing of rear seat performance.

TEST MATRIX

The vehicle sample for this study included 77 vehicles shared with the frontal compliance test programme and the frontal protection research programme. The vehicles were of model year 2003 through 2005, included passenger cars, minivans, crossover vehicles and SUV's ranging in test mass from 1400 to 2900 kg. All were equipped with a three-point lap shoulder belt in the centre rear seating position and the majority had LATCH anchors in place.

The child restraints were purchased from local retail outlets and were selected on the basis of seat geometry, advertised weight limits, tether attachment and internal harness configuration and belt guide design. In vehicles where three adjacent booster seats needed to be fitted to the vehicle, selection was based exclusively on the width of the seat as even in the largest of SUV's, the fitment of three child restraints across one bench seat was found to be a challenge.

Selection of the child seat for a particular test was dependent on seat placement and intended comparison. For example, if two outboard positions were being compared and the test was a FFRB with evenly distributed loads to the front of the vehicle, the options were either to select two identical seats and vary the attachment configuration or to select two different seat types but retain identical attachment methods. A number of comparisons were carried out including:

- Centre rear VS rear outboard with identical booster seats;
- High back booster with tether/latch attachments VS high back with only the lap/shoulder belt in two outboard positions;
- Second row VS third row with identical booster seats.

An overview of the different restraint types and the attachment configurations employed in the study are presented in Table 1. The sample contained a total of 20 high back models and 2 low back models. Each of the above comparisons was carried out with a minimum of 2 different model types. The booster seats with LATCH and tether attachment were convertible child seats. These are forward facing child seats, which can be converted to a booster seat by removing the harness.

Table 1: Test Matrix with Hybrid III Child Dummies and Small Female.

	6 Year Old	10 Year Old	5th Female
CRS	X		
Low back booster	X		
High back booster	X	X	
High back booster with harness latch & tether	X		
Lap & shoulder belt	X		X

TEST METHODOLOGY

Vehicle preparation was conducted as per the FMVSS/CMVSS 208 procedure for the full frontal rigid barrier tests. A small number of low and high speed offset deformable barrier (ODB) tests were also conducted by following the ODB procedure in FMVSS 208. Tests for the small female front to rear comparison and for the 10-year-old were carried out at 48 km/h FFRB, while tests with the 6-year-old were conducted at speeds ranging from 40 km/h to 60 km/h in FFRB and ODB.

Positioning Procedures

The Hybrid III 5th female dummy was seated in the front right passenger seat as per the FMVSS 208 procedure. In the rear seat, since there is no regulatory procedure in place, the dummy was seated in the rear right passenger seat by aligning the mid-saggital line of the dummy with the seat centerline. The head level and thorax orientation of the dummy were dependant on the vehicle seat back angle and the arms and legs were placed in a neutral position. A piece of surgical tape was placed on the dummy thorax to record pre-test shoulder belt position and to assist in identifying belt position at peak load with the indentations left on the tape.

The Hybrid III 6-year-old dummy was placed in a booster seat and restrained with the vehicle seatbelt. The seatbelt was deployed, as one would expect a child to deploy the belt that is without engaging the locking mechanism. The booster seat itself was either used as a traditional booster seat using the vehicle seatbelt alone or by attaching the booster seat to the vehicle seat with the LATCH and upper tether anchor and then using the vehicle seatbelt to restrain the child. When used as a forward facing child restraint the seat was installed in the vehicle as per manufacturer's instructions and the upper tether always used.

Instrumentation and Filtering

The dummy instrumentation included a tri-axial accelerometer at the head CG, a 6-axis load cell at the upper and lower neck; a potentiometer at the sternum, a lumbar load cell and numerous linear accelerometers on the sternum, spine box and in the pelvis. All data recording and filtering was performed in accordance with SAE J211.

Rear occupant motions were filmed at 500 to 1000 frames/second, depending on light quality, from the front and side.

RESULTS

Small female front and rear. The vehicles used for this comparison included a Japanese mid-sized sedan, a small SUV and a large SUV/truck. The two females were seated one behind the other with the front female seated in the foremost track position. The mid-sized sedan (Model 'A') was the only test where the rear occupant femurs were observed to contact the seat back. The relative timing of the dummy kinematics was such that the effect of this loading on the front dummy responses was undetectable in the front occupant lumbar or seatbelt force responses.

The responses of the Hybrid III 5th female seated behind the right front passenger were more elevated than those of the front passenger in all three FFRB 48 km/h tests. Kinematic responses sharply contrasted the kinematics of the front occupant as the abdominal belt translated up and penetrated the abdominal cavity in two of the three tests.

The comparison of sternum deflections, 3 ms chest clips and the corresponding seatbelt forces for the front and rear dummies are presented in Figures 1 through 3 respectively. The order of presentation is the same for all plots such that the sedan is Model 'A' and is always the first pair of bars; the large SUV/truck or Model 'B' is represented by the middle bars and the small SUV, Model 'C', is the final set.

Chest deflections were always greater in the rear seat. The difference for Model 'A' was of the order of 7 mm; however in Model 'B' deflection increased from 17 mm in the front to 41.7 mm in the rear. Similarly in model 'C' deflections in the rear almost doubled by increasing 22mm over the front passenger response.

The 3 ms chest clips were also more elevated in the rear, though the increment did not reflect the deflection trends. For example, in Model 'B' where deflections more than doubled, the clipped chest accelerations were essentially identical.

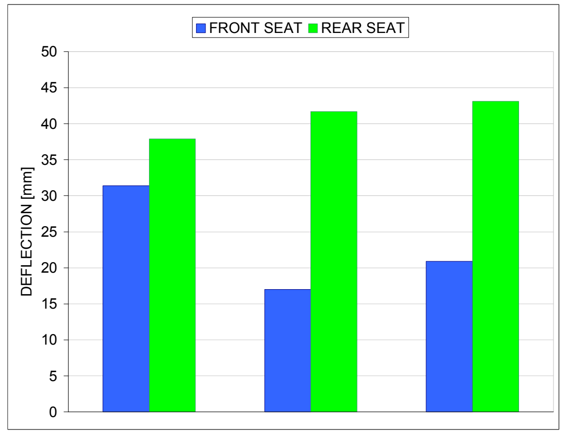

Figure 1: Deflection measures for the front & rear seat occupant in Models A, B and C respectively.

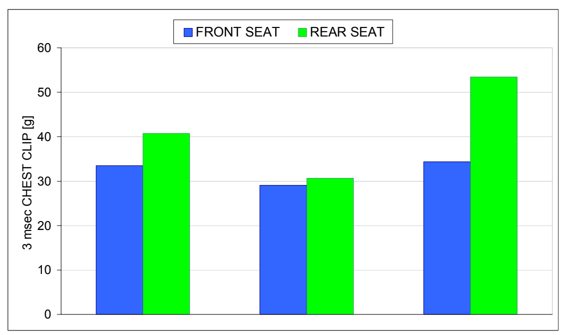

Figure 2: Chest clip for the front and rear seat occupant in Models A, B and C respectively.

Both lap and shoulder belt loads increased in the rear seat position, as did the proportion of shared load between the lap and shoulder belt. In Model 'A' the lap and shoulder belt loads are comparable but in the rear seat the relative distribution of load changes rather significantly as the shoulder belt attains three times the peak load of the lap belt. A similar change but in the reverse was observed in Model 'C' where both the shoulder and lap portions of the seatbelt attain values that are of the order of 7 kN.

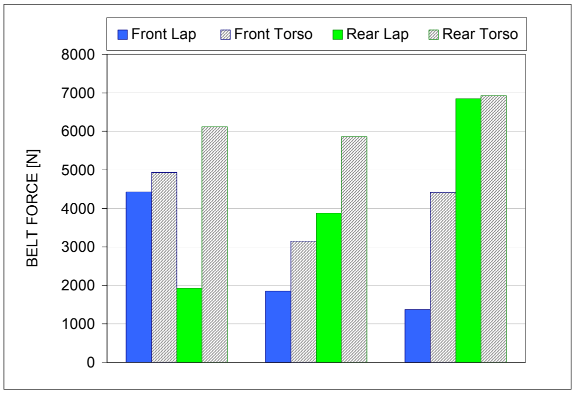

Figure 3: Lap and shoulder belt force measures for the front and rear seat occupant in Models A, B and C respectively.

Video analysis of the rear passenger kinematics suggests that lap belt migration into the abdominal cavity is most pronounced in Model 'A' and occurs in Model 'B' but to a lesser extent. In Model 'C' the lap belt appears to remain in place as the belt is loaded. These findings are consistent with the resultant lumbar forces shown in Figure 4. The elevated lumbar force resultant in Model 'A' correlates well with belt intrusion into the abdominal cavity and the associated forward pivoting motion that was observed for the rear passenger. In contrast the resultant lumbar forces are lowest for Model 'C' where the belt remained in place and pivoting was minimal.

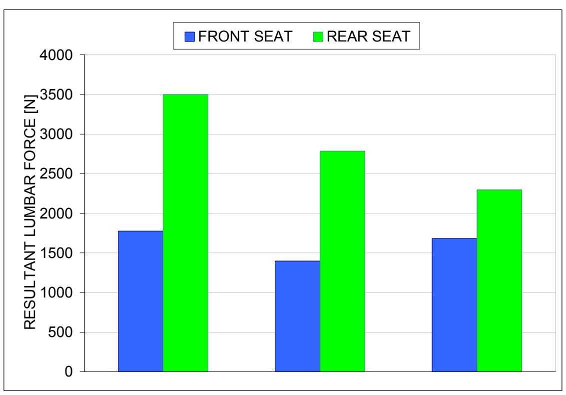

Figure 4: Resultant lumbar force for the front and rear seat occupant in Models A, B and C.

Hybrid III 6-year-old booster results: In all of the testing conducted to date, the Hybrid III 6-year-old restrained by the vehicle lap/torso belt in a booster seat, behaved in one of two ways. Either the torso belt would translate up towards the neck; or the torso belt would slide down towards the shoulder as the dummy was pitched forward.

As soon as the child dummy begins to load the shoulder belt, the webbing extends in the direction of the seatbelt anchor points. Upward motion of the belt into the neck, results in compression of the extreme upper quadrant of the chest, off-loading onto the neck and sternum deflections that are uncharacteristically low for the observed belt loads. Downward movement of the shoulder belt is associated with increased dummy excursion as the dummy slips out of the belt. At the moment of peak loading, the belt passes directly over the sternum or in very close proximity, producing high deflections. In some tests the dummy was found to slip out of the belt entirely, pivoting forward until the dummy head and thorax struck its lower extremities resulting in elevated head accelerations and chest deflections.

Figure 5 displays peak resultant head accelerations obtained in 48 km/h FFRB tests. There was one head strike into a seat frame resulting in a peak resultant acceleration of 324 g. A number of head contacts with the lower extremities resulted in accelerations that were as high as 225 g while accelerations arising from strikes with the upper extremities or chin to chest contacts were closer to 100 g. In the absence of head contact no head accelerations in excess of 80 g were observed. Accelerations above 80 g were not observed on rebound.

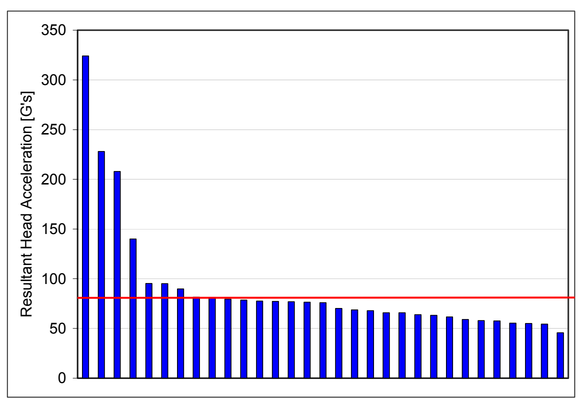

Figure 5: Peak head resultant accelerations for the HIII 6-yr-old in a booster seat 48 km/h FFRB.

Neck tension surpassed the Injury Assessment Reference Value (IARV) of 1490 N in all but the 40-km/h ODB tests. Peak upper neck tensions as high as 4500 N were observed and tended to be highest in those cases where the child dummy torso flexed forward and the head was projected to the feet.

Chest deflections ranged between 17 mm and 52 mm. Low thoracic deflections, below 25 mm, were associated with shoulder belts that translated up into the neck while the higher deflection values were observed in cases where the belt slipped off the shoulder.

Shoulder belt loads were high for all rear seating positions, irrespective of booster seat model and ranged from between 2000 N and 3000 N for low speed offset deformable barrier tests and between 4000 N to 6000 N for full frontal rigid barrier tests conducted at 40 to 56 km/h. Figure 6 displays the relationship between vehicle longitudinal acceleration at the CG and shoulder belt loads ($R^2 = 0.68$).

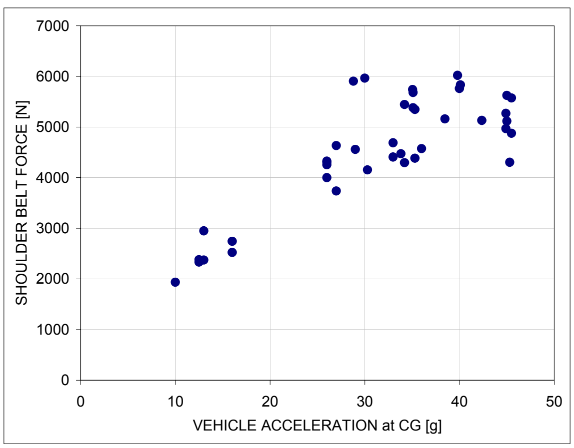

Figure 6: Correlation between rear seat shoulder belt force and longitudinal vehicle acceleration at the CG.

Six tests comparing dummy responses in outboard seating positions to responses in the centre seat position were carried out. The results of this comparison were inconclusive. Belt loads were equivalent while chest deflections were higher for three of the six centre positions and equal in one.

A preliminary comparison between second and third row seating suggests that dummy responses were equally elevated in the second and third row seats. In one case the third row centre seated child was launched upward towards the tailgate window during rebound. In another vehicle, the distance between the third and second row seat was less than between the first two rows and lead to a head strike with the second row seat frame.

Two alternative restraint configurations were explored in an attempt to reduce the dummy responses observed in lap/shoulder booster seats. The first was to place the 6-year-old dummy weighing 24 kg (52.5 lbs) into a CRS rated to 21 kg (47 lbs) or 21.3 kg (48 lbs) so that restraint now relied on the child seat 5-point harness instead of the vehicle seatbelt; and the second was to rigidly attach the convertible booster seat by way of the available LATCH and tether while still using the vehicle seatbelt to restrain the child dummy.

Figure 7 displays the normalized responses for one high speed ODB test at 60 km/h represented by the first set of bars, and 6 FFRB tests carried out at 48 km/h. The vehicle accelerations for these tests were of the order of 27 to 30 g, with the second and third sets of bars representing the results two different child seats in a single vehicle crash test. Chest deflections (Dx) were dramatically reduced though neck loads remained elevated. The two occurrences of elevated head acceleration were due to head strikes into the seatbacks of the front seats when the harness system failed. There were no failures of the LATCH or tether anchoring systems.

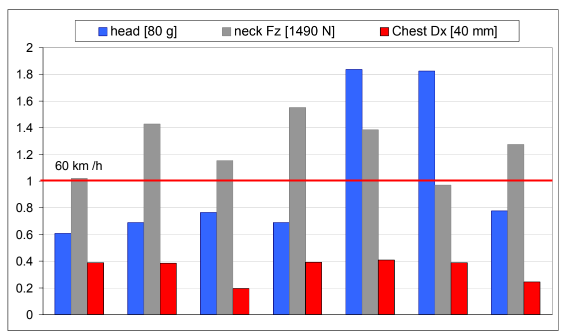

Figure 7: Normalized responses for the HIII 6-year-old in a forward facing 5-point harness CRS.

The effects of anchoring a booster seat to the vehicle seat by way of the LATCH and tether were compared to the conventional attachment method of a lap/shoulder belt and booster seat. Head accelerations, axial neck forces, chest deflections, seatbelt loads, lumbar forces and moment responses were examined as were the overall dummy kinematics.

There were no significant differences in head acceleration responses. The occasional 90 g to

100 g peak responses were typically due to upper extremity or chest to chin strikes and could not be definitely associated to seat attachment method.

Peak axial and shear neck loads were typically higher for the shoulder/lap belted booster seat. In the three cases where the latched booster seat produced higher responses the differences ranged from 5 to 15 % whereas in the remaining cases the increase in axial force for the shoulder/lap belted booster ranged from 5% to 68%.

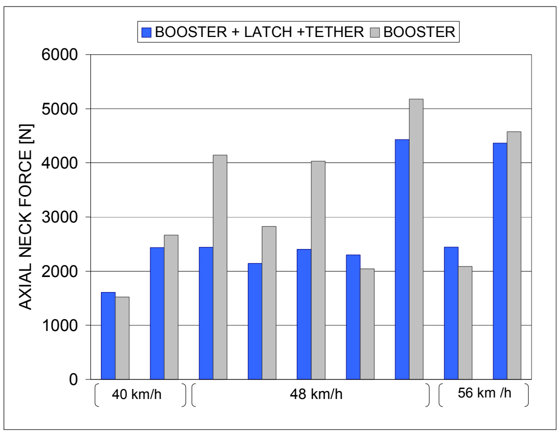

Figure 8: Comparison of axial neck force for the 6-yr-old in booster seat anchored with latch & tether and booster seat with belt only.

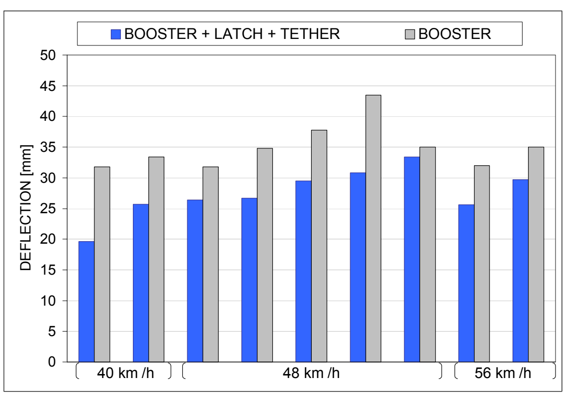

Figure 9: Comparison of peak chest deflection for the 6-yr-old in booster seat anchored with latch & tether and booster seat with belt only.

Chest deflections were consistently higher in the shoulder/lap belted booster. Figure 9 presents the comparative deflections for FFRB tests conducted at 40, 48 and 56 km/h. Consistent with the increase in deflection, shoulder belt loads were found to be higher for seven of the 9 shoulder/lap belt restrained boosters**.** Chest acceleration was unaffected by seat attachment method.

Peak resultant lumbar forces for the latched and shoulder/lab belted booster seat comparison are shown in Figure 10. The peak resultant lumbar forces found to be associated with greater abdominal penetration and rotation about the pelvis, in the small female, were more elevated in the shoulder/lap belt restrained booster**.** There were two tests wherein the lumbar forces were marginally higher (5 % to 8%) for the latched booster seat however, for the remaining tests, lumbar forces were 15 % to 90 % greater for the shoulder/lap belted boosters.

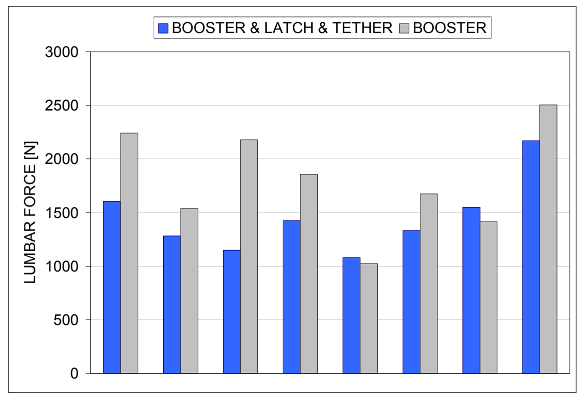

Figure 10: Comparison of peak lumbar forces for the 6-yr-old in booster seat anchored with latch & tether and booster seat with belt only.

The kinematics of the dummies in the two booster seat attachment configurations differed in their loading and rebound behaviours. During the loading phase, the shoulder/lap belted booster seats displayed greater forward excursion, rotation and vertical displacement than did the latched boosters. The dummy rebounded more rapidly and exhibited greater vertical displacement. Generally motion in the shoulder/lap belted booster seat was less controlled.

The armrests in one booster seat model sheared during impact in four tests with the LATCH attachment and once for the lap/shoulder belted both attachment configurations. There were no failures of the LATCH, or tether anchorages in the latched booster seat.

<u>Hybrid III 10-year-old with and without booster:</u> Two comparative tests were carried out to gain a better understanding of the Hybrid III 10-year-old responses and to compare responses for a 10-year-old dummy restrained in a booster seat and by the lap/shoulder belt alone. The comparison included two crash tests of identical model vehicles tested in a FFRB at 48 km/h. The longitudinal accelerations at the vehicle CG were

26 g for the non-booster seat test and 27 g for the booster seat test.

Chest deflection responses and associated shoulder belt force measurements are shown in Figure 12. The belt loads were equivalent for both test conditions, yet the deflections for the booster-seated dummy far exceeded the deflections recorded for the shoulder/lap belted dummy, both in magnitude and duration.

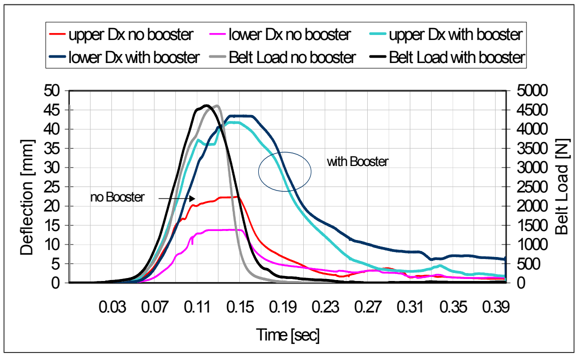

Figure 11: Comparison of upper and lower chest deflections for the Hybrid III 10-year-old with and without of a booster seat.

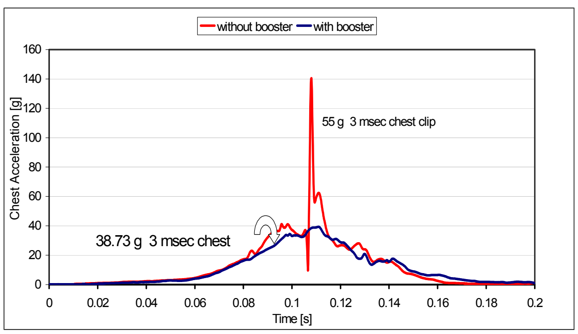

Figure 12: Comparison of resultant chest accelerations for the Hybrid III 10-year-old in and out of a booster seat.

The chest acceleration resultant shown in Figure 13 was primarily the result of a 121 g longitudinal component and a 71 g lateral component, which occurred 108 ms into the event. Video analysis failed to identify any external causes for this noise. Sternum and spine accelerometers failed to provide further information on the possible source of the noise.

The comparison of the Hybrid III 10-year-old dummy underlines the importance of including a variety of measurement parameters including video analysis in the evaluation of restraint performance. In the case of the shoulder/lap belted 10-year-old, the shoulder belt translated up into the neck placing the belt high on the chest while the lap belt penetrated the abdominal cavity. The resulting deflections which occurred in the uppermost quadrant of the chest were too remote to be detected by either of the two sternum IRTRACCs .

In the booster seat, the belt slipped off the shoulder of the 10 year-old dummy and was directly over the sternum at the moment of peak load, resulting in high upper and lower sternum deflections. There was significant excursion as the dummy rotated out of the shoulder belt. The lap belt remained on the pelvis throughout the event.

Limiting the analysis to the chest responses could lead to the false conclusion that the shoulder/lap restraint is better when in fact neither condition is desirable.

DISCUSSION

All booster seats effectively retained the lap portion of the seatbelt in the pelvic region and prevented the upward translation of the lap belt into the abdominal cavity. In contrast, the small female and 10-year-old dummies restrained with the shoulder/lap belt in the rear seat, all experienced abdominal penetration of the lap belt with one exception.

Lumbar force measurements in the small female were well correlated to lap belt migration in this small sample of six tests. Deflection should be evaluated in conjunction with the belt loads, particularly for rear seat occupancy where the dummy undergoes much less controlled displacements than the front occupant. This motion, which can in some cases be rather extreme, increases the potential for a redirection of the load application, away from the instrumented sternum. Analysis of shoulder and lap belt loads, in particular the proportion of lap to shoulder belt load should also be monitored. This can provide further insight into the relative distribution of forces between the thorax and pelvis, important in the detection of belt penetration or partial ejection of the thorax from the shoulder portion of the belt.

Booster seats were found to influence the pre-test belt placement but had insignificant effect on the kinematics of the upper body during the dynamic event. The motion and compressive response of the child dummy thorax was controlled almost exclusively by the vehicle seatbelt geometry. The belt loads generated in FFRB tests were simply too large and could not be redirected by way of plastic clips fastened to fabric or other non-structural seat components.

Other seat parameters, which may have influenced the booster seated dummy responses, include seatbelt webbing length, the relative seat pan and seatback angles, the seat stiffness and the upholstery. Such analysis was beyond the scope of this study but may be considered in future work.

The elevated chest responses for the booster seated child dummy are consistent with findings by Durbin et al. In a study investigating crashes of insured vehicles involving children the reduction of chest injury resulting from booster seat use was not statistically significant. The Transport Canada crash investigation teams will be intensifying their search for frontal crashes involving rear seated, restrained children for future crash reconstructions and dummy validation.

Restraining the dummy in a CRS rated to the appropriate weight limit may be a viable option for children between the ages of four and six. The chest is restrained by a 5-point harness, which distributes the loads well and effectively couples both the upper and lower torso of the dummy to the vehicle. Though the neck loads remained elevated, the level of injury risk that may be associated with these values is not known and further investigation, through accident reconstruction is needed to validate the biofidelity of the dummy neck. In a 2002 study Sherwood et al conducted sled testing with the Hybrid III 6-year-old dummy and compared the responses to a cadaver test. The authors concluded that the stiffness of the dummy spine contributed to high neck forces and moments that were not representative of the injury potential.

Larger children can benefit from the abdominal protection provided by booster seats. The results of this study, though still preliminary, suggest that protection, specifically of the chest, may be enhanced if the booster seat is anchored to the vehicle seat, as one would attach a CRS. Use of the LATCH and tether produce a more effective coupling than typically produced by the vehicle's lap/shoulder belt.

Child seat manufacturers are introducing more products designed for the upper weight limits and are exploring design options to improve booster seat performance. Testing of the rear seat continues to identify significant measurement parameters, test protocols and ultimately appropriate safety interventions.

CONCLUSION

Transport Canada began evaluating rear seat occupant protection in 2003 by introducing the Hybrid III 5th percentile female dummy and the Hybrid III 6 and 10-year-old child dummies in the rear seats of compliance and research test vehicles.

Balancing energy management and kinematic control of the small female dummy in a high-speed crash appeared to be problematic as either abdominal penetration occurred or very high chest responses developed.

The booster seat effectively prevented the lap belt from penetrating the abdominal cavity. However, restraint of the dummy and control of the kinematics was very strongly dependant on the vehicle seatbelt geometry and not the booster seat model type.

The evaluation of booster seat performance should be conducted during dynamic crash testing. Multiple test parameters such as head accelerations, neck forces, chest deflections and lumbar forces must be considered to obtain an accurate interpretation of the potential for injury.

ACKNOWLEDGEMENT

The authors would like to acknowledge Denton ATD for the loan of a much needed twin six year-old Hybrid III dummy, Pierre Beauchamp, Jean Marc Fiore, Alain Bussières and the staff of PMG Technologies for their assistance in the conduct of this study.

The opinions expressed and conclusions reached are solely the responsibility of the authors and do not necessarily represent the official policy of Transport Canada.

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2.Sherwood, C.G., Shaw, L., van Rooij, R.W. Kent, P.K. Gupta, J.R. Crandall, Orzechowski, K.M., Eichelberger, M.R., Kalliaris, D. Prediction of cervical spine injury risk for the 6-year-old child in frontal crashes. 46th Annual Proceedings Association for the Advancement of Automotive Medicine September 30, October 2 2002.

SAFETY FOR THE GROWING CHILD
– EXPERIENCES FROM SWEDISH ACCIDENT DATA

Lotta Jakobsson
Irene Isaksson-Hellman
Björn Lundell
Volvo Car Corporation
Sweden

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ABSTRACT

During the past 40 years, different child restraint systems (CRS) have been developed to improve protection for children of different sizes and ages. Development of more effective CRS, and a higher frequency in use of the restraints, in addition to enhanced vehicle safety, has resulted in an increased level of child safety.

This study examines accident data with Volvo cars in Sweden to evaluate child safety with respect to age, size and impact situation (including impact severity in frontal impacts); identifying optimal restraints as well as potential areas needing more attention. A total of 3670 children, aged 0-15 years, involved in car crashes 1987-2004 were selected from Volvo's statistical accident database.

The injury-reducing effect of the child restraint systems was high overall. The highest injury-reducing effect was found in rearward-facing child restraints for children up to 3-4 years of age, offering an injury-reducing effect of 90% compared to an unrestrained child. Belt-positioning boosters from 4 to 10 years of age were found to have an injury reducing effect of 77%.

Compared to adults, this study indicates that children have a generally lower AIS 2+ injury rate, except for abdominal and lower-extremity injuries. Abdominal injuries are mainly found in children using only a seat belt, emphasizing the need for belt-positioning boosters.

A tendency of higher injury risk was found when the growing child switches from one restraint to another, i.e. when the child is at the youngest age approved for the restraint. Thus, the total injury-reducing effect would increase if all children were to use the child restraint system most appropriate for their size and age. The challenge is to spread information as well as enhance design to encourage everyone to use the appropriate child restraint system and to use it correctly.

INTRODUCTION

The development of child restraint systems (CRS) for cars started in the early 60s. During the past 40 years, different child restraint systems have been developed to improve protection for children of different sizes and ages. Isaksson-Hellman et al. (1997) showed a clear trend of steadily increased safety for children in cars during these years in Sweden. This was due to the increased frequency in use of restraints, and the development of effective CRS. The rearward-facing CRS was shown to be especially effective. The percent of restrained children in Volvo cars in Sweden 1977-2004 is shown in Figure 1.

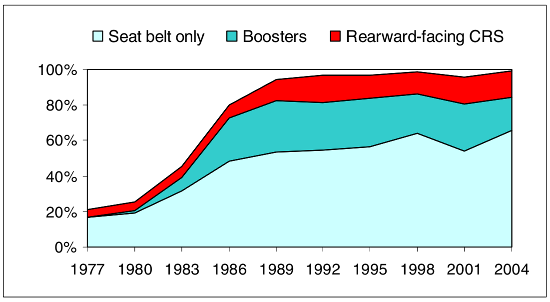

Figure 1. Percent of restrained children in Volvo cars in Sweden 1977-2004.

The different groups of restraint systems covered in this study are rearward-facing CRS (RF CRS), forward-facing belt-positioning, booster seats and cushions (boosters), and adult seat-belt only, Figure 2. Please note that forward-facing CRS for ages 1-4 with integrated child harness are very rare in Sweden, and therefore not included in this study.

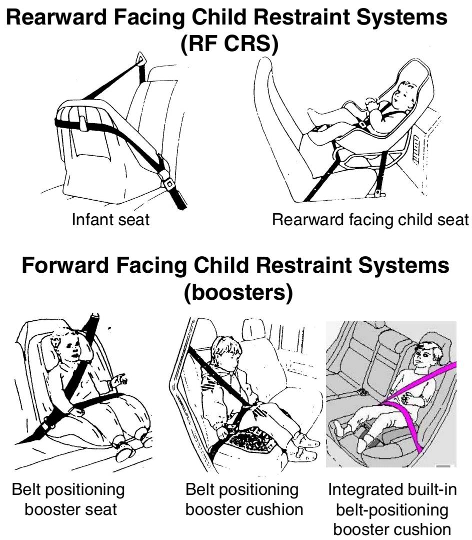

Figure 2. Analyzed child restraint systems

Rearward-Facing Child Restraint Systems

In 1964 professor Bertil Aldman introduced a rearward-facing child seat (Aldman, 1964). The purpose of this seat was to enhance support to the spine and head in the event of a frontal impact, i.e. to distribute the forces over a large part of the body. Small children have a different anatomy compared to adults; especially the proportion of the head's mass and height compared to the total body mass and height (Figure 3), and also the strength and development of the neck and cervical vertebrae (Burdi et al. 1968).

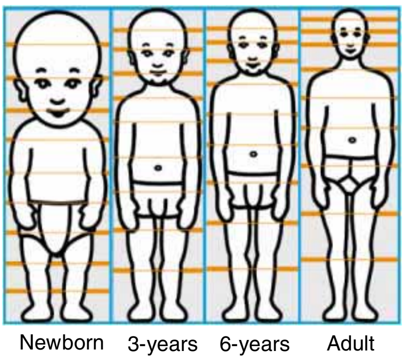

Figure 3. Body proportions for the growing child (source: Burdi et al. 1968)

The combination of high head mass and a weaker and more fragile neck in small children makes rearward-facing CRS the ultimate restraint system for this category of occupant. Several studies have pointed out the benefits of this restraint system, and it is recommended for use as long as possible; at least until 3-4 years of age (Tingvall 1987, Carlsson et al. 1991, Kamrén et al. 1993, Stalnaker 1993, Tarrière 1995, Isaksson-Hellman et al. 1997).

The two main groups of rearward-facing CRS are the infant seat and the rearward-facing child seat, Figure 2. In all rearward-facing CRS, the child is restrained by a harness, comprising a 3-, 4- or 5-point belt system. The infant seat is used from newborn to approximately 9 months old and is attached to the car by the adult seat-belt. The rearward-facing child seat, which is found mainly in the Scandinavian countries, can be used up to the age of 3-4 years. It is usually attached to the car by the adult seat-belt and an additional strap between the forward part of the CRS and the car structure below. In recent years, an international standard for attaching child restraints to cars has been introduced. It is called ISOFIX and in the USA also LATCH (Turbell et al. 1993, Langwieder et al. 2004).

Belt-Positioning Booster Seats and Cushions

When the child has reached approximately 3-4 years of age, it can be turned forward-facing in the car. The mass of the head is proportionally less and the neck is stronger. There are, however, still major differences as compared to adults. The iliac spines of the pelvis, which are important for good lap belt positioning and for reducing risk of belt load into the abdomen, are not well developed until about 10 years of age (Burdi et al. 1968). The development of iliac spines, together with the fact that the upper part of the pelvis of the sitting child is lower than of an adult, are realities that must be taken into consideration in the design, in order to give a child the same amount of protection as an adult.

Belt-positioning booster cushions were introduced in the late 70s (Norin et al. 1979). In Sweden there are three main forward-facing systems: booster cushions, booster seats and integrated booster cushions, Figure 2. The systems are used with the adult seat belt restraining the occupant together with the booster seat or cushion. The integrated (built-in) cushions were developed in order to simplify usage and to minimize misuse (Lundell et al. 1991). They can be found in the rear seats of Volvo cars from 1990, in the mid-seat or outboard position (depending on car model) and always together with 3-point seat-belts. The forward-facing CRS often used in USA, where the child is restrained by a harness or by a shield in the CRS, are very rare in Sweden and are therefore not included in the present study.

The booster allows the geometry of the adult seat belt to function in a better way with respect to the child occupant. The booster raises the child, so that the lap part of the adult seat belt can be positioned over the thighs, which reduces the risk of the abdomen interacting with the belt. An important feature regarding booster cushions is the belt-positioning device; keeping the belt in position during a crash. The booster also gives the child a more upright position, so he/she will not scoot forward in the seat to sit comfortably with their legs. This is a more safe position since slouching may result in very bad belt geometry (DeSantis Klinich et al. 1994). Other advantages of belt-positioning boosters are that the child, by sitting higher, will have the shoulder part of the seat-belt more comfortably positioned over the shoulder and will also have a better view.

Adult Seat Belt Only

When a child has grown to a height of approximately 140cm and the pelvis is also fully developed, the adult seat belt can be used without a booster. The conventional three-point belt is the best seat belt system. Volvo's studies have shown that three-point belts have a 15% better injury-reducing effect (AIS 2+ injuries) as compared to lap-belt only (Lundell et al. 1991).

Misuse

Several different definitions of misuse exist. Common types of misuse include incorrect or no mounting of the CRS, or the child not properly restrained in the CRS. Several studies have discussed these issues and can give an idea of its proportions (Tingvall 1987, Petrucelli 1986, Kamrén et al. 1993, Hummel et al. 1997). In the present study, this aspect of misuse is not possible to evaluate, since the cases are not possible to separate in the analyzed material.

Another type of misuse is a child not using the restraint designed for its size and age. The study of Isaksson-Hellman et al. (1997) showed that the maximum effect of a restraint system is not attained if the child is not using the optimal CRS for its age. Also, a tendency of higher injury risk was identified when the growing child switches from one restraint to another, i.e. when the child is at the youngest age recommended for the restraint. The present study, using the same data source complemented with more recent cases, focuses the safety of the growing child, with respect to age, stature and weight.

METHOD

A dataset of children in Volvo's statistical accident database is analyzed. Crashes involving Volvo cars in Sweden in which the repair costs exceed a specified level (currently SEK 45 000) are identified by the insurance company Volvia (If P&C Insurance). Photos and technical details of the cars (e.g. damage) are sent to Volvo's traffic accident research team. The owner of the car completes a questionnaire (shortly after the crash) to provide detailed information about the crash and the occupants. Injury data is gathered from medical records and analyzed by a physician within Volvo's traffic accident research team. Injuries are coded according to the Abbreviated Injury Scale (AIS, AAAM 1985). This forms the basis of Volvo's statistical accident database.

Occupants below 16 years of age involved in crashes occurring from 1987 to 2004 are selected for this study; a total of 3670 occupants, 47% girls and 53% boys. In Figure 4 the distribution of age, stature and weight of the children are shown. Infants are included in the 1 year old group.

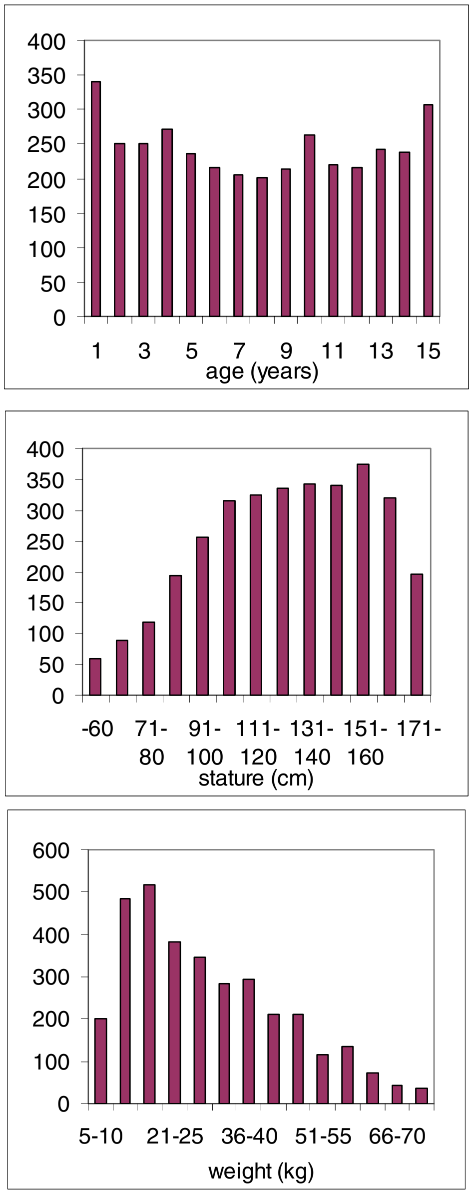

Figure 4. Distribution of age, stature and weight

The variations with respect to stature and weight of the child occupants are shown in Figure 5.

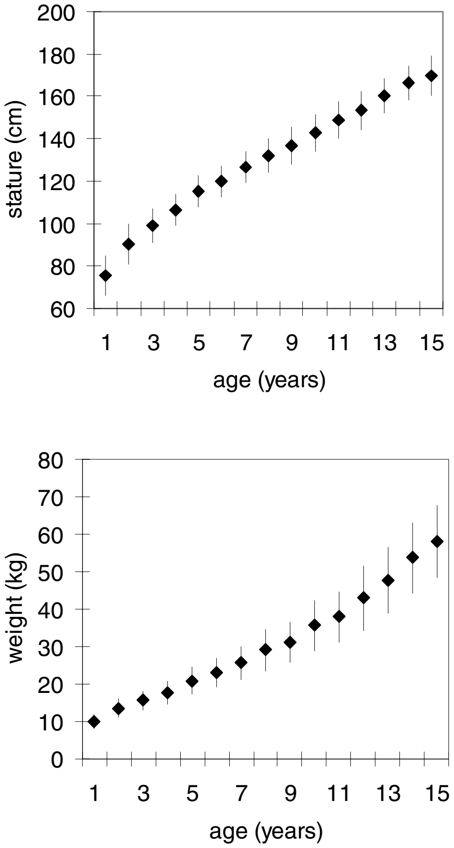

Figure 5. Variation in stature and weight, respectively, versus age; mean values and standard deviation.

The injury rate is calculated as the number of injured of a certain level of AIS divided by the total number of occupants in the group considered. Rearward-facing CRS are infant seats and rearward-facing child seats (in Sweden recommended up to age 3-4), Figure 2. The forward-facing booster includes belt-positioning booster cushions (including integrated built-in cushions) and booster seats. In these, the child together with the booster is restrained by the adult seat belt, Figure 2. Unfortunately, information regarding incorrect or no mounting of the child restraint system, or the child not properly restrained in the system is not available in the material. The number of children traveling in the different restraint systems and seating positions are shown in Table 1. The distribution of crash types is shown in Table 2. The distribution of child restraint systems versus age is seen in Figure 6.

For comparison, a subset of adult passengers is extracted from the database. A total of 3422 restrained front and rear seat passengers aged 20 to 40, involved in crashes occurring 1987 to 2004, is selected.

Table 1.
Number of child occupants with respect to seating position and restraint usage; seat belt only, rearward-facing CRS (RF CRS), forward-facing, belt-positioning booster seat (booster), belt-positioning booster cushion (cushion), integrated built-in booster cushion (int. cushion).

Restraint type	Front seat	Left rear seat	Mid rear seat	Right rear seat	**Total**
unknown	20	25	18	29	*92*
seat belt	571	535	241	634	*1981*
unbelted	16	58	41	53	*168*
RF CRS	353	21	22	58	*454*
booster	37	71	14	100	*222*
cushion	104	288	37	294	*723*
int. cushion	0	2	23	5	*30*
Total	*1101*	*1000*	*396*	*1173*	*3670*

Table 2.
Distribution of crash types.

Crash type	Number of child occupants	Distribution of crash types
Frontal impacts	1421	39%
Side impacts	869	24%
Rear end impacts	362	10%
Multiple impacts	297	8%
Rollovers and turnovers	184	5%
Multiple events	199	5%
Large animals	166	5%
Run-off road	78	2%
Side swipes	70	2%
Other	24	1%
	3670	

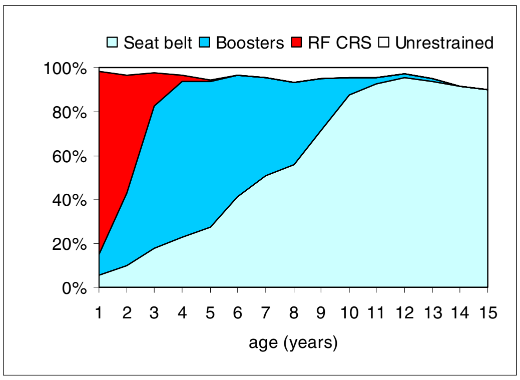

Figure 6. Distribution of restraint systems versus age.

RESULTS

Differences: Adult vs. Child Passengers

When comparing child and adult passengers (drivers are excluded), the injury rates are generally lower for restrained children as compared to restrained adults (aged 20-40), except for abdomen and lower extremities (Figure 7). The figure shows the distribution of injuries for all impact situations. Considering frontal impacts only, the same trend is seen.

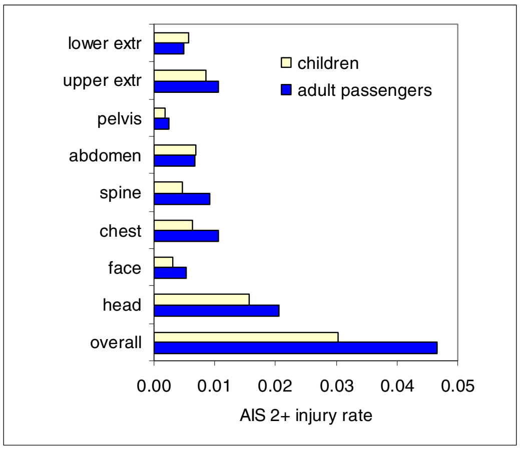

Figure 7. AIS 2+ injury rates (overall and per body part) for restrained adults in passenger seats (age 20-40y, N=3422) and restrained children (age 0-15y, N=3375), all impact situations, accident years 1987-2004.

Restraint System Effectiveness

The overall AIS 2+ (MAIS 2+) injury rates for children using/not using restraints of different types are shown in Figure 8.

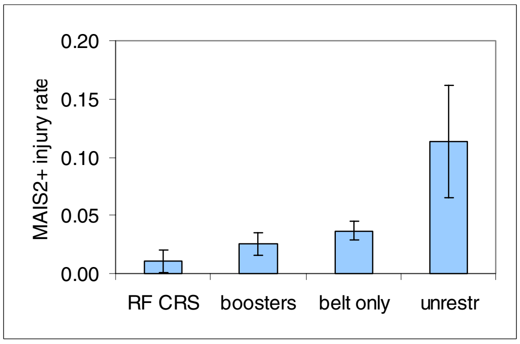

Figure 8. MAIS 2+ injury rates per restraint system (incl. 95% confidence intervals).

Figure 8 shows the very high level of protection for children in rearward-facing CRS (RF CRS). When restrained in belt-positioning booster seats or cushions (boosters), less than 3% were injured at level MAIS 2 or greater. Calculating the overall effectiveness of restrained compared to unrestrained children, the injury-reducing effect is used (Isaksson-Hellman et al, 1997). The overall injury-reducing effect MAIS 2+ for belted only is 68% with the confidence limits (C_L, C_U) = (48%, 80%), for boosters 77% with (C_L, C_U) = (60%, 87%), and for RF CRS as high as 90% with (C_L, C_U) = (74%, 96%) as compared to unrestrained children.

In Figure 7 all restrained children are included. Several of these children are not using the recommended child restraint system for their age and size. Figure 9 shows the MAIS 2+ injury rates at the age groups where the switch between the different restraint systems occur. Even though there is no statistically significant difference in injury rates, the effectiveness of the different restraint types is clearly demonstrated within the different age groups.

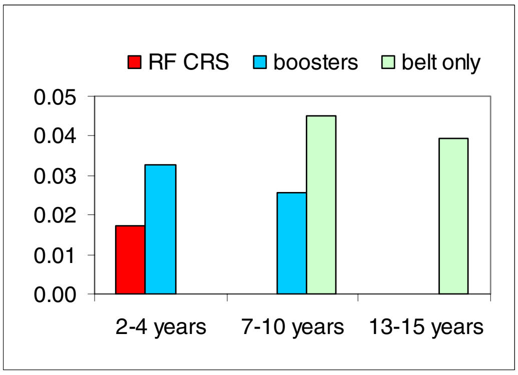

Figure 9. MAIS 2+ injury rates for children of specific age groups in different restraints.

As can be seen in Figure 9, there is a noticeable increase in MAIS 2+ injury rate if the growing child switches from rearward-facing to a forward-facing booster at around 3 years of age. The injuries to the 2-4 year-olds in boosters are mainly head injuries. Two children in frontal impacts sustained spine fractures; one of them a combination of fatal head and neck injuries. The injury rate in a booster decreases somewhat when the child grows older. At the switch to the adult belt only, between age 7 and 10, there is a remarkable increase in injury rate. The injuries for these children are spread over the whole body, with a distinct difference in abdomen injuries, which are only seen for the belted-only children. More than half of the MAIS 2+ injured belted-only children aged 7-10 had AIS 2+ abdominal injuries.

Injuries to Restrained Children

Among the 3375 restrained children (with known injury degree) there are 680 children with MAIS 1 injuries and 102 with overall AIS (MAIS) 2+ injuries. Five of the 102 injured occupants were restrained in a rearward-facing child seat. Three of them were injured in a frontal impact and two in multiple sequence accidents. The five rearward-facing children received AIS 2+ injuries to the head, chest, or lower and upper extremities.

A total of 128 AIS 2+ injuries are found for 72 children restrained by seat belt only, and a total of 38 AIS 2+ injuries to 25 children in boosters are found. Several children had injuries to multiple body areas. The AIS 2+ injuries to the restrained forward-facing children can be seen in Figure 10, divided by body part and impact situation. Head injuries are the most frequent AIS 2+ injuries, for frontal, side as well as other impact situations. Head injuries in frontal and side impacts will be explored further in this study. The head is by far the most injured body region in side impacts, while in frontal impacts the injuries are more evenly distributed over the different body parts. In the present study, injuries to the torso area, abdomen and lower extremities in frontal impacts will be studied further, as well. Upper extremity injuries are also among the most frequent AIS 2+ injuries. Six of the 20 AIS 2+ upper extremity injuries are injuries to the clavicle. They will be included in the section on injuries to the torso area. The remaining AIS 2+ upper-extremity injuries are mainly fractures to the arm bones. The mechanisms of these injuries are probably of the same type of mechanisms as for adults.

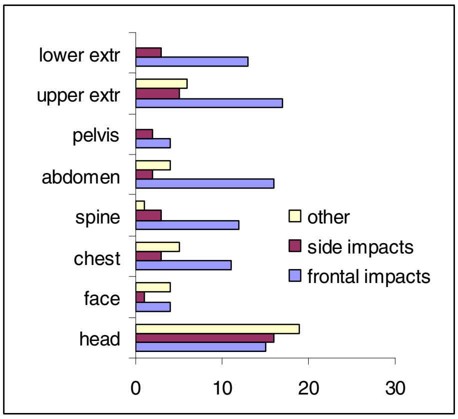

Figure 10. Number of AIS 2+ injuries to children in seat belt only (72 children) and boosters (25 children) shown by body part and impact type.

The growing child is an important aspect when designing child restraint systems. Several combinations of impact situation and body area will be discussed further in this paper with respect to occupant size and age, and when possible, with respect to impact severity. The distribution of Equivalent Barrier Speed (EBS, Mackay and Ashton 1973) versus degree of injury in frontal impacts can be seen in Figure 11. Frontal impacts account for 39% of all cases in this material and 50% of all the MAIS 2+ injured occupants. Figure 11 shows that impact severity is an important factor with respect to injury outcome in frontal impacts.

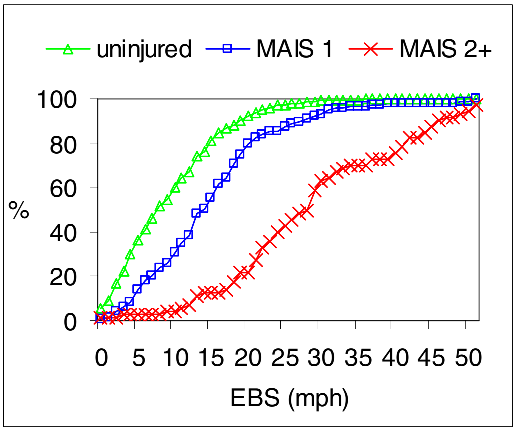

Figure 11. Cumulative distribution of EBS versus uninjured, MAIS 1 and MAIS 2+ injured occupants in frontal impacts.

Rearward-Facing CRS

The children traveling in rearward-facing CRS in a frontal impact are plotted in Figure 12, with respect to EBS and age, weight and stature.

As can be seen in Figure 12, the majority of all children in rearward-facing CRS are uninjured, even at high EBS. The children with MAIS 2+ injuries are mainly found at high EBS, while MAIS 1 injured children are found at any EBS. The severely injured one year-old child at EBS 26mph, was sitting facing rearward in the front passenger seat and sustained severe (MAIS 4) head injuries due to local intrusion. The one year-old child with MAIS 2, also sitting in the front passenger seat, sustained a lower extremity injury and minor head concussion. A third MAIS 2+ injured child, who was in a very high severity impact, sustained severe injuries (AIS 4) to the head and lungs as well as fractures (AIS 2) to the legs and one arm. The car he was traveling in collided with a large truck. The driver of the car sustained fatal injuries.

As demonstrated by Figure 12, the rearward-facing seat offers good protection for the small child in frontal impacts. In this dataset, frontal impacts account for three of five rearward-facing children with MAIS 2+ injuries. The other two were injured in multiple sequence crashes with somewhat uncommon situations. In the data, there are no rearward-facing children with injuries more than AIS 1 in side or rear-end impacts.

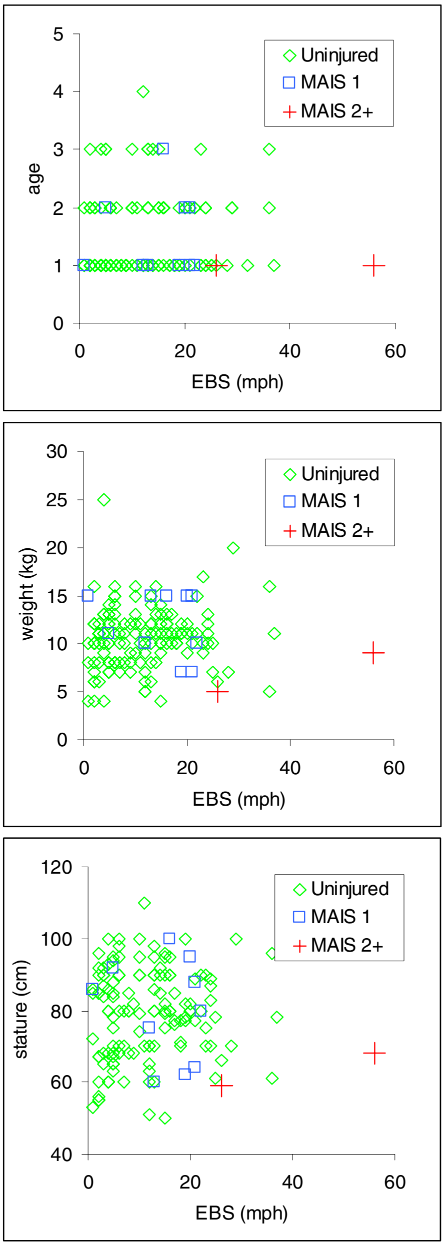

Figure 12. Distribution of injured (overall injury) and non-injured, rearward-facing children in frontal impacts, EBS vs. age, weight and stature. One injured (MAIS 4) two year-old child with unknown weight and stature is beyond the EBS scale (very high EBS).

Head Injuries in Side Impacts

In side impacts, the most common body area injured is the head (Figure 10). Head (including face) injury distribution for age versus stature is shown in Figures 13a, b, for all occupants in side impacts and near-side occupants only, respectively. The children are all restrained, belted-only or using boosters. Near-side occupants are those sitting on the struck side of the car during the crash.

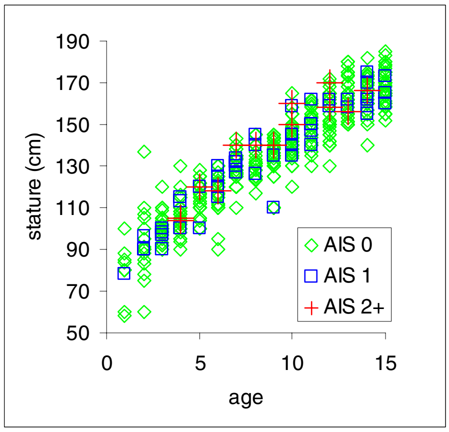

Figure 13a. Distribution of <u>head injury</u> AIS for forward-facing children (boosters and belted-only) in side impacts, stature vs. age.

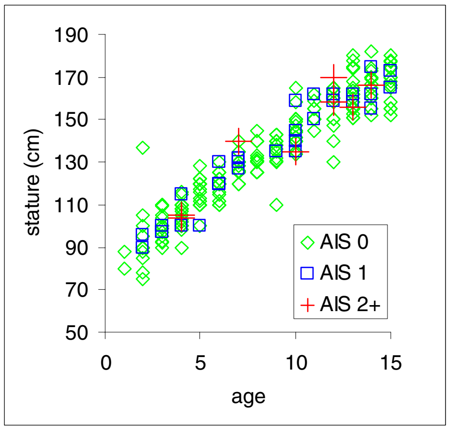

Figure 13b. Distribution of <u>head injury</u> AIS for forward-facing children (boosters and belted-only) in <u>near-side</u> side impacts, stature vs. age.

No relation between increased age/stature and injury rate is found for children in side impacts (Figures 13a, b). This is seen also when separating belted-only occupants and those using boosters, as well as near-side and far-side occupants.

The overall AIS 2+ injury rate for head (including face) injuries in side impacts is higher for children sitting on the near side; 3.1% as compared to 1.8% for those on the far side. The most frequent AIS 2+ injuries are brain concussions and skull fractures, rather evenly distributed between the children on the near side and far side. The most usual impact location is the side structure. Some of the far-side children have struck the back of the front seats. As for adults, head (including face) injuries are in most cases sustained by the occupant impacting hard structure.

Head and Face Injuries in Frontal Impacts

In Figure 14, head and face AIS is plotted for EBS vs. age and stature for forward-facing children in frontal impacts. As can be seen, EBS has the largest influence on AIS 2+ injuries. The two-year old (using lap/shoulder belt and booster) with head AIS 6 sustained a combination of fatal head injury and cervical spine fracture at EBS 50mph.

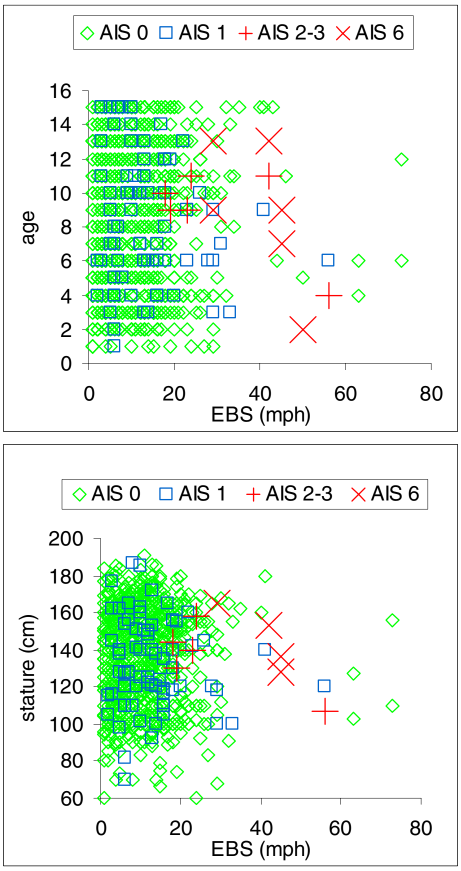

Figure 14. Distribution of head and face injury AIS for forward-facing children (boosters and belted-only) in frontal impacts, EBS vs. age and stature.

Among the total of 30 AIS 2+ head and face injuries for the 12 restrained forward-facing children in frontal impacts, the most common injuries are fractures (30%) (equally distributed between skull base, nose/maxilla and forehead) edema (26%) and concussion (20%). The most common AIS 1 injuries to the head and face are abrasions (23%), cuts (19%), contusions (17%) and pain (10%). The injury distribution for children is similar to that for adults. When studying the combinations of head injuries for the individuals, the mechanisms for AIS 2+ head injuries seem to be impact-related. The exception for this is the typical combination of fatal head and neck injury for the smallest forward facing children, as exemplified by the 2-year old at EBS 50mph, which occurred without head impact.

Abdominal Injuries in Frontal Impacts

The distribution of abdominal injuries can be seen in Figures 15 a, b, for children in frontal impacts, belted-only and in boosters, respectively. Abdominal injuries of AIS 2+ are found at higher EBS.

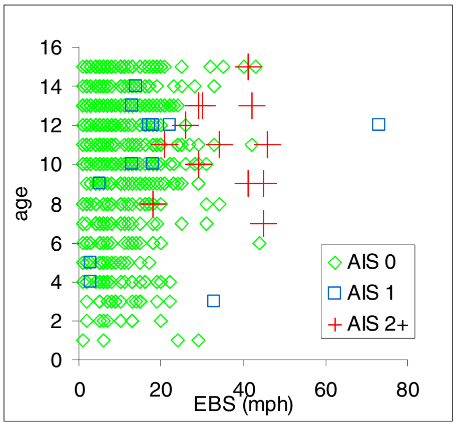

Figure 15a. Distribution of abdominal injury AIS for children restrained by belt only in frontal impacts, EBS vs. age

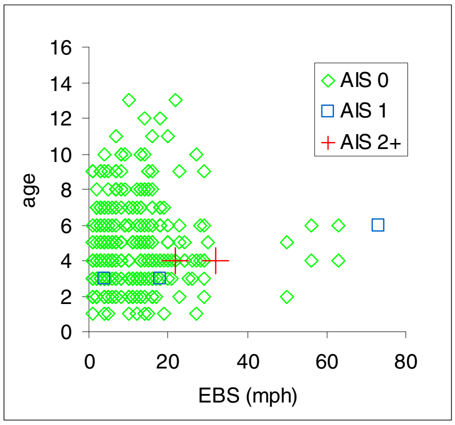

Figure 15b. Distribution of abdominal injury AIS for children in boosters in frontal impacts, EBS vs. age

The abdominal AIS 2+ injury rate is less for children restrained in boosters as compared to belt-only restrained; 0.8% as compared to 1.7%. The positive trend of reduction of AIS 2+ abdominal injuries if using a belt-positioning booster seat or cushion, as shown in Figures 15 a, b, confirms earlier studies (Isaksson-Hellman et al. 1997, Hummel et al. 1997). One of the two injured 4 year-old children using boosters was involved in a severe impact with a large truck in which only a younger sister in a rearward-facing child seat survived the crash. Both of the four year-olds were seated on booster seats with very poor guidance of the lap belt. During the crash, the belt slid up into the abdomen and the loads were transferred into the soft tissues; resulting in fatal abdominal injuries for one of them, and internal abdominal injuries, AIS 2, for the other.

Torso Injuries in Frontal Impacts

Injuries to the torso (chest, clavicle, shoulder and throat) are shown in Figure 16 with respect to age, weight and stature versus EBS, for all forward-facing restrained children (belted-only and boosters).

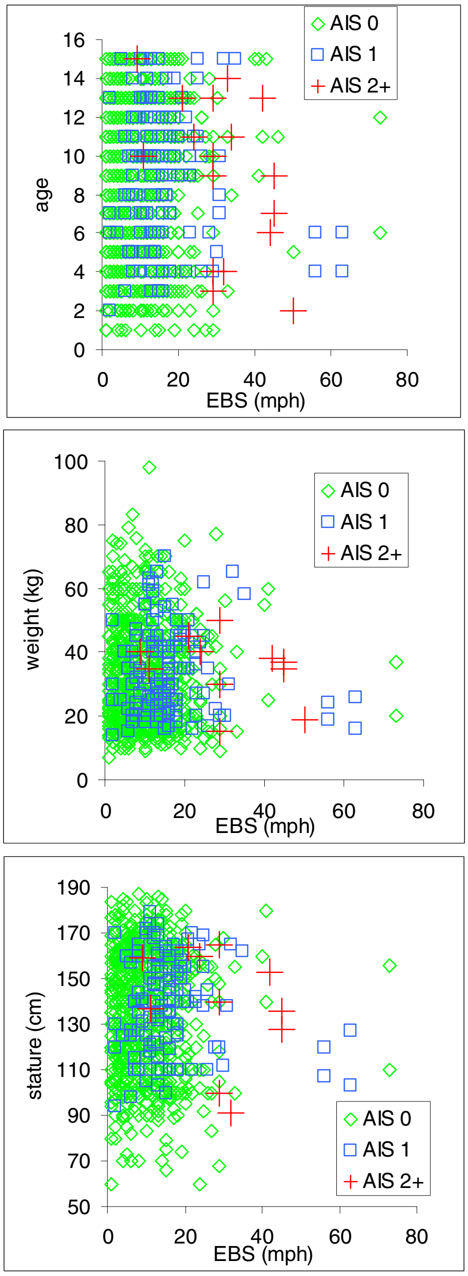

Figure 16. Distribution of torso injury AIS for forward-facing children (boosters and belted-only) in frontal impacts, EBS vs. age, weight and stature.

The injuries to the torso area are distributed evenly with respect to occupant size and age (Figure 16), and also between those wearing seat belt only and those in boosters. As can be seen in Figure 16, there is a general trend that AIS 2+ injuries are related to increased impact severity.

The most frequent AIS 2+ injuries to the torso are fractures (43%) to ribs, clavicle and sternum, together with bleeding, ruptures and contusions to inner organs (17% on each). The most common AIS 1 injuries are pain (56%), contusions (21%) and abrasions (20%). The types of injuries indicate that the most common injury mechanism for most torso injuries is probably belt interaction. Similar injury trends and injury type distribution are seen as for adults, indicating that the injury characteristics, and thus the mechanisms for these injuries, are probably not unique for children.

<u>Lower extremity injuries in frontal impacts</u>

The AIS 2+ injury rate of lower extremity injuries to children is as high as for adults, see Figure 7. Lower extremity injuries to forward-facing children are mainly found in frontal impacts (Figure 10). In order to understand the mechanisms of child lower-extremity injuries, and to evaluate whether they are different for children as compared to adults, the distribution of lower-extremity AIS is plotted for age versus EBS, for belted-only children (Figure 17a) and children in boosters (Figure 17b), respectively.

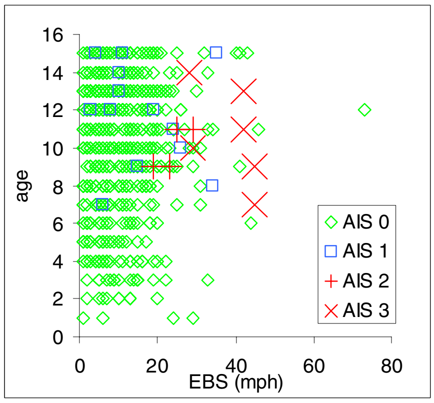

Figure 17a. Distribution of lower extremity injury AIS for <u>belted-only</u> children in frontal impacts, EBS vs. age

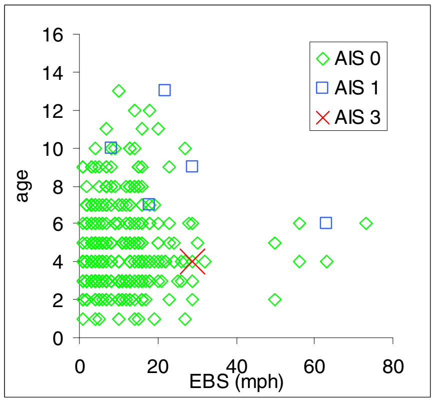

Figure 17b. Distribution of lower extremity injury AIS for children in <u>boosters</u> in frontal impacts, EBS vs. age

The AIS 3 injuries in Figures 17a,b, are femur fractures. They occur typically at higher impact severity than the AIS 2 injuries, which are fractures below the knee. As can be seen in Figure 17b, there is only one small child (with lap/shoulder belt and booster) who sustained lower extremity AIS 2+. All the other AIS 2+ injured children were 7 years or older, and all of them were 130 cm or taller and restrained by seat belt only. For these children, the injury mechanisms would be similar as for adults; the knees interact with structure in front of them and are broken when high loading is transferred

DISCUSSIONS

Over the last three decades, total protection for children has increased through a combination of increased usage (Figure 1) and the performance of the child restraint systems (Isaksson-Hellman et al.1997). The data in this study is from Volvo cars in Sweden. In Sweden, the way of transporting children in cars differs to some extent from other countries. Also, the overall use of restraints for children is as high as 95%, which might not be representative for several other countries and car brands. The aim of this study was to show the great benefits of the existing child safety systems, and to use the detailed data to suggest potential areas for further improvement.

The protection of the growing child in the car is a question of designing child-restraint systems specifically for the needs of the child. Age as well as stature and weight are important aspects with regard to the specific needs. Earlier studies have found that children are best protected if they travel rearward-facing up to the age/size when the mass of the head is proportionally less and the neck is stronger; at least to age 3-4. After this, the restraints need to compensate for the development and the size of the pelvis to accommodate a good belt geometry; at least up to age 10, preferably older. This study emphasizes the good performance of the safety systems evaluated. The switch of restraint is also highlighted. An increase in injury rate indicates that children turn forward-facing too early and do not stay in belt-positioning seats long enough. This also suggests that adaptable booster seats are desirable; that is, seats that can be adjusted to the size of the growing child.

In this paper, the good performance of rearward-facing CRS is demonstrated. The performance of rearward-facing seats is shown by the low injury rate. Only three children sustained MAIS 2+ injuries in frontal impacts and they were all exposed to relatively high severity impacts. In contrast to this, the two-year old forward-facing child (in a lap/shoulder belt and booster) sustaining the combination of fatal head injury and cervical spine fracture typically illustrates the vulnerability of the neck and head for small children in forward-facing boosters. This child's five-year old sister, sitting next to him in the rear seat (using the same type of restraints), sustained no injuries. This two-year old would have been better protected in a rearward-facing CRS. Other cases of this type of injury mechanism are described in Fuchs et al. (1989) and Stalnaker (1993). The rearward-facing child seats are designed primarily for frontal impacts, however the outcome for side and rear-end impacts indicates a good performance also in these situations. In this data, no rearward-facing child sustained MAIS 2+ injuries in side or rear-end impacts.

A large part of this study deals with injuries to restrained, forward-facing children, mainly aged 3 and over. In this data, the head is the most frequently injured body region. In frontal impacts, injuries to head/face as well as the torso area, abdomen and lower extremities are studied in detail, and will be discussed with respect to the possible mechanisms. For the youngest children in boosters, injuries to the cervical spine in a frontal impact are the highest priority, although they are not frequent in this data. Because of relatively few children below age 4 in boosters in this data, only one case is available to illustrate this mechanism. Among the injuries studied, abdominal injuries for belted only children and the combined head and neck injury for the smallest booster children in frontal impacts were found unique for children, and need special care. Injuries to the torso area, head and lower extremities seem to be of the same mechanisms as for adults, and need general care and focus on adaptivity in all safety system development.

For most head injuries to forward-facing children, in frontal impacts as well as in side impacts, the main injury mechanism is the head impacting into something. The exception is the fatal combination of skull base fracture and neck injury for small forward-facing children in frontal impacts, which does not require a head impact to occur. The head impact mechanisms, both in frontal and side impacts, are not unique for children. In side impacts, measures for adults will probably benefit children as well. In frontal impacts, measures to avoid head impacts are encouraged for children as well as for adults.

For forward-facing children in frontal impacts, the injury mechanisms of the injuries to the torso area (chest, shoulder, clavicle and throat) are probably interaction to the shoulder part of the lap/shoulder belt. There are no major differences in injury pattern as compared to adults. Injuries of AIS 2+ were mainly found at higher impact severity. Injuries to the lower extremities of forward-facing children were explored to evaluate if there were any unique mechanisms for children as compared to adults. All except one of the lower-extremity injured children were rather tall (>130 cm) and restrained by seat belt only. The lower extremity injuries that children were exposed to occurred at rather high impact severity, especially the femur fractures. The mechanisms of lower-extremity injuries for these children would be of the same kind as for adults; the knees interact with the structure in front of them and are broken when high loading is transferred.

The importance of a belt-positioning boosters for forward-facing children, in order to avoid abdominal injuries by the abdomen slipping under the belt, has been shown previously (DeSantis Klinich et al. 1994, Isaksson-Hellman et al. 1997, Warren Bidez and Syson 2001). The data presented in this study support these findings and emphasizes the importance of belt-positioning systems, and that the booster is designed to hold the belt firmly on the pelvis or thighs during a frontal impact. It is recommended for children up to the age of 10 to use a belt-positioning booster. However, Figures 15 a, b, suggest that even the 11-12 year-old child would probably benefit from such a device.

The injury reducing effect of the child restraint systems is high. However, the total injury-reducing effect would increase if all children used the child restraint system most appropriate for their size and age. Future challenges for improved protection are to spread information as well as enhance designs to encourage everyone to use the appropriate child restraint system and to use it correctly.

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CRASH PERFORMANCE EVALUATION OF BOOSTER SEATS FOR AN AUSTRALIAN CAR

Judith L. Charlton
Brian Fildes
Ron Laemmle
Sjaanie Koppel
Liam Fechner
Kristen Moore
Monash University Accident Research Centre
Australia
Stuart Smith
Fiona Douglas
Irena Doktor
Holden
Australia
Paper Number 05-0425

ABSTRACT

The aim of this study was to examine the relative effectiveness of two booster seats for use by children across a wide age span, from around 3 years to 10+ years. The study was part of a broad research program to identify suitable child restraint systems (CRS) to fit a large sedan vehicle to maximise protection in a crash. Dummies were restrained in the rear seat of a vehicle buck in three restraint configurations with boosters: (i) with a standard adult lap-sash seatbelt, (ii) with a seatbelt plus H-harness and crotch strap, or (iii) with a seatbelt plus H-harness with the crotch strap disengaged (to simulate the effects of real-world misuse conditions), and a fourth condition (iv) with an adult seatbelt only. Boosters were fitted in the vehicle with two different anchorage systems: a standard seatbelt and a system including a retrofitted rigid ISOFIX attachment and top tether. HyGe sled tests were conducted to simulate a 64 km/h offset deformable barrier frontal impact with a change in velocity of around 71 km/h. Preliminary investigations were also conducted using side impact simulations with a change in velocity of around 15 km/h. Overall, the booster seats for both the 6 year old dummy and the 3 year old dummy when a (with harness and crotch strap) provided superior crash protection than use of the adult seatbelt. For tests when the H-harness was used to restrain the dummy, use of the crotch strap was critical in eliminating 'sub-marining'. The research highlighted the potential for serious injury with misuse of child harness systems and identified several areas for design improvement of booster seats.

INTRODUCTION

Motor vehicle crashes are one of the leading causes of child death and acquired disability (NHTSA, 2002). Recent figures for the State of Victoria in Australia show that between 1998 and 2002, an average of 148 children aged 10 years and younger were killed or seriously injured each year in car crashes. This equates to around 900 child injuries or deaths per year in Australia. Of those killed or injured, 62 per cent were aged 5 to 10 years; 32 per cent were aged 1 to 4 years; and six per cent were under 12 months of age. This suggests that a significant effort is warranted to reduce child occupant injury, particularly in the 5-10 year old age group.

Child restraint systems (CRS) for vehicles are designed to provide specialised protection for child occupants of vehicles in the event of a crash. Recent estimates of CRS effectiveness have suggested that they may reduce injury by approximately 70 percent compared with unrestrained children (Mackay, 2001; Webber, 2000). Adult seatbelts, on the other hand, are not designed for children. Hence, it is not surprising that although children wearing adult seatbelts are better protected (53 percent less likely to be seriously injured) than children who are unrestrained, children in appropriate CRS or booster seats are 60 percent less likely to be seriously injured than children wearing adult seat belts (Durbin, 2001). In an in-depth study of Australian fatal crashes involving child occupants, Henderson (1994) reported a 26 percent reduction in MAIS2+ injuries for those restrained in a CRS over a seat belt.

The effectiveness of child restraint systems, however, is critically dependent on correct installation of the restraint in the vehicle, correct harnessing of the child in the restraint, and use of the appropriate restraint for the child's size and weight. Incorrect and inappropriate fitment and use of restraints may reduce or nullify safety benefits (Henderson, 1994; Paine & Vertsonis, 2001).

Australian legislation pertaining to child restraint use requires that children less than one year must be restrained in an approved, properly fitted and adjusted CRS. However, the law relating to use of child restraints by older children is less definitive and states that children over one year must be in either an appropriate child restraint or use a suitable seat belt (National Transport Commission, 2000). In the absence of more clear guidelines for CRS use, the responsibility largely rests upon parents to determine what type of restraint is 'appropriate' for their child. Notwithstanding this shortcoming in the legislation, usage rates of child restraints in Australia are relatively high. An observational study conducted in Australia in 1994 estimated that usage rates exceeded 95 percent (Henderson, Brown & Paine, 1994). However, the survey techniques used to obtain these estimates do not allow for accurate estimates of correct installation and appropriateness of restraint for the child's height and weight (Paine & Vertsonis, 2001). Hence, although compliance estimates are high, these figures belie reported error rates in CRS use, as discussed below.

While CRS manufacturers provide adequate instructions for fitment, it is generally acknowledged that installation and use of child seats and boosters is somewhat complicated and prone to error. Indeed, studies show that inappropriate use and misuse of the fitment of CRS is widespread (Glanvill, 2000; Paine, 1998; Paine & Vertsonis, 2001; Wren, Simpson, Chalmers, & Stephenson, 2001). In a recent survey of parental attitudes and behaviours in relation to child restraints, Glanvill (2000) reported a number of gaps in knowledge about correct use of child restraints, the risks associated with incorrect installation, and of children travelling in restraints that are inappropriate for their size.

It is important that as children grow, they use a restraint that is appropriate for their size (particularly, height and weight) (Winston, Durbin, Kallan & Moll, 2000). A number of researchers have reported that a relatively high proportion of children who grow out of a CRS suitable for young children move directly into an adult seat belt rather than using a booster seat (Winston et al., 2000; Ramsey, Simpson & Rivara, 2000). Durbin reported that booster usage rates in the United States varies across this age range, from 33 percent amongst 4 year olds to 10 percent for 8 year olds (2000). More recent U.S. figures show that booster usage amongst children in the weight range 18.6-22.7 kg (41-50lb) has increased from 5 percent in 1999 to 17 percent in 2002 (Winston, Chen, Arbogast, Elliott & Durbin, 2003). The authors note that these improvements in usage rates suggest the success of a number of community, corporate and government campaigns to promote appropriate restraint for children.

In Australia, there have been recent efforts to address appropriate CRS usage, particularly for boosters amongst older children, aged 4-10 years (Charlton, 2004). Anecdotal evidence suggests that few children at the upper end of this age range use boosters, although no recent data exist for usage rates for this specific age group. In the absence of more definitive legislation regarding appropriate CRS for older children, we have sought alternative solutions to promote the use of boosters for children up to 10 years. One possible approach is to offer a restraint system that takes children from toddler age to booster age in one child restraint system. This could be thought of as a *hybrid child seat/booster*, which would function as a forward-facing child seat with an H-harness for younger children and as a booster when used with a seatbelt only for older children. While a small number of restraints of this kind are currently available on the market in Australia, none have been subjected to vigorous crash testing and none have been developed with ISOFIX anchorage systems which are currently under consideration for the Australian Standards on CRS (AS1754). With these developments in mind, the current study aimed to examine the relative effectiveness of selected boosters, used as *hybrid child seat/boosters,* that would be suitable for children across a wide age span, from around 3 years, when a forward-facing CRS with harness would be suitable, to 10+ years, when conventional boosters with seatbelts, or adult seatbelts only may be appropriate. Of particular interest was the effectiveness of the boosters compared with a standard adult seatbelt. In addition, we considered the crash effectiveness of selected boosters when used with a harness with the crotch strap disengaged (to simulate the effects of real-world misuse conditions);

METHOD

Two child booster seats were tested. These are referred to in the paper as Booster A and Booster B and are designed for children in the weight range 14-26 kg and 15-36 kg respectively. Four tests were conducted: Three with booster seats with either a (i) standard adult lap-sash seatbelt, (ii) a seatbelt plus H-harness and crotch strap combination (H+C), or (iii) a seatbelt plus H-harness with the crotch strap disengaged from the lap part of the seatbelt (H-C) (to simulate real-world misuse conditions). (iv) A fourth test restrained the dummy in an adult lap-sash seatbelt only.
In the case of tests (ii) and (iii), where the booster was used as a forward-facing restraint (with harness) suitable for a toddler, both ISOFIX and top-tethers were retrofitted. This modification is in line with proposed changes to the AS1754 which apply to this type of restraint (but not required for boosters). The ISOFIX anchorage system comprised two connectors

that were attached in a rigid fashion to the base of the booster seat. The connectors were then attached to the vehicle at two prototype ISOFIX anchorage points, which were welded to the sedan buck and located at the junction between the vehicle seat cushion and seat back.

Booster A was selected on advice from the local manufacturer. Booster B was a European import, selected because of the wide side wings around the head and adjustable height of seat back; characteristics thought to be important in crash protection.

HyGe sled tests were conducted using a large sedan vehicle buck. The booster seats were fitted in the right or left side rear seating positions in a simulated offset deformable barrier frontal impact with an impact speed of 64 km/h. A limited number of side impact simulations (near-side) were also conducted representing an impact speed of around 15 km/h. New seatbelts and booster seats were used in each test and the rear seat belt anchor points were reinforced to withstand numerous tests. The front seats were positioned mid-way between full forward and the 95th percentile positions and the front seatback angle was 25º from vertical.

Kinematics from Hybrid III 6 year old and 3 year old dummies were used for frontal tests and from a TNO P6 6 year old dummy for side impact tests. A sub-set of these measures are reported in this paper. These are Peak Head Acceleration values and Neck Injury Criteria (N_{ij}), which were computed from the neck axial forces and flexion bending moments. High-speed digital video footage was captured from two on-board cameras for each test. The digital images were analysed using digitising software to estimate the maximum head displacement (mm). These measures were computed as the distance travelled by the centre of gravity of the dummy head from the commencement of the test to its point of maximum forward motion in the horizontal plane. In side impacts, maximum lateral head displacement was measured both for the initial (impact) phase as well as the rebound phase of the test. Video recordings were inspected by two independent observers for evidence of contact with the vehicle interior (and other contact points) and ‘sub-marining’. Sub-marining is an undesirable effect in which the dummy slides pelvis first, forward and under the harness/seatbelt. Since there were no discrepancies in observer judgements, a single measure of these data is presented.

Due to the limited biofidelity of the child dummies and the lack of biomechanical knowledge about injury mechanisms in infants and young children, dummy kinematics were compared across restraint systems rather than against specified criteria.

RESULTS

The results are presented in two sections: First, a comparison frontal test results of the restraint types suitable for older children using a 6 year old dummy. In this section we compare performance of the two boosters with each other and with an adult seatbelt only. Side impact tests for the two boosters are also compared. In the next section we compare the results of frontal tests for restraint types suitable for younger children using a 3 year old dummy. Comparisons were made between the two boosters with ISOFIX anchorages and top tethers and used with a full H-harness (with and without a crotch strap). In addition, the boosters are compared with an adult seatbelt only.

Comparison of restraint systems with a 6 year old dummy

Figure 1 shows the peak acceleration of the head for tests with the 6 year old dummy. Peak Head Acceleration was highest for the 6 year old dummy restrained in a seatbelt only (89g). Head acceleration values for Booster A and Booster B with the standard seatbelt were not notably different (62g and 60g, respectively). The head acceleration results suggest that both boosters provided a considerably superior level of protection to a seatbelt.

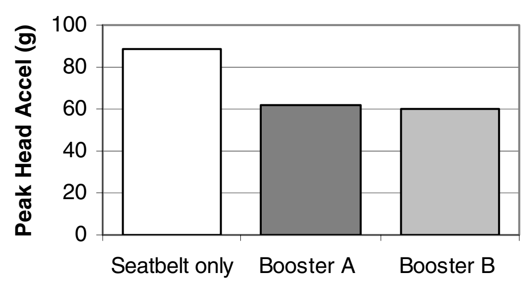

Figure 1. Peak Head Acceleration for tests with HIII 6 year old dummy

Neck injury (N_{ij}) values were calculated from the axial forces and flexion bending moments, providing a composite neck injury indicator. While no direct comparisons are made with the conventional injury threshold of 1.0 (FMVSS 208), the higher the N_{ij} value, the higher the potential for neck injury. As shown in Figure 2, the pattern of results for N_{ij} across restraint types mirrors the results for head acceleration. That is, the highest neck injury value was recorded with use of the adult seatbelt only. The two boosters used with the conventional seatbelt restraint did not differ notably (0.82 and 0.75 for Boosters A and B, respectively). Taken together with the Peak Head Acceleration measures, the results for neck injury suggest that use of the adult seatbelt only offers the weakest level of occupant protection for the 6 year old dummy.

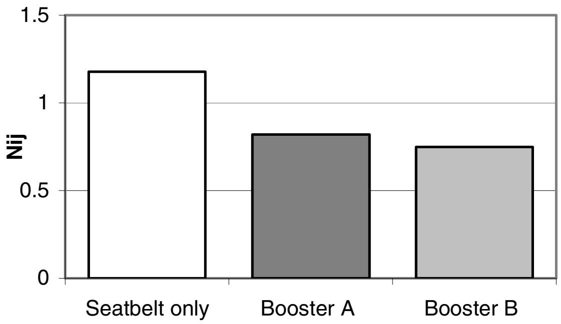

Figure 2. N_{ij} Tension Flexion for tests with HIII 6 year old dummy

Analysis of the maximum forward motion of the dummy head showed similar patterns for boosters with seatbelts (see Table 1). Interestingly, the restraint of the 6 year old dummy in a seatbelt only, provided better restraint of head motion than the two boosters which did not differ notably from each other. This is likely to be due to the added mass of the booster behind the dummy, contributing to its forward momentum. Despite the increased forward motion, inspection of the video recordings showed that the dummy head was well clear of the rear of the front seat in each of the three tests and, importantly, none of the restraint systems permitted the dummy's head to contact the vehicle interior or the dummy's knees. The seatbelt guides also maintained the sash and lap belts in a good position over the dummy's shoulder and pelvis throughout the tests.

Table 1. Summary measures for frontal tests with 6 year old dummy

Restraint Type	Max Head Excursion Impact Phase (mm)	Head Contact
Seatbelt only	530	No
Booster A Seatbelt	850	No
Booster B Seatbelt	800	No

Results for the side impact tests with the TNO P6 dummy are summarised in Table 2. Peak Head Acceleration values were the same across the two restraints (24g). Similarly, head excursion during impact did not differ (360mm and 330mm). Despite its considerably wider side wings around the head, Booster B failed to contain the dummy head during the rebound phase and the amount of head motion was considerably greater than for Booster A (680mm and 260mm, respectively). In crash configurations with multiple rear seat occupants, this could place the Booster B occupant at risk of an occupant-to-occupant collision. Given the very low head acceleration values, it could be argued that head contact was not problematic. However, it is noted that this result occurred in a relatively low crash speed; hence, it would be prudent to repeat the test at high crash speeds and with multiple rear seat occupants.

Table 2. Summary of dummy measures for side impact tests with 6 year old dummy

Restraint Type	Head Accel Peak (g)	Max Head Excursion (mm) Imp	Max Head Excursion (mm) Rb	Head Contact
Booster A Seatbelt	24	360	260	No
Booster B Seatbelt	24	330	680	No

Comparison of restraint systems with a 3 year old dummy

Figure 3 shows the Peak Head Acceleration values for tests with the 3 year old dummy. The head acceleration for Booster A, used with harness and crotch strap, was 10g higher than for Booster B tested in the same configuration (78g and 68g, respectively). In addition, when used with the harness and crotch strap, both boosters performed better than the test in which the 3 year old dummy was restrained in a seatbelt only (102g). Notable increases in Peak Head Acceleration were evident for the boosters plus harness combination when the crotch strap was disengaged. Indeed, in the case of Booster A, these 'misuse' simulations yielded head acceleration values that were comparable to the seatbelt only condition.

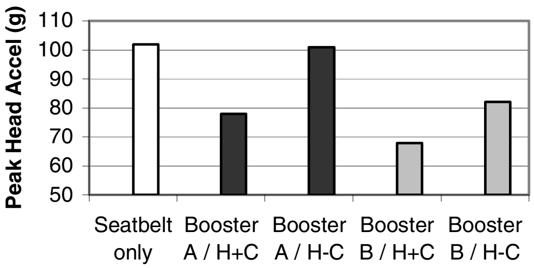

Figure 3. Peak Head Acceleration for tests with HIII 3 year old dummy

Results for N_{ij} Tension Flexion for tests with and without a crotch strap followed a similar pattern to the head acceleration values as discussed above (see Figure 4), with higher neck injury values observed when the crotch strap was disengaged. Interestingly, N_{ij} values for the seatbelt only condition did not differ from the boosters with the harness and crotch strap.

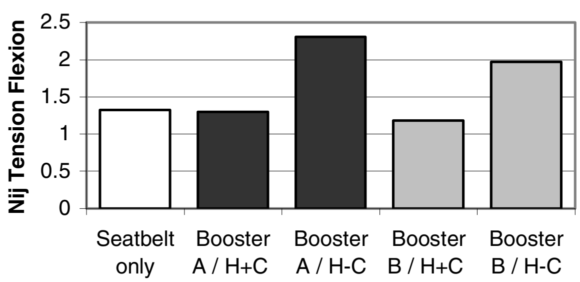

Figure 4. N_{ij} Tension Flexion for tests with HIII 3 year old dummy

Results of analyses of the video data are summarised in Table 3. The maximum forward motion of the dummy head at the time of impact did not vary greatly across restraint types (range was 386 – 443 mm). There was no evidence of head contact with either the vehicle interior or with the dummy's knees. However, sub-marining was evident for both boosters when the crotch strap was not engaged.

Table 3. Summary measures for frontal tests with 3 year old dummy

Restraint Type	Max Head Excursion Impact Phase (mm)	Head Contact	Sub-Marine
Seatbelt only	443	No	No
Booster A			
Seatbelt/H+C	426	No	No
Seatbelt/H-C	401	No	Yes
Booster B			
Seatbelt/H+C	386	No	No
Seatbelt/H-C	397	No	Yes

Figures 5 and 6 show the post-test dummy position for frontal tests for Booster A. Figure 5 shows the dummy restrained with a harness and crotch strap, remaining well positioned at the end of the test. Figure 6 demonstrates the sub-marining effect that resulted when tested with the crotch strap disengaged. Booster B with harness and crotch strap disengaged also resulted in the same sub-marining effect. As demonstrated in Figure 6, without the crotch strap, the dummy slides forward and under the lap portion of the adult seatbelt. This effect is likely to place the occupant at serious risk of injury.

Figure 5. Frontal test for Booster A with seatbelt and harness with crotch strap

Figure 6. Frontal test for Booster A with seatbelt and harness with the crotch strap disengaged

CONCLUSIONS

This study aimed to explore the suitability of two boosters for use with children across the age range for which toddler child seats and boosters would be appropriate. The motivation for this was that if parents were offered a single seat (a hybrid child seat/booster) that could take the child through the transition from forward-facing restraint to booster, that this might promote greater use of boosters amongst older children and reduce the complexity of decisions about what restraint might be appropriate for a child once they 'graduate' from the forward-facing child seat.

The results demonstrated that the two boosters selected for this study provided a suitable option for children represented by the 3 year old and 6 year old dummy. Based on head acceleration and neck injury measures, these restraint systems provided superior protection to that of an adult seatbelt. Importantly, no head contact was observed with the vehicle interior during any of the tests.

Of some concern, however, was the considerable lateral motion of the dummy and failure to contain the head during the rebound phase in the side impact test for Booster B. In contrast, Booster A restrained the 6 year old dummy in a good position throughout both impact and rebound phases. The result for Booster B was somewhat surprising given its considerably larger side wings and higher back. These findings need to be explored further at higher impact speeds.

An important finding was that when the H-harness was used to restrain the 3 year old dummy, correct use of the crotch strap was critical in eliminating 'sub-marining'. The effect of sub-marining places the occupant at serious risk of injury to the neck region, including vital airways, blood vessels and spinal cord. This finding raises serious concerns, given the relative ease with which the crotch strap can be disengaged by a child occupant during a trip, or not engaged with the lap portion of the adult seatbelt when fitting the child into the restraint.

The simulated misuse errors highlight the need for a crotch strap connection mechanism that cannot be easily disengaged. Ideally, this might be a mechanism similar to that used in the integrated 6-point harness provided in forward-facing restraints for toddlers. If the same booster is to be used with a seatbelt for older children, the design would need to allow for such an integrated harness to be removed.

Several other areas for design improvement should be explored further. For example, it will be important to develop a design feature that would allow the ISOFIX connectors to telescope into the child seat when used as a booster seat for older children. Alternatively, it is possible that the ISOFIX anchorages could offer a desirable method of attachment for both booster configurations. In addition, there is a need to consider the child restraint and rear seatbelt restraint as an integrated system. For example seatbelt pretensioners and belt load limiters should offer desirable solutions.

Limitations

The validity of these outcomes is constrained by the limited biofidelity of the dummies. It would be expected that the human body, being less stiff than a dummy, would be subjected to greater excursions and hence is more likely to contact the vehicle interior in the event of a crash. While the Hyge sled tests presented here provide useful information about the interaction of both dummy and restraints in a real vehicle, they do not demonstrate the likely effects of intrusions, particularly in a side impact crash. Further research is needed to examine intrusion effects using full-scale vehicle crash tests. In addition, it would be prudent to conduct more tests to gain a full set of data across the various restraint types in side impact and also to verify the repeatability of key test outcomes.

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ASSESSING NEW CHILD DUMMIES AND CRITERIA FOR CHILD OCCUPANT PROTECTION IN FRONTAL IMPACT

Kate de Jager, Michiel van Ratingen
TNO Science and Industry (The Netherlands)
Philippe Lesire, Hervé Guillemot
LAB (France)
Claus Pastor, Britta Schnottale
BASt (Germany)
Gonçal Tejera
Applus+IDIADA (Spain)
Jean-Philippe Lepretre
UTAC (France)

On behalf of EEVC WG12 & WG18
Paper no. 05-0157

ABSTRACT

The European Enhanced Vehicle-safety Committee wants to promote the use of more biofidelic child dummies and biomechanical based tolerance limits in regulatory and consumer testing. This study has investigated the feasibility and potential impact of Q-dummies and new injury criteria for child restraint system assessment in frontal impact.

European accident statistics have been reviewed for all ECE-R44 CRS groups. For frontal impact, injury measures are recommended for the head, neck, chest and abdomen. Priority of body segment protection depends on the ECE-R44 group.

The Q-dummy family is able to reflect these injuries, because of its biofidelity performance and measurement capabilities for these body segments. Currently, the Q0, Q1, Q1.5, Q3 and Q6 are available representing children of 0, 1, 1.5, 3 and 6 years old. These Q-dummies cover almost all dummy weight groups as defined in ECE-R44. Q10, representing a 10 year-old child, is under development.

New child dummy injury criteria are under discussion in EEVC WG12. Therefore, the ECE-R44 criteria are assessed by comparing the existing P-dummies and new Q-dummies in ECE-R44 frontal impact sled tests. In total 300 tests covering 30 CRSs of almost all existing child seat categories are performed by 11 European organizations. From this benchmark study, it is concluded that the performance of the Q-dummy family is good with respect to repeatability of the measurement signals and the durability of the dummies. Applying ECE-R44 criteria, the first impression is that results for P- and Q-dummy are similar.

For child seat evaluation the potential merits of the Q-dummy family lie in the extra measurement possibilities of these dummies and in the more biofidelic response.

INTRODUCTION

Each year, 700 children are killed on European roads and 80,000 are injured [1]. It represents an unacceptably high burden on Europe's society and economy. The fact that such poor results are observed, despite normal use of CRS (Child Restraint Systems) complying with the ECE 44 Regulation, underlines the high social importance of continued child safety research. Despite many initiatives being taken in Europe and elsewhere, progress made in child safety in the last decade can be considered small, in particular compared to the advancements made in adult occupant protection in that same period. Important contributors to this situation are the lack of biomechanical knowledge on injury mechanisms and associated physical parameters, specifically for children.

The European Commission (EC) has recognized that it is only through a decisive increase of the basic scientific knowledge that major steps can be achieved towards improved standards and more efficient design of CRSs. For this reason the CREST (Child Restraint Standards, 1996-2000) and CHILD (Child Injury Led Design, 2002-2006) projects were initiated to develop the knowledge on child behavior and tolerances. The outcomes of EC-CREST and EC-CHILD can be used to propose new test procedures for determining the effectiveness of CRS using improved child dummies and injury measures [1-3]. As a result of these projects the Q-series of child dummies are currently available for CRS testing [4, 5].

The European Enhanced Vehicle-safety Committee (EEVC) wants to promote the use of more biofidelic child dummies and biomechanical based tolerance limits in regulatory and consumer testing. It initiates the assessment of new child dummies and criteria for child occupant protection in frontal impact. Therefore, EEVC WG12 and WG18 carried out collaborative research following four basic steps: (i) identification of child injury causation in frontal impacts based on real world data, (ii) completion and consolidation of the specifications of the Q-series of advanced child dummies, (iii) recommendation for new injury criteria and tolerance limits for frontal impact, and (iv) a validation test program based on ECE R.44 test conditions, comparing P and Q dummy performance in frontal CRS tests. For the latter part, eleven European organizations including OEMs, research institutes and child restraint manufacturers performed 300 tests covering 30 available child seats. These seats represent the majority of existing child seat categories on the European market.

The paper starts with an overview on child injury causation. This overview presents a synthesis of frontal crash investigations including those performed under the CREST and CHILD projects. Next, the development and evaluation of the Q-dummy family (including Q0, Q1, Q-18 months, Q3 and Q6) are described. In addition, the situation regarding newly proposed child dummy injury criteria is given. Thereafter, the validation of P- and Q-dummies and criteria are described. An in-depth analysis of 300 test results covering 30 child seats will be presented, showing the effect and potential benefit of introducing new test dummies and criteria into legislation. Finally, conclusions are drawn and recommendations are given.

CHILD INJURY CAUSATION

The EEVC WG 18 on Child Safety was created in October 2000. One of the first tasks of this group was to review the European accident statistics with respect to child car occupants and injuries in all type of car crashes. For this purpose, the most important existing databases in Europe have been examined. Data from the International Road Traffic Accident Database (IRTAD) show that in 1998 on average 2 children were killed each day. The tendency for Europe over the past ten years is that the total number of children killed as car occupant is decreasing. This can be seen as one of the effects of the general adoption of a European regulation on child restraints. An overall positive effect of restraint use by children when travelling in cars is found. The rate of severe injuries is more than twice as high for unrestrained children than for restrained children in frontal impact, which is the most common crash configuration. The risk of being severely injured as car occupant is very small for correctly restrained children up to a delta V of 40 km/h in a frontal impact. However, special attention should be paid to avoid CRS misuse and to make sure that clear information is forwarded to the public area about child safety and injury risk related to accidents.

In order to draw more detailed conclusions, WG18 has accessed and examined the following databases: CREST (as developed in the European collaborative research project), CCIS (the Co-operative Crash Injury Study in the UK), GIDAS (German In Depth Accident Study), GDV (German Insurance), IRTAD and LAB (Laboratory of Accidentology and Biomechanics in France). All of these databases have specific definitions and data collection methods, which makes it difficult to merge the data for analysis. Nevertheless for frontal impact, generally sufficient information was available in each database to classify injury causation according to the different group of child restraint system used. The CRSs were put in categories according to the weight group existing in the ECE R44-03:

Carrycots (Group 0): The number of crash cases available with this kind of restraint system is too low to conclude about the general injury mechanism.

Rearward facing infant carrier (Group 0/0+): These systems seem to offer good protection to their users in frontal impact. Severe head injuries are most frequently observed injuries with such CRS suggesting that introduction of effective padding may significantly reduce head injury risk. Three different injury mechanisms are hypothesised: impact by the shell with the dashboard (67% of rearward facing infant carriers is put on front passenger seats), direct impact of the head on supporting object and rebound. For these systems, limbs are also representing a high number of injuries, but only a few are considered as severe injuries. Therefore limb injuries are of less priority.

Rearward facing system with harness (Group I): Most popular in Northern Europe, rear facing CRS have been seen to be more effective in frontal impact when compared to forward facing CRS. Severe head injuries are less frequent in frontal impact with such devices than with rearward facing infant carriers. Limbs (especially arms) can also be injured.

Forward facing systems (Group I): For this type of system head injury is still a big issue. Impacts are one cause, but diffuse brain injuries are also observed due to angular acceleration that can occur either with or without impact. The neck is an important area to protect for children in such devices (younger than 4 years of age) even if these injuries are not very frequent. Chest and abdominal injuries are not very frequent with such systems but are found.

Forward facing system with shield (Group I and shield systems (Group II): The main sources of data are from the UK and France where these devices are not very popular. Therefore, no accident data are available at this time but some observations from experts were collected. Head contact with the top of the shield and risk of ejection (total or partial) are likely scenarios causing injuries.

Forward facing seat and adult seatbelt (Group I/II/III): In most of the analysis of databases these systems were considered as booster seats (see below). In addition, the risk of neck injuries is as high as for forward facing systems with harness (see forward facing systems (group I) above).
Booster seat and adult seatbelt (group II/III): Head is still the most important body area in terms of frequency of injury, but the relative importance of abdominal injuries increases with such restraint systems. The penetration of the seatbelt in the soft organs creates injuries at the level of liver, spleen, and kidney. For these systems, the protection of the abdominal area is clearly a priority to ensure a good protection of children using a CRS on which they are restrained by the adult seatbelt. The chest does not seem to be a priority in terms of frequency of injuries, nevertheless, as the chest cavity protects vital organs, it remains an important body segment. Focussing on severe injuries, ribs fractures are not very common because of the chest compliance for children, and internal injuries occur by compression of the chest by the seatbelt. No injury due to inertial loading has been noticed. The pelvis is not a priority body region in frontal impact. Limb fractures are numerous for children on booster seats and booster cushions, but do not seem to be a priority in terms of child protection for the moment.
Booster cushion and adult seatbelt (group II/III): The situation for these systems is the same as for booster seats with an increase of the number of chest injuries, certainly due to the fact that children using these CRS are generally older than the ones using booster seats.
Adult seatbelt: It was observed that a lot of children were only restrained by the adult seatbelt, while they could be better protected by using an additional CRS. The body segments that are protected for children restrained by the adult seatbelt only are the same as for the ones using booster cushions but with worse injury outcome, especially in the abdominal region.

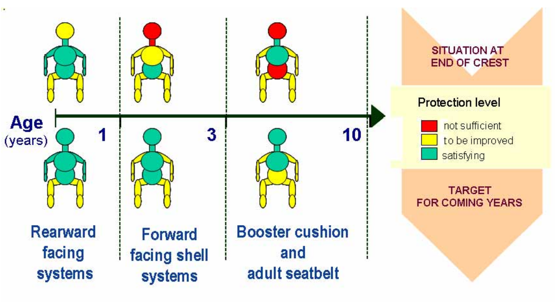

Figure 1. Level of protection for well-restraint children on appropriate CRS in the year 2000 and the target for the years thereafter.

The review of child occupant injuries related to CRS systems used in frontal impact has demonstrated that the whole priority should lie on protecting the head and neck from injury for infants and toddlers (Group 0/1), shifting to head, chest and abdomen as children grow up and starting to become taller (Group 2/3/adult belt) which is illustrated in Figure 1. It is important that new dummies and criteria reflect these injuries observed in the field. Consequently, injury measures were recommended for the head, neck, chest, abdomen/lumbar spine and pelvis.

DEVELOPMENT AND EVALUATION OF Q-DUMMIES

Background

The P-series is a series of crash test dummies representing children in the age of six weeks (P0), 9 month (P3/4), three year (P3), 6 year (P6) and 10 year (P10) old. The P-dummies ('P' from Pinocchio) were the first European child dummies to become official in 1981, when the ECE-R44 [6] regulation came into force. Later, the dummies were also adopted by other standards. The P-series, despite being simple in design and limited in measurement capability, gave a substantial contribution to the protection of children in cars. However, more knowledge on biomechanics related to children and the changing nature of exposure (airbags, belt systems) meant that the P-series became less appropriate over time. In the nineties the CRABI (Child Restraint AirBag Interaction) and Hybrid III child dummies were developed in particular to address the growing problem of child-airbag interaction in the US. In Europe, research has been focused on the development of a new child dummy series that would bring major improvements in terms of biofidelity and instrumentation and that could be used for a range of applications including side impact.

In 1993, the international Child Dummy Working Group (CDWG) was formed with the mission to develop the Q-series as the successor of the P-dummy series. This group, consisting of research institutes, CRS- and dummy manufacturers and OEM's, determined the anthropometry, biofidelity, measurement capabilities and applications for these dummies [7-10]. Under their surveillance, also the development of the first Q dummy, Q3, started. In 1997, this work was continued under the EC sponsored CREST (Child REstraint System sTandard) research program. Within the CREST and the consecutive CHILD (CHild Injury Led Design) projects, altogether the new-born (Q0), the 12-month (Q1), three-year old (Q3) and six-year-old (Q6) dummies were delivered and used in accident reconstruction. In 2003, the most recent dummy was added to the series: the Q1.5, representing a child of 18 months old. Figure 2 shows the Q-dummy family.

Figure 2. The Q-dummy family: (from left to right) Q1.5, Q3, Q0, Q6, Q1 and Q1 without suit.

Since their original release, the Q-dummies have undergone updates, in particular to improve the overall durability in frontal impacts, while maintaining the overall biofidelity and side impact performance of the dummy. The Q-dummies were particularly tailored to meet the (high-end) demands of EuroNCAP and NPACS testing [11]. This section summarises the status of the Q-dummy series today. The dummy design and performance particularly for frontal impact are described. In addition, the main differences with the US child dummy series are given.

Dummy description

Specific design features of the Q-dummies are the anatomical representation of body regions, use of advanced materials, dummy-interchangeable instrumentation, multi-directional use (frontal & side impact) and easy handling properties (limited components, easy assembly/disassembly, and simple calibration).

The dummy layout of the Q1, Q1.5, Q3 and Q6 is similar. The design of the head, the neck, the shoulder, the clavicle, the thorax, the lumbar spine, the abdomen and the extremities show a realistic anatomy compared to the human anatomy. The head and the clavicle are made entirely from plastics. The neck and the lumbar spine have a similar design: a combination of metal and a natural rubber. It is flexible and allows shear and bending in all directions. The thorax consists of a deformable ribcage and a rigid metal thoracic spine. The plastic clavicle is connected to the thorax at the front of the ribcage and to the shoulders at the arm side. The shoulders are made of natural rubber with metal end plates which are connected to the upper arm on one side and the thoracic spine on the other side. The lumbar spine is mounted between the pelvis and the thoracic spine. The abdomen is skin covered foam, which is enclosed by the ribcage and the pelvis. The pelvis consists of two parts: a metal pelvic bone and a plastic pelvis flesh. The extremities are a combination of plastics and metal. The Q1, Q1.5, Q3 and Q6 have a kinematical representation of the elbow, shoulder, hip and knee joints.

The anthropometry of a new-born child makes it difficult to maintain the dummy lay-out of the other Q-dummies for the design of the Q0. The limited space reduces the anatomical representations of body parts. For the Q0, its design results into eleven body parts: head, neck, shoulder block, two arms, thoracic spine, lumbar spine, thoracic flesh, pelvis block and two legs. The materials used are similar to those used in the other Q-dummies. The legs and arms have no knee and elbow joints, respectively; instead, the angles between upper and lower leg and upper and lower arm are fixed. The torso flesh foam part represents the ribcage and the abdomen. It is made of foam covered by a vinyl skin. The neck and lumbar spine have a similar design [4].

<u>Anthropometry</u> - To establish human-like dimensions for the Q-dummies, a special Child Anthropometry Database, CANDAT, has been built [8]. For this database, the newest available child data from birth to 18 years have been collected from different regions (US, Europe and Japan). Inconsistencies have been solved and gaps have been filled to calculate the averages for important body dimensions and mass (Figure 3).

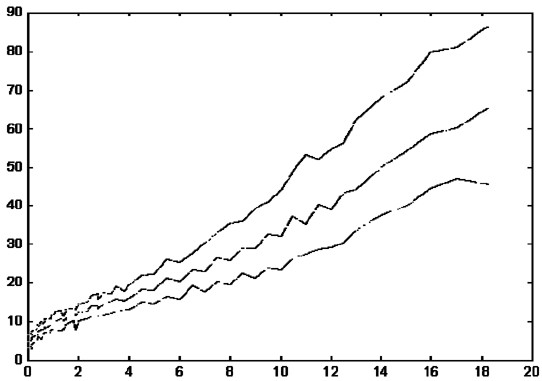

Figure 3. 5th, 50th and 95th Percentile child body mass (in kg) vs. age (in years) in CANDAT.

For adoption of the Q-dummy series, it is important that the body mass corresponds with the manikin body mass as defined in the regulations. In ECE-R44, a child restraint system falls into one of the five defined mass groups. Each mass group has a lower and upper boundary. Therefore, two child dummies are necessary to validate a child restraint system. Below, in Table 1, the body mass of the Q-dummy series is compared with the weight groups of ECE-R44. In Annex A, the main dimensions and the segment masses of each Q-dummy are compared with the manikin requirements as defined in ECE-R44.

Table 1.
ECE-R44 mass groups with corresponding Q-dummy

R44 Group	Limits	R44 Mass [kg]	Dummy age	Dummy mass [kg]
0	Lower	-	Q0	3.4
	Upper	<10	Q1	9.6
0+	Lower	-	Q0	3.4
	Upper	<13	Q1.5	11.1
I	Lower	9	Q1	9.6
	Upper	18	Q3	14.6
II	Lower	15	Q3	14.6
	Upper	25	Q6	22.9
III	Lower	22	Q6	22.9
	Upper	36	-	-

It can be observed that the mass groups are covered by the Q-dummy series except for the upper boundary of a group III seat. This Q-dummy is not yet available. The segment masses and the main dimensions of the Q-dummy series are slightly different from the manikins as defined in ECE-R44 which are based on the P-dummy anthropometry. The Q-dummy family, however, is based on a more recent anthropometric database (CANDAT).

Biofidelity - The availability on biomechanical data of children is poor due to the ethical difficulties with obtaining such data. Therefore, the following approach was chosen to derive a set of biomechanical response requirements for the Q dummy series. First, a set of human body responses to frontal and side impact have been discussed [12-17]. Second, a study was made of the characteristics of the human body, both of adults and children [9, 10]. Finally, scaling methods, combined with the data on human body tissue characteristics were used to derive child response characteristics from adult data. The scaling is based on differences between adult and child subjects in terms of geometry and stiffness [18-21]. For frontal impact, biofidelity requirements have been set-up for the head, the neck, the thorax, the abdomen and the lower extremities. For lateral impact the set of biofidelity requirements is extended with requirements for the shoulder and the pelvis. It should be noted that due to the (many) assumptions made in the scaling process, these requirements should be treated as design targets rather than strict specifications.

For the assessment of the biomechanical response in frontal impact, the head, the neck, the thorax and the abdomen are considered the most important body parts (head and neck only for Q0).

The biomechanical target of the Q-dummy heads is based on the rigid surface cadaver drop tests conducted by Hodgson and Thomas [22]. The head biofidelity for frontal impact has been assessed by a free-fall head drop test with a drop height of 130 mm. Table 2 shows the head biofidelity test results for the Q-dummy family.

Table 2.
The head biofidelity target vs. test result for the Q-dummy family

Peak resultant head acceleration [G]		
Q-dummy	Target	Test result
Q0	124 ± 33	120 ± 3
Q1	108 ± 29	112 ± 1
Q1.5	111 ± 29	111 ± 2
Q3	121 ± 29	116 ± 1
Q6	139 ± 37	122 ± 1

The neck response requirement for flexion-extension has been established by scaling human volunteer and cadaver data of Mertz and Patrick [23]. The assessment of the neck biofidelity head-neck pendulum test responses were performed for the assessment of the neck biofidelity of the Q-dummy series. Figure 4 shows the biofidelity performance of the Q3 dummy for flexion.

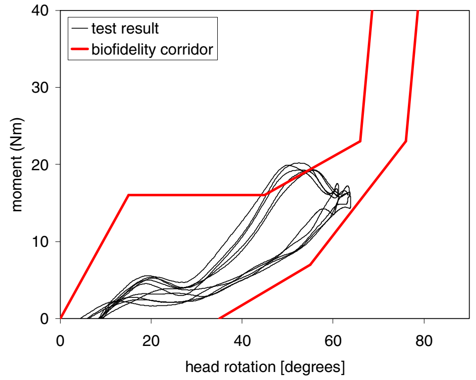

Figure 4. Q3 neck biofidelity corridor for flexion vs. test results.

The thorax frontal response requirement is based on two series of blunt-frontal, mid-sagittal impactor tests reported by Kroell [24, 25], Nahum [26] and Stalnaker [27]. Thorax impactor tests, using a dummy specific pendulum, were performed to assess the biofidelity of the thorax. Two different impact velocities are used, 4.3 m/s and 6.7 m/s. Figure 5 shows the thorax biofidelity performance of Q3 compared to linearly scaled corridors. It should be noted that linear scaling does not take into account damping and therefore is likely to underpredict the true force response of the actual child [28].

For the abdomen, a frontal belt loading requirement has been defined. It is based on living porcine experiments [16, 29]. Previous abdomen tests indicated that the segment is meeting the corridors, but additional test are being run to document the abdominal response for all dummies. The complete biofidelity results of the individual Q-dummy family dummies will be reported separately to the EEVC along with the recommendation for its use. From the data available at this time it is concluded that the biofidelity responses of the head (see Table 2) and the neck of all Q-dummies are within the corridor. The biomechanical performance of the Q1, Q1.5, Q3 and Q6 thorax show that it is a bit stiffer than its (linearly scaled) target, in particular at lower impact velocity.

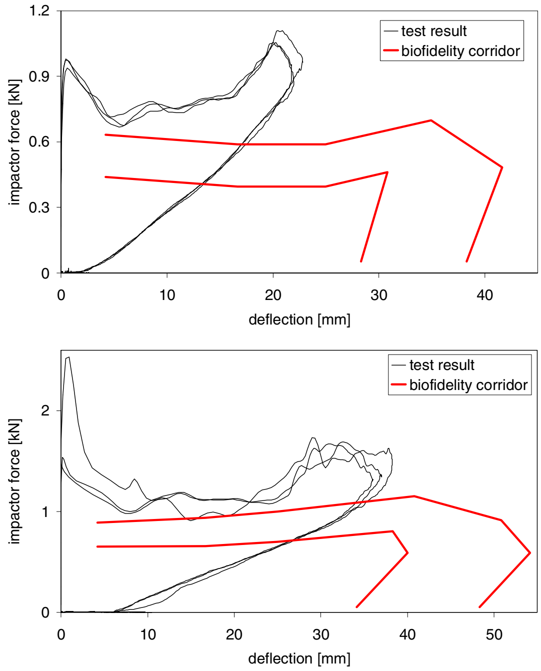

Figure 5. Q3 thorax biofidelity corridors vs. test results for 4.3 m/s (upper graph) and 6.7 m/s (lower graph) impactor velocity.

<u>Injury assessment</u> - The Q-dummy series allow the measurement of a number of responses covering the needs that follow from the field accident research. The set of instrumentation is similar for Q1 and Q1.5 and for Q3 and Q6. The type of load cells, the head angular velocity sensors and the accelerometers are generally interchangeable for all Q-dummies. Figure 6 shows the set of the instrumentation for Q1.5.

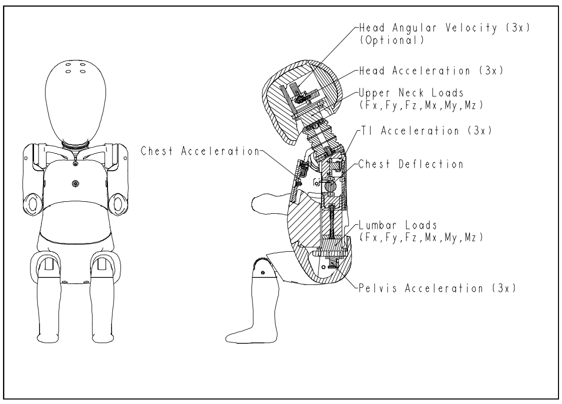

Figure 6. Q1.5 dummy instrumentation set.

In addition to the set of instrumentation for Q1/ Q1.5, the Q3 and Q6 have a lower neck loadcell (6-axis). Q3 and Q6 abdominal sensors are under evaluation [1, 30]. For Q0, the set of instrumentation is limited compared to Q1/Q1.5 due to space and performance limitations. The Q0 can, due to its size, only be equipped with head, T1 and pelvis accelerometers (3-axis) and an upper neck loadcell (6-axis).

<u>Durability</u> - The anticipated use of the Q-dummy series in EuroNCAP full-scale and NPACS body-in-white/sled testing make that the dummies have to be durable under test conditions that are more severe than ECE-R44. The definition and assessment of the durability level required for the Q-series are assessed on the ECE-R44 sled equipped with a rigid wooden seat instead of a CRS [31]. The crash pulse is based on a generalized vehicle B-pillar acceleration taken from actual EuroNCAP tests. The Q1 and Q1.5 are restrained with a 5-point belt over the shoulders and upper legs. For Q3 and Q6 a standard 3-point belt system is used. Thirty tests were carried out with each dummy with intermediate visual inspection to ensure that the dummies meet the durability requirements without any damage. It is concluded that the Q-dummies sustained the durability tests showing no damage.

<u>Repeatability</u> - The level of repeatability of dummy responses is often expressed in the coefficient of variation. For adult dummies, a coefficient of variation of 10% is considered to be acceptable. In case of child dummies, the coefficient of variation in sled tests depends on more factors compared to adult dummies. For example, in most test conditions the child dummies are restrained in a CRS and the CRS is attached with the car belt, both adding variability to the system. To assess the repeatability of the Q-dummy series the peak responses in the durability sled tests of Q1 and Q3, as described above, were used. In this test program the interference of a CRS has been avoided. The peak responses from the biofidelity test results of all Q-dummies have also been analysed to assess the repeatability of the Q-dummies. It shows that the coefficient of variation is 12% or less for relevant dummy measurements, which is considered acceptable from the user point of view.

<u>Certification</u> - For frontal impact tests a certification test is derived for the head, the neck, the thorax, the lumbar spine and the abdomen. All certification tests are component tests carried out on individual components or the full dummy. For Q0, only the head and neck have certification requirements. To perform the certification tests special equipment is required: a head drop table, a wire suspended pendulum for the thorax impactor test, a dummy specific pendulum (weight and diameter are dummy specific), an abdomen compression device, a part 572 pendulum and a dummy specific head form for the neck and lumbar spine certifications. The certification procedures and criteria are described in the respective dummy manuals [32, 33].

The recommended frequency of Q-dummy certification and the number of tests that can be performed between certifications strongly depends on the number, type and severity of the tests in which the dummy is used. Which certification tests have to be carried out depends on the dummy application (ECE-R44, NCAP) performed.

Comparison with US Child Dummy series

It is recognized that the development phase of the Q-series has largely run parallel in time to the development and enhancement of the Hybrid-III series in the US. The Hybrid-III family is fundamentally different from the Q-series in terms of design philosophy (scaling methodology), lay-out, and source information used.

In 1987 the development of the CRABI and Hybrid III child dummies started by two SAE task groups, the Hybrid III dummy family task group and the Infant dummy task group. The CRABI (Child Restraint Air Bag Interaction) dummies represent children in the age of 6, 12 and 18 month old for use in assessing airbag interactions with rear facing child restraints. The Hybrid III child dummies are representatives of 3, 6 and 10 years old children. These dummies are designed primarily for use in frontal loading conditions, with special attention given to OOP (Out-Of-Position) test conditions [34]. The anthropometry of these dummies has been derived from children in the United States. The biofidelity requirements were obtained by scaling the biomechanical response corridors for the mid-size adult male that were used to develop the Hybrid III dummy [34, 35], using dummy dimensions.

The main differences between the US child dummies and the Q-dummy series are identified on the anthropometry sources used, the biofidelity and the application. The anthropometry of the US child dummies focuses on US-databases, whilst the Q-series is based on a more global anthropometry. The set of biofidelity requirements as defined for the Q-dummy series is more elaborated than for the US child dummies. It has resulted in different mass and weight distribution between the two dummy types. The US child dummy biofidelity concerns mainly head, neck (and chest for the older dummies) requirements while the Q-dummy series also have requirements for all relevant body regions in front and side impact. The interpretation of biofidelity also shows differences. For example the head biofidelity requirement of the Q-dummy series is based on the non-fracture zone of impact while the CRABI and HIII child dummy head requirement focuses on the fracture zone. The Q-dummy series have a different field of application than the US child dummies since the Q-dummy series are optimised for CRS testing in ECE and side impact testing, while the US dummies have their background in airbag interaction testing and are used in FMVSS 213 and FMVSS 208.

CHILD DUMMY INJURY CRITERIA

One of the most challenging tasks in child safety is to establish correlations between the child injuries and child dummy measurements. Biomechanical tests with child subjects are undertaken very seldom, for obvious ethical reasons. Besides, a child is not a “small” adult and the scaling approach does not allow the direct transfer of knowledge from adult to child. For these reasons, crash test reconstructions of actual crashes with fully instrumented dummies having a comparable anthropometry, constitutes a right and appropriate methodology to acquire the missing biomechanical knowledge relative to the children. This approach is taken in the EC-CREST and EC-CHILD projects. It is clear, however, that this methodology requires many reconstructions to be performed. At this point in time insufficient reconstructions have been carried out to recommend new injury limits for all dummies. It is expected that the EC-CHILD project will supply sufficient reconstructions by mid 2006.

What is available at this time is based on child free-fall studies, aircraft field investigations and animal testing combined with response scaling from adults and dummies. The Hybrid III child dummies series have the most extensive set of injury criteria, based on the scaling methodology developed by Irwin and Mertz [35]. For the head, neck, chest and lower extremities injury criteria are determined:. For the P-dummies the set of injury criteria is limited to the head and the chest. These criteria are described in ECE-R44.

Injury criteria for the Q-dummy family have yet to be reviewed by EEVC WG12. Awaiting the outcome of the EC-CHILD project, the scaling methodology as used for the Hybrid III child dummies has been studied by EEVC WG12 and may be applied to the Q3-dummy head, neck and chest criteria. This has proven to be less straight forward as expected since biofidelity responses of the dummies are not identical. The results of the P-dummy and Q-dummy comparison presented below therefore focuses firstly on the existing ECE R44 criteria. In addition, the extra measurements taken for the Q-dummies are assessed with regards to their potential merits for child seat evaluation.

VALIDATION OF DUMMIES & CRITERIA

In this paper, the Q-dummies are compared to the existing P-dummies in an extensive validation program performed by eleven European organizations including OEMs, research institutes and child restraint manufacturers. Below, the test set-up and test matrix are described in detail. In addition, the data analysis with its preliminary test results is given.

Test set-up

The test procedure is essentially based on the current ECE-R44 (status of 4th February 2004; including Supplement No. 6). The test series exclusively focuses on the dynamic test procedure as described by ECE-R44 paragraph 8.1.3, frontal impacts. However, on the following points the test procedure deviates from the ECE-R44 dynamic test protocol. Firstly, only frontal impact sled tests are performed. No tests on trolley and vehicle body shell (ECE-R44 8.1.3.2) or tests with a complete vehicle (ECE-R44 8.1.3.3) have been conducted. Secondly, CRS with support legs (ECE-R44 7.1.4.9) have been tested. The test laboratory has chosen one suitable position for the support leg and has repeated this test. The position of the support leg on the floor is photographed. Thirdly, for all classes of ISOFIX CRS (ECE-R44 7.1.4.10) it is decided to perform one test with the anti-rotation device in use, if any. One change from the specification, given in Annex 6 of ECE-R44, is that the EEVC WG12/18 program allowed the use of a double sled with two benches on the trolley. Furthermore acceleration and deceleration based sleds are allowed.

The complete Q-dummy family is assessed and from the P-dummy family the P10 is excluded.

Table 3.
Assessment of dummies for a CRS per ECE-R44 group

Dummy		ECE-R44 group			
		0+	I	II	III
Small	P	P0	P¾	P3	P6
	Q	Q0	Q1	Q3	Q6
Intermediate	P	P¾	P1.5	-	-
	Q	Q1	Q1.5	-	-
Large	P	P1.5	P3	P6	-
	Q	Q1.5	Q3	Q6	-

Both dummy families are fully instrumented. Modelling clay for the P dummies is only used for appropriate kinematics and not as injury risk assessment. The Q3 and Q6 abdominal sensors are under evaluation in the EC-CHILD project and therefore not included in the dummies. The temperature of each child dummy has been stabilised in the range of 18°C to 22°C. Table 3 shows the assessment of dummies for a CRS per ECE-R44 group.

To fix the dummy position in the pre crash phase, masking tapes on the heads and arms are used, if necessary. Each test is repeated once with a new CRS. In case of breakage of the CRS, breakage of the dummy or "strong differences" between the two conducted tests, a third test is conducted.

Test matrix

The test matrix covers almost all existing CRS categories, including rear infant carry cot (isofix/universal), seats with harness (forward/ rearward, isofix/universal), shield systems (isofix/ universal), boosters with backrest, booster cushions and multi-group. Therefore 30 CRSs are selected and 300 tests are Table 4 summarizes the test matrix. More details are given in Annex B.

Table 4.
Test matrix P- & Q-dummy comparison

Type of CRS	# of tests	# of CRS
G0+ RWD FC	**68**	**6**
Infant carrier universal	36	3
Infant carrier isofix basis	12	1
Combination CRS used RWD	8	1
Combination CRS-RWD isofix	12	1
GI FWD & RWD HARNESS	**116**	**11**
FWD FC universal	64	6
FWD FC isofix + top tether	20	2
FWD FC isofix + support leg	12	1
RWD FC classical (non-isofix)	8	1
RWD FC isofix	12	1
GI FWD SHIELD	**12**	**1**
FWD FC isofix	12	1
BOOSTER + BACK	**32**	**4**
Universal	32	4
MULTI I,II,III same config	**40**	**3**
Universal	40	3
MULTI I, II, III differ config	**52**	**5**
Universal – shield	20	1
FWD universal – harness	32	4
	300	**30**

Data analysis

For the data analysis a database has been developed to compile all test results: measurement signals, photographs and videos. In addition, a summary of all test results per laboratory will be given. Because of the fact that the test program is recently completed, only a preliminary data analysis can be conducted at this time. This analysis focuses on current ECE-R44 requirements, dummy kinematics, and extra measurement signals of the Q-dummies.

ECE-R44 – It is known that sled acceleration will influence the dummy measurements of ECE-R44. In addition, it is known that different types of CRS belonging to the same ECE-R44 group may show slightly different outcomes. Therefore, for a valid comparison between P- and Q-dummies, only tests where both P- and Q-dummies are tested at the same test facility and in the same CRS are selected for the data analysis. For the maximum head excursions, this means that 276 of in total 300 tests are available for the current analysis. For chest accelerations, data on ECE-R44 requirements are not yet complete at the time of this analysis. Therefore, only a subset of 106 tests can be used for this analysis.

Kinematics – The dummy kinematics are studied by analyzing the videos of the test and the timing of the maximum head excursions. The results of the timing are obtained in the same manner as the maximum head excursion itself.

Extra measurements for Q-dummies – The analysis of the extra measurement signals taken for the Q-dummies focuses on the findings of the individual laboratories. Injury criteria for these extra measurement signals are not yet available as mentioned before.

Data are expressed as means with the standard error of the mean (s.e.m.). A 95% confidence interval for the mean can be determined as mean ± 1.96*s.e.m.. The s.e.m. gives an impression of the variability of the estimated mean (standard deviation is s.e.m. * n). Differences between means were tested by t-tests. A t-test probability of $p<0.05$ is considered as statistically significant.

Test results

The test program is performed without any notifying problems. The Q-dummy family shows good durability under ECE-R44 frontal impact test conditions. No Q-dummy part has been replaced during the test program.

ECE-R44 – The preliminary results of the P- and Q-dummy comparison according to ECE-R44 requirements for head and chest are summarized in Table 5 and Table 6, respectively.

The similarity of the sled acceleration for the P- vs. Q-dummy tests is evaluated. This is done by comparing the mean values and the s.e.m. of the maximum sled accelerations for both P- and Q-dummies per ECE-R44 group. For all dummy comparisons, t-test results have shown that the maximum sled acceleration can be considered to be similar for P and Q. This means that if the input of the test (maximum sled acceleration) is similar for P- and Q- dummies, the dummy responses (head excursion) may be compared. The head excursions are compared between P- and Q-dummies in the same manner as the maximum sled acceleration. The maximum sled acceleration is compared between P- and Q-dummies by determining the mean and s.e.m..

Table 5.
Comparison of the maximum sled acceleration and head excursion in X and Z for P vs. Q in ECE-R44 group 0+, I and II

ECE-R44 group 0+ CRS				
Dummy	**P0**	**Q0**	**P1.5**	**Q1.5**
Maximum sled acceleration [G]				
mean	24.6	24.6	23.1	22.8
s.e.m.	0.8	0.8	0.7	0.6
n	6		6	
Head excursion in X [mm]				
mean	465	455	581	584
s.e.m.	17.5	16.5	33.4	30.1
n	6		6	
Head excursion in Z [mm]				
mean	459	459	598	632
s.e.m.	29.4	20.4	22.9	3.8
n	6		6	
ECE-R44 group I CRS				
Dummy	**P¾**	**Q1**	**P3**	**Q3**
Maximum sled acceleration [G]				
mean	23.0	22.7	22.4	22.4
s.e.m.	0.3	0.3	0.3	0.4
n	22		22	
Head excursion in X [mm]				
mean	399	398	457	457
s.e.m.	13.7	15.4	18.7	18.9
n	22		22	
Head excursion in Z [mm]				
mean	432	437	494	499
s.e.m.	57.0	60.5	60.5	64.0
n	20		22	
ECE-R44 group II CRS				
Dummy	**P3**	**Q3**	**P6**	**Q6**
Maximum sled acceleration [G]				
mean	23.5	24.1	22.4	23.1
s.e.m.	0.8	0.6	0.3	0.3
n	12		10	
Head excursion in X [mm]				
mean	396	389	490	453
s.e.m.	33.0	32.6	14.8	20.9
n	12		10	
Head excursion in Z [mm]				
mean	424	414	485	448
s.e.m.	72.6	93.9	81.0	92.9
n	12		8	

Table 6.
Amount of tests (in %) in which the max. res. chest acc. and the max. z-chest acc. are above 55 G and 30 G, respectively, for P vs. Q in ECE-R44 group 0+, I and II

Resultant chest acceleration > 55G			
ECE	**Dummy type**	**P**	**Q**
0+	P0, Q0 (n=4)	n.a.*	0
	P1.5, Q1.5 (n=8)	25	25
I	P¾, Q1 (n=14)	0	21
	P3, Q3 (n=13)	38	31
II	P3, Q3 (n=6)	67	33
	P6, Q6 (n=8)	25	0
Chest acceleration in Z-direction > 30G			
ECE	**Dummy type**	**P**	**Q**
0+	P0, Q0 (n=4)	n.a.*	0
	P1.5, Q1.5 (n=8)	25	25
I	P¾, Q1 (n=14)	0	21
	P3, Q3 (n=13)	38	31
II	P3, Q3 (n=6)	67	33
	P6, Q6 (n=8)	25	0

* no instrumentation

Table 5 shows that head excursions for P- and Q-dummies are similar under similar test conditions. None of the comparisons between P- and Q-dummy head excursions show statistical significant differences. This means that P- and Q-dummies do not discriminate for head excursion under ECE-R44 conditions.

The results in Table 6 give the impression that the P-dummies more frequently exceed the maximum resultant chest acceleration of 55 G and also the maximum chest acceleration in z-direction of 30 G. This is caused by the less stiff thorax of the Q-dummies. These findings are in line with previous comparison studies of P- and Q-dummies [31]. A statistical data analysis on the chest measurements has yet to point out if P- and Q-dummies show significant differences for these ECE-R44 chest criteria.

Kinematics – The video analysis shows two major differences in the general kinematics of P- and Q-dummy comparison. Firstly, the Q-dummy reaches a less 'wrapped' or 'pinned' position during the whole movement compared with the P-dummy. In ECE-R44 group I and II, the P-dummy rotates first upwards, then flexes forwards and so far downwards that the P-dummy head contacts the legs while, in most of the tests, the Q-dummy starts immediately with bending forwards and downwards. Secondly, the video analysis shows that the rebound of the Q-dummy starts earlier than for the P-dummy. These findings can be explained by the differences in dummy neck and thoracic-lumbar spine design. The Q-dummy neck design is able to induce neck moment. The P-dummy neck consists of an inner core of nylon rings and an outer shape made of urethane rings. This neck design makes it impossible to induce neck moments. For the thoracic-lumbar spine, the Q-dummy design has a lumbar spine which is similar to the neck design. Therefore the lumbar spine in the Q-dummy is also able to induce neck moments. The thoracic-lumbar spine of the P-dummy is completely rigid, which explains the large rotation in the pelvis.

The mean and the s.e.m. are also determined for the timing of the maximum head acceleration and the maximum sled acceleration. For ECE-R44 group 0+ and group II none of the comparisons between P- and Q-dummy head excursion timings show statistical significant differences. This result is also found for P¾ vs. Q1 in ECE-R44 group I. Not statistically significant are the results of the timing of the head excursions in X-direction for P3 vs. Q3 in ECE-R44 group I. It is assumed that an explanation for this result will be found by investigating the videos and the complete set of dummy measurements in parallel.

Although this preliminary result shows kinematical differences between P- and Q-dummies, the results for the ECE-R44 head requirements are not influenced by these findings.

Extra measurements for Q-dummies – The results of the extra measurements taken for the Q-dummies show a good repeatability of the test results for one Q-dummy type in the same CRS. The preliminary results indicate that the Q-dummies can discriminate between different CRSs in one type of ECE group, which is illustrated in Figure 7. These findings promote the added value of Q-dummies for child seat evaluation.

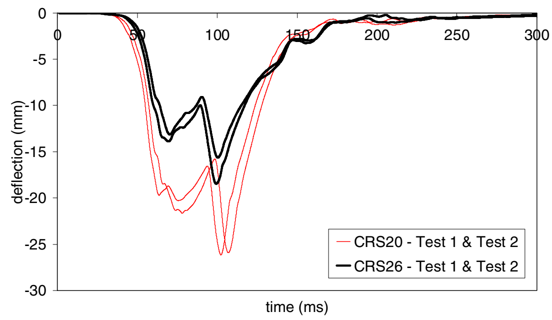

Figure 7. Chest displacement of Q3 in two different seats of ECE-R44 group I.

DISCUSSION & CONCLUSIONS

Within the EEVC, work is performed in order to promote new advanced child dummies and criteria for regulatory and consumer testing of child seats. The work, as presented in this paper, focuses on child occupant protection in frontal impact.

Starting from real world child injuries, priorities have been established with regards to what injuries are observed for what child ages and child seat types. The review of child occupant injuries covers all ECE-R44 CRS groups and the adult seatbelt. It has demonstrated that the priority should lie on protecting the head and neck from injury for infants and toddlers (ECE-R44 CRS group 0/0+/I), and the head, chest and abdomen for the older children (ECE-R44 CRS group II/III and adult belt).

The new child dummies, the Q-series, are able to reflect these injuries. More knowledge on biomechanics resulted in a new child dummy family which is more biofidelic and applicable for a range of applications. The Q-dummy family consists of Q0, Q1, Q1.5, Q3 and Q6 representing a newborn, 12 months, 18 months, 3 years and 6 years old child, respectively. These ages of the Q-dummies currently include the most important sizes required for testing the majority of child seats available on the market. However, in comparison with the P-dummy family, a dummy representing a child of 10 years is not available in the Q-series. The background information on which the Q-dummies are developed is collected and derived with ECE-R44 and side impact testing in mind. Through European cooperation (CDWG, EC-CREST, EC-CHILD) specifications have been agreed and dummies have been developed and validated. In this study, only the frontal impact biofidelity requirements are evaluated. For the head and the neck, the Q-results are within the corridor. The Q-thorax response is too stiff for its (linearly scaled) target. The measurement capabilities of the Q-series cover all needs of the injury causation study, except for the Q3 and Q6 abdomen. Abdominal sensors for these two dummies are under evaluation. In the final phase of development, most effort has gone into ensuring that the durability of the Q-dummies is up to the standard required for ECE, EuroNCAP and NPACS testing.

New child dummy injury criteria are under discussion in EEVC WG12. Therefore, the ECE-R44 criteria are assessed by comparing the existing P-dummies and new Q-dummies in ECE-R44 frontal impact sled tests. In this study, the most popular child seat configurations on the European market are taken into account. In total 300 tests are performed.

From the validation program, it can be concluded that the Q-dummy family is durable and the measurements show good repeatability. Applying ECE-R44 criteria, the P- and Q-dummy show similar results for head excursion in x- and z-direction. An in-depth analysis on the chest results of the P- and Q-tests is required, to be able to compare P- and Q- dummies according to the ECE-R44 chest criteria. Note that the actual velocity change of a deceleration sled is typically 52 to 54 km/h. This is more than the prescribed test speed of 50 +0/-2 km/h due to the rebound, which is typical for the ECE-R44 deceleration sled. At the time of this analysis, investigations towards the velocity change of the sleds were not yet completed. However, the similar maximum sled accelerations for all P- and Q-dummy comparisons indicate that the influence of the actual velocity change does not affect the outcome for the ECE-R44 head requirements (see table 5). From the findings of the extra measurements for Q-dummies, it is indicated that these measurements are able to distinguish between the performance of CRSs of one particular ECE-R44 group. This indication can be considered as the added value of the Q-dummy family for child seat evaluation.

From the results of the assessment of Q-dummies and ECE-R44 injury criteria in frontal impact as presented in this paper, the following conclusions are made:

- Head, neck, chest and abdomen need priority in protection (focus depends on age).
- Q0, Q1, Q1.5, Q3 and Q6 are available.
- ECE-R44 mass groups are covered as soon as Q10 is available (expected in 2006).
- Biofidelity targets, based on scaled criteria, are derived for the Q-dummies.
- Q-biofidelity results are good, except for the (linear scaled) thorax requirement.
- Q-measurements show good repeatability.
- Q-dummies are durable for ECE-R44 and EuroNCAP test conditions.
- P- and Q-dummies show similar results with respect to ECE-R44 requirements.
- For CRS evaluation, potential merits of Q-dummy family lie in the extra measurement capabilities.

RECOMMENDATIONS

Using the Q-dummy family for the assessment of all available ECE-R44 CRS groups, the Q-series need to be extended with a dummy, representing a 10 years old child. As mentioned in the conclusions the potential merits of the Q-dummy family for child seat evaluation lie in the extra measurement possibilities. Therefore it is recommended to further investigate new injury criteria. Subsequently, these criteria will be assessed with the Q-dummy test results from the validation program as presented in this paper. In near future, the analysis of the validation program will be finalized. Then, a recommendation for the implementation of the Q-dummy family in ECE-R44 can be made.

The child dummy assessment as described in this paper focuses only on ECE-R44 frontal impact loading. It is recommended to assess a similar program with child dummies for side impact, because side impact legislation is expected in the near future.

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ANNEX A: Anthropometric data of Q-dummies vs. ECE-R44

Body part	**Q0 [kg]**	**“new-born” [kg]**
Head & neck	1.1	0.7
Torso	1.5	1.1
Arms	0.25	0.5
Legs	0.55	1.1
Total mass	3.4	3.4

Dimension	**Q0 [mm]**	**“new-born” [mm]**
Chest depth	90	100
Shoulder width (maximum)	141	150
Hip width seating	98	105
Seating height	354	345

Body part	**Q1 [kg]**	**“9-months” [kg]**	**Q1.5 [kg]**	**“18 months” [kg]**
Head & neck	2.41	2.2	2.8	2.73
Torso	4.46	3.4	5.04	5.06
Upper arms	0.45	0.7	0.58	0.54
Lower arms	0.44	0.45	0.62	0.5
Upper legs	1.00	1.4	1.14	1.22
Lower legs	0.82	0.85	0.92	0.96
Total mass	9.6	9	11.1	11.01

Dimension	**Q1 [mm]**	**“9-months” [mm]**	**Q1.5 [mm]**	**“18 months” [mm]**
Back of buttocks to front knee	211	195	235	239
Back of buttocks to popliteus, sitting	161	145	185	201
Chest depth*	117	102		113
Shoulder width (maximum)	227	216	227	224
Hip width seating	191	166	194	174
Seating height	479	450	499	495
Shoulder height (sitting)	298	280	309	305
Stature	740	708	800	820

*Chest depth is measured on the centre line of the fixation point for the displacement sensor.

Body part	**Q3 [kg]**	**“3-years” [kg]**	**Q6 [kg]**	**“6 years” [kg]**
Head & neck	3.17	2.7	3.94	3.45
Torso	6.40	5.8	9.62	8.45
Upper arms	0.75	1.1	1.27	1.85
Lower arms	0.73	0.7	1.22	1.15
Upper legs	2.00	3	3.98	4.1
Lower legs	1.54	1.7	2.92	3
Total mass	14.6	15	22.9	22

Dimension	**Q3 [mm]**	**“3-years” [mm]**	**Q6 [mm]**	**“6-years” [mm]**
Back of buttocks to front knee	305	334	366	378
Back of buttocks to popliteus, sitting	253	262	299	312
Chest depth*	145.5	125	141	135
Shoulder width (maximum)	259	249	305	295
Hip width seating	200	206	223	229
Seating height	544	560	601	636
Shoulder height (sitting)	329	335	362	403
Stature	985	980	1143	1166

*Chest depth is measured on the centre line of the fixation point for the displacement sensor.

ANNEX B: Test matrix of P & Q-dummy family comparison

TYPE OF CRS	**CRS CODE**	**P0 Q0**	**P 3/4 Q1**	**P 1,5 Q1,5**	**P3 Q3**	**P6 Q6**	**Nb tests**
G0+ RWD FC							
Infant carrier Universal	"01"	X	X	X			12
	"02"	X	X	X			12
	"03"	X	X	X			12
Infant carrier Isofix basis	"04"	X	X	X			12
Combination CRS used RWD	"05"	X		X			8
Combination CRS-RWD isofix	"06"	X	X	X			12
GI FWD & RWD HARNESS							
FWD FC Universal	"07"		X	X	X		12
	"08"		X		X		8
	"09"		X	X	X		12
	"24"		X		X		8
	"11"		X		X		8
	"12"		X		X		8
FWD FC isofix + top tether	"13"		X	X	X		12
FWD FC isofix + support leg	"14"		X		X		8
	"15"		X	X	X		12
RWD FC classical (non-isofix)	"16"		X		X		8
RWD FC isofix	"17"		X	X	X		12
GI FWD SHIELD							
FWD FC isofix	"19"		X	X	X		12
BOOSTER+BACK							
Universal	"20"				X	X	8
	"21"				X	X	8
	"22"				X	X	8
	"23"				X	X	8
MULTI I/II/III same config							
Universal	"10"		X		X	X	12
	"25"		X	X	X	X	16
	"26"		X		X	X	12
MULTI I/II/III differ config							
Universal - shield	"27"			X	X		8
FWD Universal - harness	"29"		X		X		8
	"30"				X	X	8
	"31"		X		X		8
	"32"				X	X	8

CHILD POSES IN CHILD RESTRAINT SYSTEMS RELATED TO INJURY POTENTIAL: INVESTIGATIONS BY VIRTUAL TESTING

Lex van Rooij, Christelle Harkema, Ronald de Lange, Kate de Jager, Marian Bosch-Rekveldt, Herman Mooi
TNO Science and Industry
The Netherlands
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ABSTRACT

In Europe approximately 1250 children younger than 15 years of age die in traffic each year. The number of children severely injured in traffic is dramatically higher. Within the ECE-R44 regulation the safety of children in cars has been regulated by means of certification of child restraint systems (CRS). Much has been achieved, but further reduction of injuries seems possible. The ECE-R44 regulation provides a simplified set up and test configuration, that may be different from the real-world environment in which a child is injured.

In this study, a virtual testing approach was followed to explore the effect of one particular aspect, i.e. the posture of a child in a CRS, on the injury potential in a typical car crash. The investigation focussed on the vulnerable child population seated in ECE-R44 Group I seats. A photo-study was performed with 10 children in the age group from one to three years. Their positions were recorded on short and longer drives. Few children remained seated in the standard position. Most children slouched, slanted and turned their head and rested it on the side-support of the CRS. Extreme positions such as leaning forward, escaping from the harness or holding feet were observed. In the MADYMO simulation environment a non-deforming finite element model of a CRS was combined with multi-body models of Q1.5 and Q3 dummies and of human child models representing 1.5 and 3-year-olds. They were set up in realistic poses. The dummy models were adapted to enable these poses, while the human models were used to compare the biofidelity performance. From the simulated response between the ECE-R44 prescribed position and various common and extreme positions children were found to be in, it was shown that children are at an increased risk in relatively common positions. High lateral neck loads were observed in slanted positions, while correctly restrained children that managed to escape from their shoulder harness sustained large amounts of head excursion. Virtual testing was shown to be a valuable tool to predict trends in situations that are more closely related to the actual automotive environment than current regulations or hardware testing do.

INTRODUCTION

In the 15 countries that comprised the European Union until 2005, approximately 1250 traffic fatalities were recorded among children up to 15 years of age in the year 2002 [EU, 2005]. Approximately half of these fatalities were from child occupants, the rest from pedestrian or cyclists. Although the number of fatalities is relatively small compared to adult fatalities, the number of injuries children sustain is dramatically larger.

Serious or fatal injuries in child occupants have various causes. The use of an appropriate child restraint system (CRS) is a key requirement for protecting a child. A CRS prevents the child to impact vehicle interior structures and it ensures a belt restraint condition that is designed specifically for the smaller anthropometry of a child. However, the CRS needs to be installed properly, which often is difficult to do and hence causes potentially dangerous situations [Quintero del Rio, 1997]. In addition, child restraint systems are designed for a specific range of body weight or length. When a child is seated in a CRS that is inappropriate for its weight or length, potentially hazardous restraint conditions may exist. Parents are often prone to prematurely graduate their child to a larger seat, which causes an inappropriate belt fit. The latter may result in submarining, the lap belt cutting into the abdomen while the child's pelvis slides underneath.

Even when the proper CRS is installed in the vehicle and the child is positioned correctly with no slack in any of the belts, a serious injury risk may exist. Current CRS designs allow children a certain amount of freedom to move around in their seats. Meissner et al. showed that children seated in booster seats have a large tendency to move with respect to their CRS and to move the belt restraint around [Meissner, 1994]. Whether any posture other than the standard posture a child is in when positioned in the CRS has an effect on the injury risk is unknown.

The currently existing test method for evaluating the performance of child restraint systems in the EU is ECE-R44 [ECE, 1998]. Within this regulation a frontal impact sled test is performed at an impact speed of 50 km/h. A child dummy that is appropriate for the tested seat is installed properly, while all belt restraints are pretensioned. Since no vehicle interior is implemented, no information is provided on a possible interaction of the child with the vehicle. Euro-NCAP installs child restraints on the rear seat of vehicles in their full-scale crash tests in order to evaluate the child safety performance of the vehicle in both frontal as well as side impact. However, the vehicle manufacturer is free to choose any CRS and will therefore always choose the seat with the best performance rating, thus eliminating the gross of seats on the market. Various consumer testing programs are being developed in order to ensure safer CRS designs that are easier to install and will reduce injury risk of child occupants.

Currently, within both ECE-R44 and Euro-NCAP testing child dummies of the TNO P-series are used. While these dummy designs have been successful in terms of reproducability and durability, the biofidelity of their response is limited. A new series of child dummies, the Q-family, was designed in order to overcome the lack of biofidelity. Currently, the dummy performance is being tested in a research environment [de Jager, 2005].

Virtual testing, or numerical simulation, is a useful method for extrapolating beyond currently existing test methods and dummies. While current experimental test methods are limited to hardware dummies and a limited amount of test conditions, parametric simulation studies are virtually unlimited in size and amount of parameters. Simulations are only valid within the range they are validated for, but extending outside the range of validation might be useful in showing possible trends.

The objective of this study was to investigate the effect of various poses on the injury response of children in child restraints. In a virtual testing environment first of all the model setup needed to be created. Human surrogate models of two anthropometries were developed; 1.5 and 3-year-old. Dummy models of the Q-family were developed and validated against component tests. Human child models were generated in order to compare the biofidelity response of the models. In order to find out which poses were common and which poses were extreme a photo study was performed. Common poses and some extreme poses were simulated in a crash environment model with dummy and human geometry in order to indicate a potential increase of injury risk in poses different than the standard one.

METHODS

In the methods section, first the development and validation of the modeling environment needed for the posture study will be discussed. Secondly, the posture study itself will be discussed, subdivided into photo study and simulation study.

Q3 dummy model development

A multi-body model of the latest version of the Q3 hardware dummy [de Jager, 2005] was created based on a pre-existing Q3 ellipsoid dummy model [MADYMO, 2004]. The model consisted of 32 rigid bodies that were interconnected by 32 kinematic joints. Mass and inertia properties were attributed to the rigid bodies, while force models were implemented in the joints. The outer geometry of the model was represented by 40 ellipsoids and 4 cylinders for which contact characteristics were defined. The resulting model is shown in figure 1.

In order to compare properties of the developed Q3 dummy model with the actual hardware dummy, the segment mass distributions between various body parts is shown in table 1 for the production dummy specifications as well as for the dummy model. The total mass of the dummy was approximately 14.5 kg.

Table 1: Segment mass distribution of Q3 dummy and dummy model.

Segment mass [g]	Q3 product specs.	Q3 model
Head	2784	2784
Neck	382	381
Torso Upper	1976	2047
Torso Lower	4032	4245
Arms Upper	750	760
Arms Lower	728	740
Legs Upper	2000	1980
Legs Lower	1542	1540
Suit	390	0
Total	14584	14477

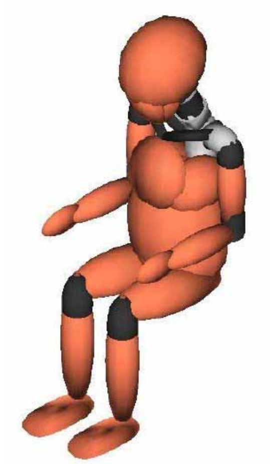

Figure 1. The MADYMO Q3 dummy model.

The Q3 dummy model was developed to represent the Q3 hardware dummy, allowing for validation against component test data of the current hardware dummy.

Q1.5 dummy model development

A Q1.5 dummy model was developed to represent the Q1.5 hardware dummy. This dummy can be applied in both Group 0+ as well as Group I child restraint evaluation and is therefore seen as an important dummy in the Q-family. Geometrical data of the Q1.5 dummy was obtained from a CT scan database that was processed with visualization package Mimics [Mimics, 2005]. From this database external anthropometric dimensions were computed as well as internal geometrical landmarks such as joint locations. A cross-section scan of the Q1.5 dummy in the frontal plane is shown in figure 2.

The Q1.5 dummy model was developed by anthropometrical scaling of the Q3 dummy model, followed by manual corrections to obtain correct segment mass distributions. The MADYMO/Scaler module scaled based on a set of 35 anthropometric parameters, such as seated height and hip breadth [MADYMO, 2004]. This set was defined for the Q3 dummy model based on the dimensions of the model. For the Q1.5 dummy model, the set of parameters was derived from the CT database. From these two anthropometric datasets, scaling parameters were determined for 14 body regions in three directions. The Q1.5 dummy model was created by applying scaling rules, with the obtained scaling parameters, to dimensions, mass and inertia, stiffness and damping parameters. The resulting dummy model resembled the Q1.5 model in terms of anthropometry and internal joint locations. The resulting dummy model is shown in figure 2.

In order to make the model comply with the segment mass distributions as specified for the production dummy, the mass and inertia parameters of the rigid bodies were altered. A comparison of segment mass distribution between actual dummy and developed model is provided in table 2.

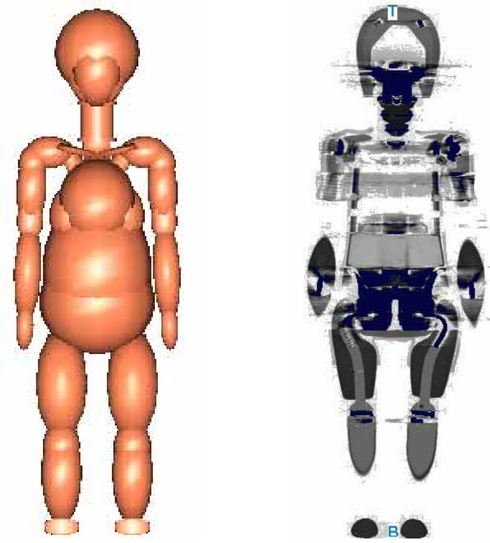

Figure 2. The MADYMO Q1.5 dummy model and CT scan of actual dummy.

The mass of the actual Q1.5 dummy neck did not differ from the mass of the Q3 dummy neck. An identical neck design was used for both dummies. Accordingly, scaling rules in the neck assembly of the model were suppressed in order to leave the neck of the model unchanged.

Table 2: Segment mass distribution of Q1.5 dummy and dummy model.

Segment mass [g]	Q1.5 product specs.	Q1.5 model
Head	2400	2400
Neck	382	381
Torso Upper	1324	1336
Torso Lower	3408	3658
Arms Upper	575	556
Arms Lower	620	625
Legs Upper	1140	1152
Legs Lower	922	899
Suit	305	0
Total	11076	11007

The Q1.5 dummy model, as developed by scaling from the Q3 dummy model, resembled the actual Q1.5 dummy as far as anthropometry, internal dimensions and mass distributions goes. The stiffness values of soft tissue contact models and kinematic joint stiffness were scaled from the Q3 model. Component validation was needed to show validity of the applied scaling technique.

Dummy model component validation

Validation of the dummy models was performed against component tests as specified in the dummy design requirements [FTSS, 2003]. No full scale sled test data was available, hence no validation on the whole body dummy was performed.

Q3 dummy model – An overview of validation simulations performed for the Q3 dummy model is shown in table 3. For all test conditions corridors or peak response and timing of peak were available, as defined for assessing dummy biofidelity. For frontal and lateral head drop tests, for frontal thorax impactor tests and for lumbar flexion tests hardware dummy test data was available, in addition to corridor requirements.

Table 3: Overview of validation performed on Q3 dummy model.

Test description	Specifications	
Head		
Frontal drop*	Drop height	130 mm
Frontal drop*	Drop height	376 mm
Lateral drop*	Drop height	130 mm
Lateral drop*	Drop height	200 mm
Neck		
Pendulum extension	Impact velocity	3.9 m/s
Pendulum flexion	Impact velocity	3.9 m/s
Pendulum lateral	Impact velocity	3.5 m/s
Thorax		
Impactor frontal*	Impact velocity Impactor mass	4.3 m/s 3.8 kg
Impactor frontal*	Impact velocity Impactor mass	6.7 m/s 3.8 kg
Impactor lateral	Impact velocity Impactor mass	4.3 m/s 3.8 kg
Impactor lateral	Impact velocity Impactor mass	6.7 m/s 3.8 kg
Lumbar		
Pendulum frontal*	Impact velocity	4.4 m/s

* hardware dummy test data available

The frontal head drop test simulated a facial impact and was performed at two heights to evaluate the rate-dependency of the foam of the dummy head. The lateral head drop test simulated an impact of the dummy head under an angle with a side structure of a vehicle. The impacted plate was rigid. The acceleration of the head center of gravity was computed. Simulation setups are shown in figure 3.

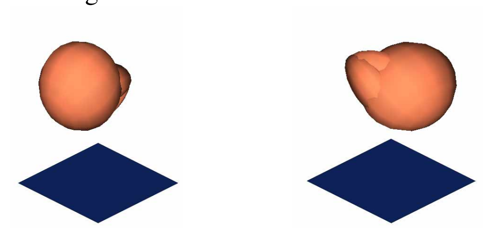

Figure 3. Simulation setup of Q3 head drop test in frontal (left) and lateral (right) direction.

The neck pendulum test was designed to evaluate the performance of the neck in three bending directions: flexion, extension and lateral flexion. The neck was disassembled from the dummy and mounted to a pendulum on the proximal side, while it was mounted to a standardized test mass representing a dummy head on the distal side, as shown in figure 4. The pendulum was stopped by a 3 inch layer of honeycomb, which was modeled in a contact characteristic with a crush force of 2500 N and 75% allowable compression. The velocity decrease of the pendulum and the total head rotation were computed.

Figure 4. Simulation setup of Q3 dummy neck pendulum test just before impact (left) and at time of maximum neck bending (right)

The thorax impactor test was designed to evaluate the thoracic response to impact in frontal and lateral direction. The free-flying impactor with a mass of 3.8 kg struck the sternum in frontal and the ribs in lateral impact at speeds of 4.3 m/s and 7.6 m/s. The thoracic response was characterized by a force-deflection plot for frontal impact and a force history plot for lateral impact. The simulation setups for thorax impactor tests are shown in figure 5.

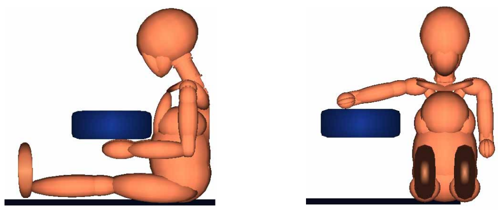

Figure 5. Simulation setup of Q3 dummy thorax impactor test in frontal (left) and lateral (right) direction

The lumbar pendulum test setup was similar to the neck pendulum test, where the neck assembly was replaced by the lumbar spine assembly. The pendulum impact speed was increased from 3.9 m/s to 4.4 m/s. For an image of the test setup it is referred to the setup for the neck as shown in figure 4.

Q1.5 dummy model – To evaluate the Q1.5 dummy model response, simulations similar to those performed for the Q3 dummy model validation were setup. A smaller amount of tests was available for the Q1.5 dummy, as shown in table 4. Since the neck of the Q1.5 dummy and corresponding dummy model were identical to the neck of the Q3 dummy and corresponding model, no validation of the Q1.5 neck was performed. The mass of the thorax impactor was reduced from 3.8 kg to 2.6 kg, as prescribed in the dummy requirements [FTSS, 2003]. For head frontal drop, thorax frontal and lumbar flexion tests, hardware dummy test data was available in addition to dummy design requirements. For a more detailed description of the test setup it is referred to the Q3 dummy component validation paragraph.

Table 4: Overview of validation performed on Q1.5 dummy model.

Test description	Specifications	
Head		
Frontal drop*	Drop height	130 mm
Thorax		
Impactor frontal*	Impact velocity Impactor mass	4.3 m/s 2.6 kg
Impactor lateral	Impact velocity Impactor mass	4.3 m/s 2.6 kg
Impactor lateral	Impact velocity Impactor mass	6.7 m/s 2.6 kg
Lumbar		
Pendulum frontal*	Impact velocity	4.4 m/s

* hardware dummy test data available

Human child model development

Human child models of a 1.5 and a 3-year-old child were developed in order to be able to compare responses between the two human surrogate models. The models were developed by a scaling technique similar to the technique used for the development of the Q1.5 dummy model. The model was scaled from the MADYMO human occupant model in a 50th percentile male configuration [Happee, 2001]. Different from the ellipsoid dummy models, these models are characterized by a mesh representing the skin, by flexible bodies representing a fully deformable thorax and abdomen and by additional joint models representing all spinal flexibility.

The 35 anthropometric scaling parameters of the 50th percentile adult were derived from RAMSIS anthropometric database [Seidl, 1994]. The anthropometry datasets for 1.5-year-old and 3-year-old were based on the design specifications of the Q3 and Q1.5 dummies, which were derived from the CANDAT child anthropometry database [Twisk, 1993]. Less important anthropometric parameters that were not available within CANDAT were taken from the GEBOD database [MADYMO, 2004]. These less important parameters were altered to improve the segment mass distribution resulting from the scaling routine. An overview of CANDAT mass specifications and resulting model segment mass distribution is given in table 5. The scaling routine optimization target included total dummy mass, which was reached. The models resulting from the scaling procedure are shown in figure 6.

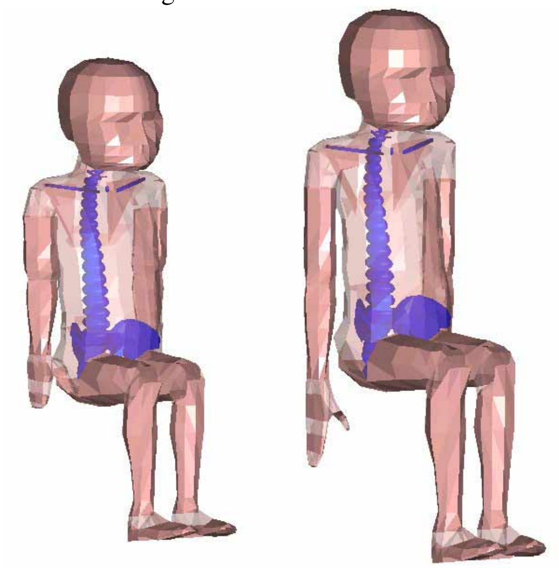

Figure 6. The MADYMO 1.5-year-old (left) and 3-year-old human child models.

Table 5: Segment mass distribution as specified in CANDAT database and as resulting from human models of 3-year-old and 1.5-year-old children.

Segment mass [g]	3 yo model	3 yo specs.	1.5 yo model	1.5 yo specs.
Head	3190	3220	2510	2540
Neck	330	300	300	300
Torso	6220	6410	5120	5100
Arm upper	370	370	280	270
Arm lower	210	210	150	150
Hand	100	130	080	100
Thigh	950	980	580	570
Leg lower	500	500	300	290
Foot	250	240	170	160
Total	14500	14500	11020	11020

The developed human child models represented children of approximately 1.5 years and 3 years of age. While the anthropometry was defined from anthropometric databases, the stiffness and damping parameters were scaled based on geometrical scaling on body part level and did not take structural changes and differences in material properties between children and adults into account.

Group I CRS simulation

All developed child models, Q1.5 and Q3, as well as human 1.5-year-old and 3-year-old, can be positioned in a Group I child restraint system (CRS). A Group I seat model positioned in a typical car seat was developed to evaluate the behavior of the developed child models in a typical vehicle crash environment.

A mesh of the outer geometry of a production CRS was generated and implemented in MADYMO. The finite element CRS was considered undeformable, while seat compliance was modeled by means of a contact characteristic between child model and CRS. The CRS was positioned in two planes with a contact characteristic representing a rear seat of a typical passenger car. The seat was mounted to the vehicle with the vehicle's three-point belt system, modeled by a multi-body belt. The child model, either human or dummy, was positioned on the CRS after which FE belts were wrapped around the child that represented the internal 5-point harness of the CRS. An image of a Q3 dummy model in the modeled CRS on the vehicle rear seat is shown in figure 7.

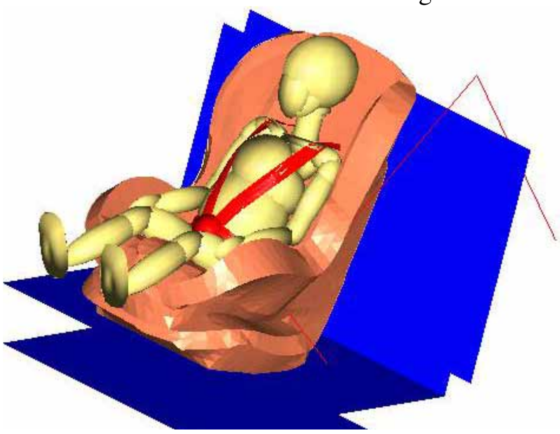

Figure 7. Generic child restraint model of Group I seat, mounted on vehicle rear seat with positioned Q3 dummy model and FE internal harness.

A frontal crash was simulated by prescribing an acceleration field to child model and CRS, while the vehicle seat was mounted to reference space. The supplied pulse was taken from ECE-R44 regulations [ECE, 1998]. Since the Q3 dummy is the proposed upper limit dummy for Group I CRS evaluation, simulations have been performed with the Q3 dummy model and with the 3-year-old human model.

No experimental validation data was available to validate the CRS model or to evaluate the response of child dummy or human model in a full scale crash environment.

Injury Reference Values for 3-year-old children

Injury Reference Values (IRV) for a 3 year-old child are shown in table 6. IRV's indicate a reference value at which injury may occur. Some of these values have been defined in regulations [ECE, 1998], others are proposed values for regulations [Eppinger, 2000], some were scaled by body mass ratio from adult cadaveric data [Mertz and Patrick 1971, Cavanaugh 2002], while others were adapted from advanced scaled data [Ivarsson 2004, 2005]. The values presented here were meant to be mere indications of injury severity. They were used for ease of normalizing various responses, not to predict the occurrence of injury per se.

In order to present a normalized injury indicator, the relative Injury Reference Value (rIRV) was defined as follows:

$$rIRV = \frac{response}{IRV} \qquad (1)$$

Table 6: Injury Reference Values (IRV) of 3-year-old children with source and subsequent comment.

Criterion	IRV	Source
Head Injury Crit. **HIC**	570	*Eppinger 2000*
Head Excursion	0.55 m	*ECE-R44*
Neck Injury Predict., tension-extension **Nij TE**	1	*Eppinger 2000*
Neck Injury Predict., compression-ext. **Nij CE**	1	*Eppinger 2000*
Neck Lower Force, lateral shear **Fy**	273 N	*Mertz and Patrick 1971*, mass scaled
Neck Lower Force, axial **Fz**	-1380 N	*Eppinger 2000*
Neck Lower Moment, lateral bending **Mx**	16.5 Nm	*Ivarsson 2005*, Nij intercept
Sternum Displacement	0.034 m	*Eppinger 2000*
Thorax Viscous Criterion **VCmax**	0.74	*Cavanaugh 2002*, mass scaled

Hence, the model response divided by the in table 6 defined IRV is a normalized measure for injury potential and was used as such in presenting results. When rIRV equals 1, the computed response is equal to the IRV defined.

Posture study

A posture study was performed since it was hypothesized that children rarely sit in the same posture for a long period. At first, a real-world photo study was performed to indicate which posture children regularly take on longer drives. The resulting most common or most extreme postures were then simulated with a Q3 dummy model and a 3-year-old human model in a Group I seat at ECE-R44 impact level.

Real-world photo study – To investigate what postures children seated in Group I seats take on long drives, parents of a total number of 10 children were asked to take pictures of their children and to fill in a questionnaire. Three series of photos were taken.

- **A**: one picture of the child in the CRS every 15 minutes during drives of at least one hour, taken by the co-driver.
- **B**: one picture of the child just after it was positioned in the CRS and one right after the drive, so that the picture could be taken from outside the vehicle to obtain a better view.
- **C**: pictures of the child when it was observed that it took a strange or extreme position.

The resulting photographs were organized into two categories. First of all, in order to find common postures that often occurred and that many children took. Secondly, in order to find some extreme postures that children take, which are potentially dangerous in case of a crash.

Modeling of poses – To evaluate the effect of various poses or postures, some of the observed postures were simulated in the CRS on the vehicle rear seat model environment that was developed within this study. By changing the positions and orientations of the kinematic joints of dummy and human model the postures observed in the photo study could be modeled. With the changed position also the belt routing changed, which made rewrapping of the FE internal CRS harness necessary.

The Q3 dummy model joint characteristics needed to be changed in order to be able to position the dummy in many of the postures. Most dummy joints provide resistance to any position other than the reference position, which in the modeling environment with the original dummy model resulted in a transient effect at the start of the simulations, where the joint relaxed into its reference state. In order to eliminate this undesired transient effect, the dummy joint characteristics were altered so that no force or torque was generated at the desired joint orientation. This increase of range of motion of dummy joints, reduced the quality of validation of the model. However, the effect on dummy response was found to be small and acceptable for performing trend studies.

RESULTS

The results from this study consist of the simulation responses from the dummy component tests and the response of the generic CRS model simulations, followed by results from the photo study and the numerical analysis on the various postures.

Dummy model component validation

The component validation results of both Q3 and Q1.5 dummy models will be presented in this section. Additional figures can be found in appendix 1 and 2 for Q3 and Q1.5 respectively.

Q3 head – The validation of the Q3 dummy head in frontal drop test at 130 mm was compared with three hardware dummy tests, as shown in figure 8. The x-acceleration of the head was higher than observed in any of the three experiments, which indicated that the head contact stiffness was too high in the model. However, the maximum resultant acceleration was 1254 m/s^2, which fulfilled the requirement that the acceleration should lie in between 981 and 1275 m/s^2. At a drop height of 376 mm the model response was slightly lower than the hardware dummy responses. The lateral drop test at 130 mm drop height showed that the model fulfilled the requirement of minimum 1177 m/s^2 and maximum 1472 m/s^2 at a resultant head acceleration of 1257 m/s^2. At 200 mm drop height, the model response was comparable to the test responses.

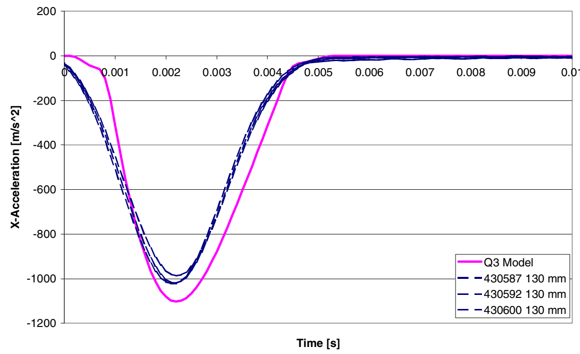

Figure 8. X-acceleration of Q3 head frontal drop test validation at 130 mm height.

Q3 neck – Validation of the neck of the model in flexion resulted in a plot of pendulum velocity decrease, as shown in figure 9, and in total head rotation, as shown in figure 10. The velocity decrease plot showed that the pendulum speed decreased slightly more rapid than allowed by the corridor. The total head rotation was slightly higher than allowed by the dummy requirements and the angular rate of the neck was slightly lower, as shown by the peak occurring later than allowed by the bounding box.

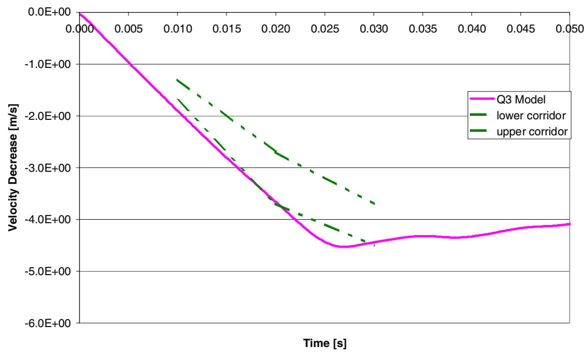

Figure 9. Velocity decrease of Q3 neck flexion pendulum test validation at 3.9 m/s impact speed.

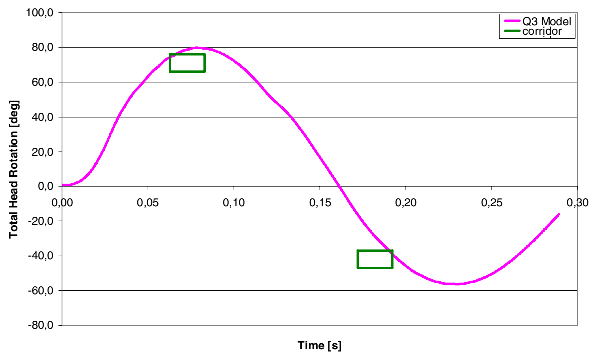

Figure 10. Total head rotation of Q3 neck flexion pendulum test validation at 3.9 m/s impact speed.

The results of the neck pendulum tests in the other loading directions, extension and lateral flexion, showed similar results. The velocity decrease was on the high side of the corridor, while the maximum allowable head rotation was slightly larger and occurred slightly later than the maximum set in the requirements.

<u>**Q3 thorax**</u> – The thoracic response of the Q3 model was evaluated with the force-deflection response resulting from the frontal thoracic impactor test, as shown in figure 11. The hardware dummy showed a stiffer performance than the corridor, on which it is elaborated by de Jager et al. [de Jager, 2005]. The model response resulted in an even higher impact force and a maximum deflection close to 30 mm. A similar trend was observed at higher impact speed of 6.7 m/s. The odd shaped fluctuation of the model response was attributed to a vibration in the impactor force history, which indicated a lack of damping in the dummy model.

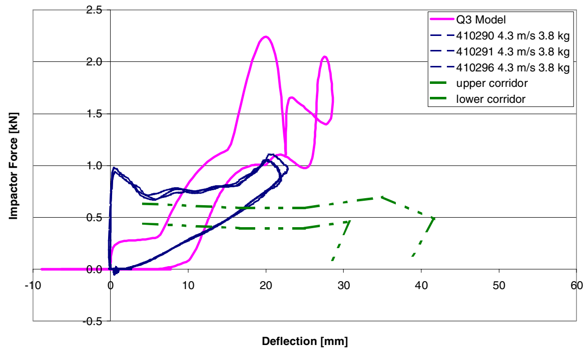

Figure 11. Force-deflection of Q3 thorax frontal impactor test at 4.3 m/s with 3.8 kg mass.

The lateral thoracic force response was approximately four times higher than defined in the corridor, while timing was fairly correct, as figure 12 indicates.

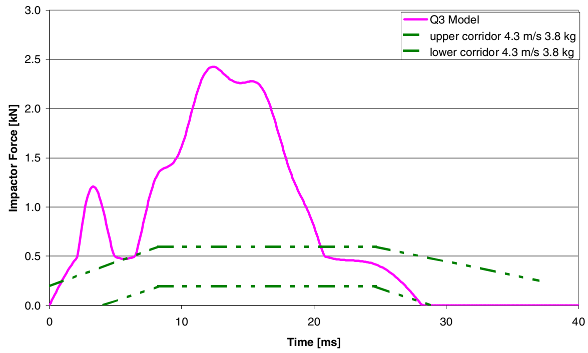

Figure 12. Force history of Q3 thorax lateral impactor test at 4.3 m/s with 3.8 kg mass.

<u>**Q3 lumbar**</u> – The lumbar pendulum test results are plotted in figures 13 and 14. The model response was very similar to the three hardware dummy test responses, but the bounding boxes for maximum total head rotation and timing of maximum total head rotation were not completely met.

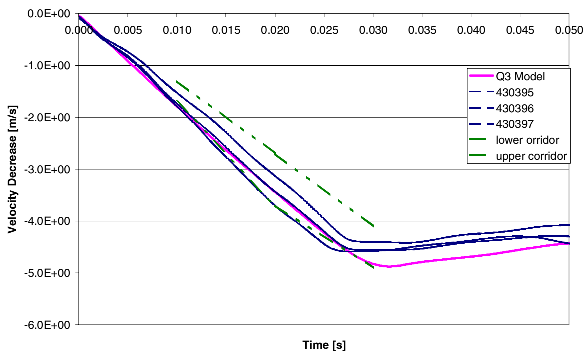

Figure 13. Velocity decrease of Q3 lumbar flexion pendulum test validation at 4.4 m/s impact speed.

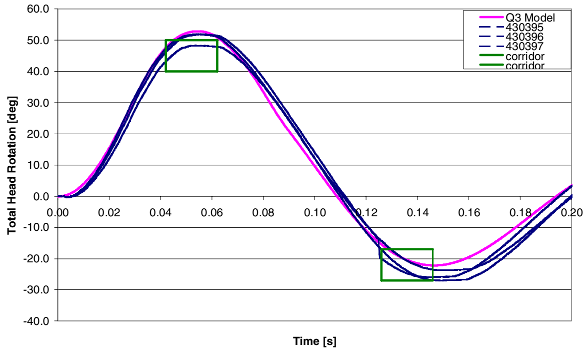

Figure 14. Total head rotation of Q3 lumbar flexion pendulum test validation at 4.4 m/s impact speed.

From the Q3 dummy model component validation it was concluded that the model response did not always meet the requirements. However, most of the responses fell only slightly outside the corridor, except for thoracic force response that was approximately four times higher than required.

Q1.5 head – The x-acceleration response of the head frontal drop test at 130 mm for the Q1.5 dummy model is shown in figure 15. The model response was about 15 % higher than the hardware dummy response. Also, the timing of the peak acceleration occurred slightly earlier. The resultant head acceleration fell inside the requirement of 1089±284 m/s^2 at 1299 m/s^2.

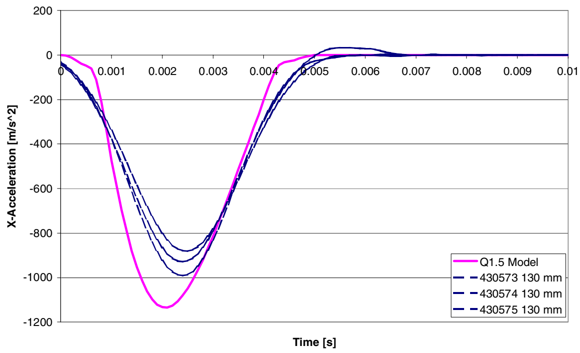

Figure 15. X-acceleration of Q1.5 head frontal drop test validation at 130 mm height.

Q1.5 thorax – The frontal thoracic impactor response of the 4.3 m/s test is plotted in figure 16. The hardware dummy performance is slightly different from the corridor, on which it is referred to de Jager et al. [de Jager, 2005]. The force response of the model was about four times higher than the corridor, while the maximum deflection did not increase up to the range set in the corridor. Lateral thoracic tests resulted in a similar trend, where the force was approximately four times higher for both high and low impact speed tests.

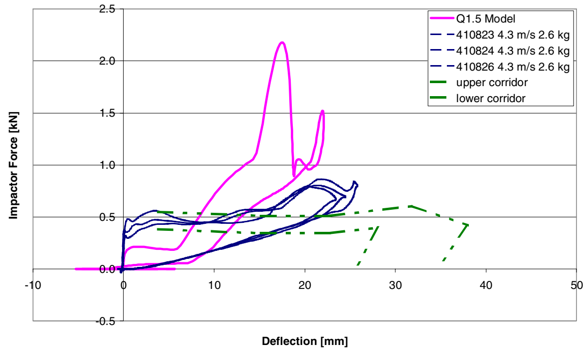

Figure 16. Force-deflection of Q1.5 thorax frontal impactor test at 4.3 m/s with 2.6 kg mass.

Q1.5 lumbar – The lumbar pendulum test results are shown in figures 17 and 18. While the pendulum velocity decrease of the model fell inside the corridor during the larger part of the simulation, the maximum total head rotation was larger than allowed by the requirements. Nevertheless, the timing was correct and the model response was very similar to the hardware dummy response.

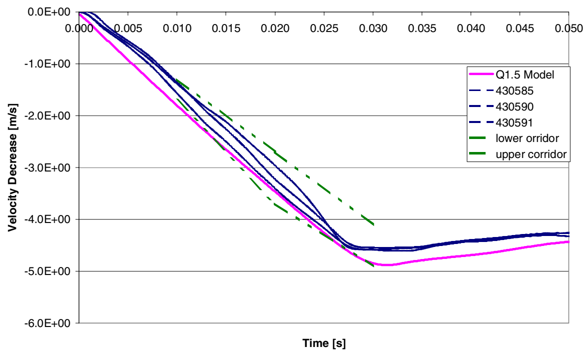

Figure 17. Velocity decrease of Q1.5 lumbar flexion pendulum test validation at 4.4 m/s impact speed.

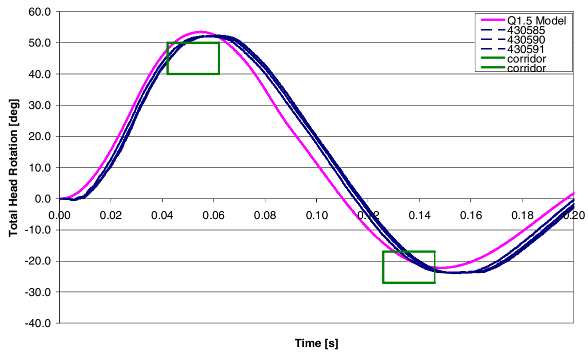

Figure 18. Total head rotation of Q1.5 lumbar flexion pendulum test validation at 4.4 m/s impact speed.

From the Q1.5 dummy component validation simulations it was concluded that similar trends were observed as in Q3 component validation. The model response met the requirements or just did not meet the requirements for most all responses, except for thoracic impactor force response.

Group I CRS simulation

Kinematic simulation results of the Q3 dummy model and 3-year-old human child model seated in a Group I CRS at ECE-R44 impact are shown in figure 21. At first, the CRS and child model moved forward. After 50 ms the lower neck went into flexion, while the upper neck went into extension, caused by the unrestrained head moving relative to the restrained torso. At 100 ms after impact a difference between dummy and human was observed. The dummy allowed more forward movement of the shoulder than the human model. The human model's shoulder was restrained more. Therefore, later in the event, the human torso stayed more upright than the dummy torso. Even though the belt friction coefficient was identical at 0.4, differences in the contact algorithms between ellipsoid and rigid FE models might have caused this. The kinematics of head and neck was fairly similar throughout the rest of the event.

A comparison of model response in terms of resultant head acceleration between Q3 dummy model and human model, as shown in figure 19, indicated that both timing and maximum value of head acceleration were fairly similar between human and dummy model.

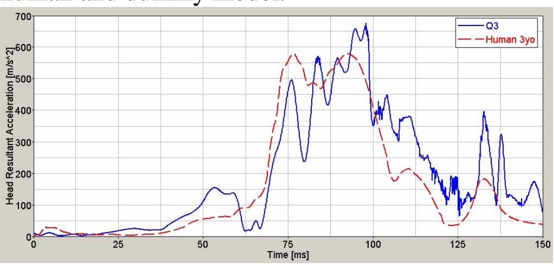

Figure 19. Head resultant acceleration of Q3 dummy model and human 3-year-old model in Group I seat with ECE-R44 pulse.

In terms of force response, the axial force generated in the lower neck indicated that both dummy and human model predict neck tension at first and neck compression later in the event, as figure 20 shows. The maximum compressive force in the lower neck is similar for both at around -1500 N.

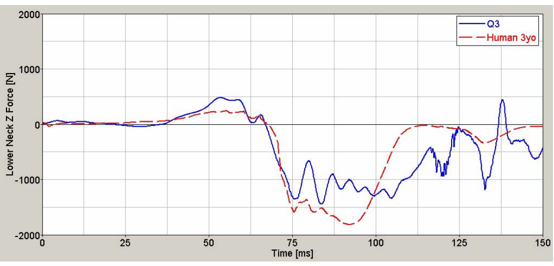

Figure 20. Lower neck axial force of Q3 dummy model and human 3-year-old model in Group I seat with ECE-R44 pulse.

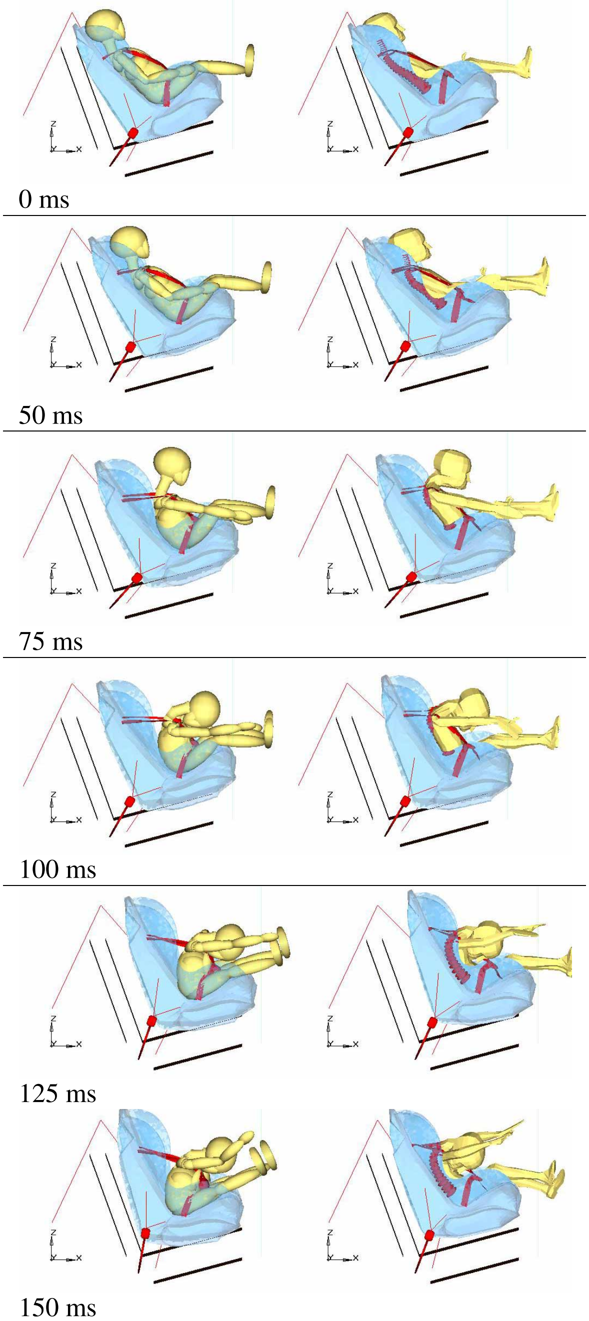

Figure 21. Simulation of Q3 dummy model (left) and human 3-year-old model (right) in Group I seat with ECE-R44 pulse.

In order to quickly compare the effect of a certain impact configuration on the model response, in figure 22 the relative Injury Reference Values (rIRV), as defined in table 6, for the performed simulations are shown. From the Q3 model and human 3-year-old model comparable rIRV values were computed for HIC and head excursion. The values for Nij in the human model simulation were far lower than the Q3 model, due to almost non-existing extension in the human model upper neck. Forces and moments in the lower neck were comparable. Large differences existed for the sternum displacement and VCmax as well. The lower dummy sternum displacement matched with the low sternum displacement observed in thoracic impactor simulations, as shown in figure 11.

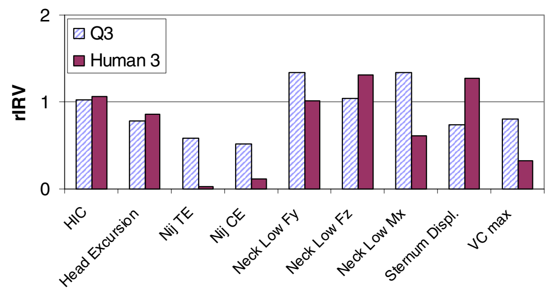

Figure 22. Relative Injury Reference Values (rIRV) of simulations with Q3 dummy model and human 3-year-old model in Group I seat with ECE-R44 pulse.

Real-world photo study on poses

The photo study resulted in a total of 141 photographs. A division was made between children based on age and weight. Four out of ten children were best represented by a 3-year-old model, while the other six were best represented by a 1.5-year-old model. Obviously, a younger child has more freedom of movement in its seat. As a result, a larger variety of postures was found for smaller children.

The standard posture, sitting up straight, was found most often. This posture is shown in figure 23 on the left. An extreme posture is shown in figure 23 on the right. The child in the 1.5-year-old group was slouched and managed to hold her feet in her hands.

Figure 23. Photo of child sitting straight up (left) and of child holding her feet in her hands (right).

Often, children were hanging to either one side of their CRS, resting their heads on the wings of the CRS. This posture is shown in figure 24 on the left. Either the whole body was slanted, or just the neck was laterally flexed. The child's neck was often hanging in the shoulder belt. A posture typical for older children was to stretch one of their legs against the front row seat, as shown in figure 24 on the right. Often this was combined with the other limb pulled up and resting on the knee of the stretched limb. In order to reach the front row seat with their feet, children were often slouched in their CRS.

Figure 24. Photo of child sleeping slanted (left) and of child stretched out, one leg pulled up and right foot against front seat (right).

An uncommon position, but observed with two children, was to escape from the shoulder harness and then lean forward. The parents of these specific children stated clearly that they removed all slack from the internal harness system during installation, but the child still managed to escape. A photo of this position is shown in figure 25.

Figure 25. Photo of child escaped from shoulder belts.

Additionally to these five poses, children were sometimes leaning forward in their shoulder restraint, completely hunched. At other times, children managed to escape from their shoulder harness and rotate their whole body so that they could look backward. Many variations on all poses existed as well.

Modeling of poses

The five poses shown in figures 23-25 were chosen to be modeled in the Group I CRS simulation environment discussed before. Simulations were performed with both the Q3 as well as the 3-year-old human model.

Standard pose – The standard pose, of a child sitting straight up, was considered to be identical to the standard model setup that was discussed earlier. This posture and the accompanying model are referred to as the base posture and model in the following figures.

Child holding feet in her hands – The pose of a child holding her feet in her hands, was modeled by changing the orientation of the joints such that all extremities were stretched and the hands were in the proximity of the feet. The child model was positioned somewhat slouched in its seat. The initial setup is shown in figure 26. While in reality this pose was observed at a child representing the 1.5-year-old age group, simulations were performed with the 3-year-old child models.

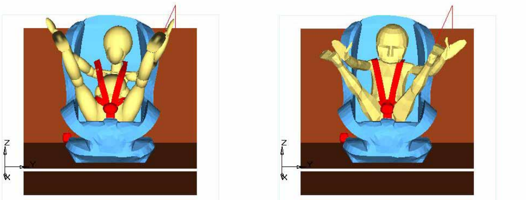

Figure 26. Simulation setup of Q3 dummy model (left) and human 3-year-old model (right) holding feet in hands.

The resulting kinematic response consisted of upper and lower limbs flinging in the direction of the force vector, e.g. frontal. This behavior was observed in the standard posture as well, only with a different initial orientation of the limbs. The kinematic response is therefore similar to the standard posture response.

Child sleeping slanted – The child sleeping slanted in the CRS was modeled by rotating the body slightly and by adding lateral flexion in the neck, as shown in figure 27. The left shoulder belt was proximal to the neck, causing an asymmetric load condition.

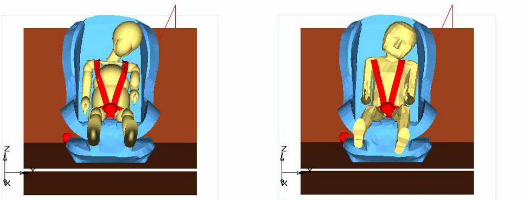

Figure 27. Simulation setup of Q3 dummy model (left) and human 3-year-old model (right) sleeping slanted.

The kinematic simulation results, shown in figure 28 at 95 ms after impact, showed that the asymmetric load condition resulted in lateral components of head movement.

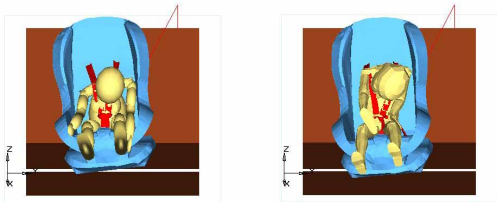

Figure 28. Simulation of Q3 dummy model (left) and human 3-year-old model (right) sleeping slanted, 95 ms after impact.

Child with one leg stretched against front row seat – In order to simulate the effect of a front row seat on the child's lower extremity response, a plane was added with an assumed contact stiffness characteristic for a seat back. Initially, the right lower limb of the child was stretched and in contact with the plane. The left limb was pulled up with the left foot resting on the right knee. This setup is shown in figure 29. Additionally, the child was slanted causing an asymmetric load condition at the shoulder harness.

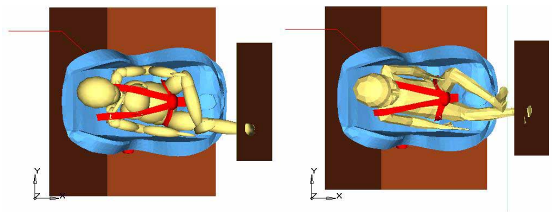

Figure 29. Simulation setup of Q3 dummy model (left) and human 3-year-old model (right) stretched out, one leg pulled up and right foot against front seat.

In figure 30 the simulation results are shown at 85 ms after impact. The simulation results were similar to those of the standard position, except for the lower extremities that did not stretch out but were compressed against the seat, inducing knee flexion. The forces generated in the tibia were at 400 N well below the fracture threshold defined from scaled adult data at 1860 N [Ivarsson, 2005]. However, in the current simulation setup knee flexion was present, while in reality a fully extended knee might induce higher forces in the lower extremity.

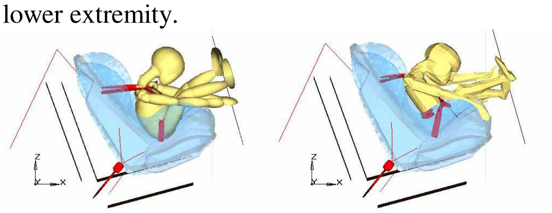

Figure 30. Simulation of Q3 dummy model (left) and human 3-year-old model (right) stretched out, one leg pulled up and right foot against front seat, 85 ms after impact.

Child that escaped from shoulder belts – The child that escaped from the shoulder harness and then leaned forward was modeled as shown in figure 31. The FE internal harness was routed differently, with the shoulder belt going underneath the armpits.

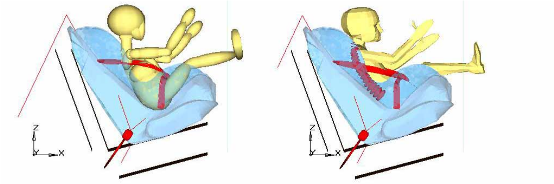

Figure 31. Simulation setup of Q3 dummy model (left) and human 3-year-old model (right) escaped from shoulder belts.

Simulation results of the child that escaped from the shoulder belts are shown in figure 32. The child was correctly restrained by the lap belt of the harness and therefore full body excursion relative to the CRS did not occur. However, the unrestrained upper torso, head and neck moved forward causing large lumbar flexion and the dummy spine being lined up with the force vector from the impact. The head excursion limit was exceeded. In a full-scale setup the head would have impacted the front row seat, possibly implicating severe head injury. However, this was not modeled since the Head Injury Criterion (HIC) is very sensitive to the contact stiffness of the front row seat, for which no validated model was available.

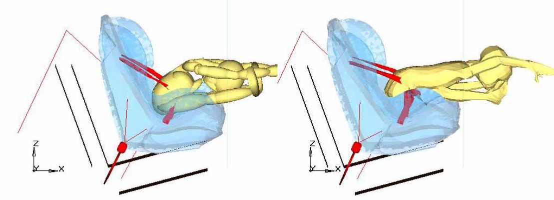

Figure 32. Simulation of Q3 dummy model (left) and human 3-year-old model (right) escaped from shoulder belts, 120 ms after impact.

Injury criteria of modeled poses – The effect of various poses on the response of a child model was evaluated in terms of relative injury reference values. The Nij value in compression-extension mode was higher than in the base or standard case for two postures; feet in hand and sleeping slanted. At both these postures the body was slouched, resulting in a changed neck orientation with respect to the impact direction. The changed orientation caused a different loading condition at the upper neck, where Nij was computed.

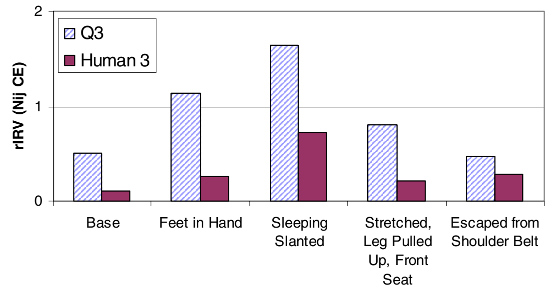

Figure 33. Relative Injury Reference Values (rIRV) of Nij compression-extension of simulations with Q3 dummy model and human 3-year-old model in Group I seat with ECE-R44 pulse at different poses.

The head excursion rIRV was similar for all simulations except for the simulation where the child escaped from the shoulder belts. There, the ECE-R44 excursion limit was exceeded for both dummy and human model. The human model exceeded the head excursion limit more than the dummy model, which was caused by the spinal elongation that can be observed in figure 32. Due to the absence of a front row structure in an ECE-R44 setup, as well as in the current simulation setup, the effect of exceeding the head excursion limit was not quantified. However, high head accelerations and neck loads are likely to occur.

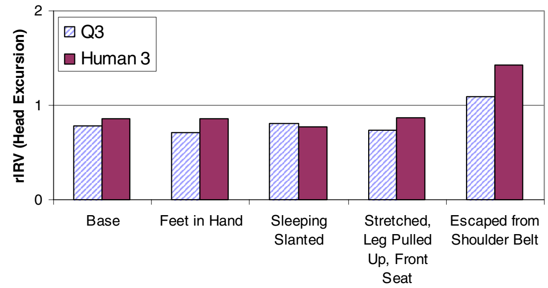

Figure 34. Relative Injury Reference Values (rIRV) of head excursion of simulations with Q3 dummy model and human 3-year-old model in Group I seat with ECE-R44 pulse at different poses.

Due to the asymmetric loading condition, lateral motion and computed forces and moments were expected to occur. Two postures where the body was slanted and the neck was laterally flexed initially were the child sleeping slanted and the child with one leg stretched against the front row seat. For these two simulations a large lateral shear force in the lower neck was observed, as shown in figure 35. The forces were up to two times higher than observed in the base model and were also twice as high as the IRV.

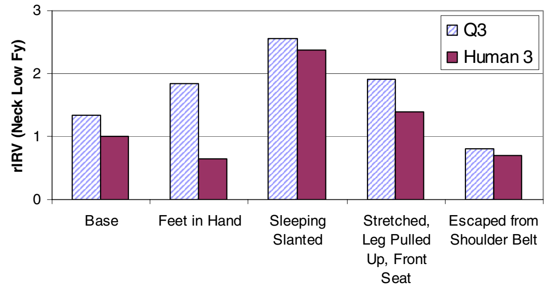

Figure 35. Relative Injury Reference Values (rIRV) of lower neck lateral shear force of simulations with Q3 dummy model and human 3-year-old model in Group I seat with ECE-R44 pulse at different poses.

In order to further investigate the occurrence of lateral loading, the lateral bending moment in the human model neck is plotted for all poses in figure 36. Lateral neck bending occurs in the base model, due to the asymmetric mounting configuration of the CRS on the test bench by means of a three-point belt restraint. However, when additional asymmetry was introduced by slanting the human child model, the lateral bending moments that occurred in the lower neck were up to twice as high and exceeded the IRV for lateral neck moment that was defined at 16.5 Nm in table 6.

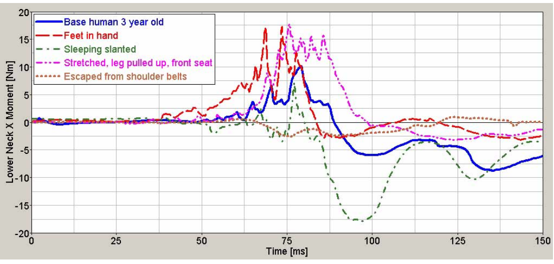

Figure 36. Lower neck lateral bending moment of simulations human 3-year-old model in Group I seat with ECE-R44 pulse at different poses.

The results from the different poses indicated that loading levels were not dramatically different from a properly restrained child. However, the simulation of the child that escaped from the shoulder belt exceeded the head excursion limit. A child that was not symmetric relative to the shoulder restraint sustained lateral forces and moments in the neck of a level that might potentially induce injury.

DISCUSSION

The development of the Q3 dummy model, based on the pre-existing model, resulted in a model that showed comparable results with respect to hardware dummy tests. Additionally, the design requirements for the dummies were met or almost met for all tests except for the thoracic impactor tests. The thorax of the model was consistently stiffer than the corridors. This resulted in forces that were approximately 4 times higher than required at deflections lower than required. Improvements to the dummy model thorax are necessary in order to show a similar response to the hardware dummy. The need to fit the thorax corridors is subject to discussion since these corridors were developed based on scaling from cadaveric adult data, an approach that involved a large number of assumptions.

Q1.5 dummy development through anthropometrical scaling was a useful process in scaling the outer dimensions of the dummy. The stiffness properties of contact characteristics and joint resistance models were scaled accordingly. The latter approach was validated by the component simulations in which the stiffness characteristics were tested. Manual adaptations were necessary in order to achieve a correct segment mass distribution, since the MADYMO/Scaler routine did not scale based on those. Component validation showed that the developed Q1.5 model response fell inside or was just outside the requirements, except for the thorax. The lack of thorax validation was a direct consequence of the scaling approach used to develop the dummy from the Q3 model. It must be stated however, that the amount of tests performed was limited.

The developed human child models were scaled from an adult anthropometry. The resulting models met anthropometrical requirements from the CANDAT database. However, the procedure involved large scaling ratios in which the potential for errors in scaling the various stiffness and force models was large. Within the current human child model development, structural differences between humans and children and variation of material properties by age were not taken into account. For example, the long bones of children have growth plates and their bone tissue is generally more elastic than that of adults [Ivarsson, 2004]. In an FE environment these structural and material differences between children and adults can be taken into account intrinsically [Okamoto, 2002], while in a multi-body environment they can be taken into account by using more advanced scaling techniques, incorporating additional joints and material properties.

Validation of human child models was not performed in this study. Since cadaveric child data is unavailable due to ethical considerations, validation needs to be performed differently. A possible approach in validating a human model can be to perform crash reconstruction. Correlation data between impact severity and resulting injuries is often available in detailed car crash databases, also for children. When a large number of crashes are simulated and the computed injury criteria from the human model match up with the injuries recorded in the medical records, confidence in the injury predictive capacity of the human models can be achieved. In a first attempt to validate the human child models, in this paper the response of the validated Q3 model was compared with the 3-year-old child human model. Most responses were similar. However, large differences existed in upper neck extension moment and in sternum displacement.

In the posture study, the dummy model was applied for conditions it was not developed for. The extreme joint orientations resulted in transient effects at the beginning of the simulation. As a result model adaptations were necessary. The human model could be positioned in any of the postures without these transient effects, due to larger ranges of motion in the joints. The model response in the full scale CRS environment was comparable between human and dummy model, which indicated that both dummy and human model were valid tools to perform this type of investigation. No validation data was available for the CRS itself and for the human and dummy model in a full scale environment and therefore the results from this study should be considered as indications of trends that might occur in various poses.

In many of the most common poses the body was slanted, which caused an asymmetrical loading condition at the shoulder belts. This involved lateral movement of the head and lateral forces and moments in the neck of a level that is potentially hazardous. When a child was escaped from its shoulder belt restraint, the ECE-R44 criterion for head excursion was exceeded. Besides large axial forces in the spine, impact with a front seat can cause severe head and neck injuries.

CONCLUSIONS

From this study the following conclusions were drawn:

- The Q3 model update was validated against component test data and met the corridors, except for the thoracic response, which was approximately four times too stiff.
- The Q1.5 model, developed through anthropometrical scaling, largely fulfilled the dummy design requirements, again except for thoracic response
- The human child models were developed based on anthropometrical scaling, which resulted in human models resembling a 1.5 and a 3-year-old from CANDAT database.
- A posture study showed that children tend to move around in their CRS on longer drives, resulting in slanted and slouched positions.
- Correctly restrained children in a Group I seat were able to escape from their shoulder restraint, which increases risk of injury.
- Simulation of the various poses with the above discussed human surrogate models indicated that lateral neck loads were twice as high in slanting positions. Slouching resulted in higher neck loads as well.
- The simulation of the child that escaped from the shoulder belt was shown to be hazardous since the head excursion limit was exceeded by over 20 cm.
- Virtual testing was shown to be a useful method to investigate the types of crash conditions that may occur in the field, but that are difficult to test in an experimental environment.

ACKNOWLEDGEMENTS

The authors would like to acknowledge Westeinde Hospital in The Hague for the use of their CT scanner.

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APPENDIX 1: Q3 component validation

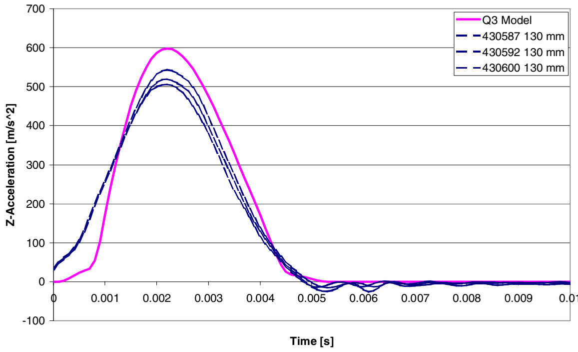

Figure 37. Z-acceleration of Q3 head frontal drop test validation at 130 mm height.

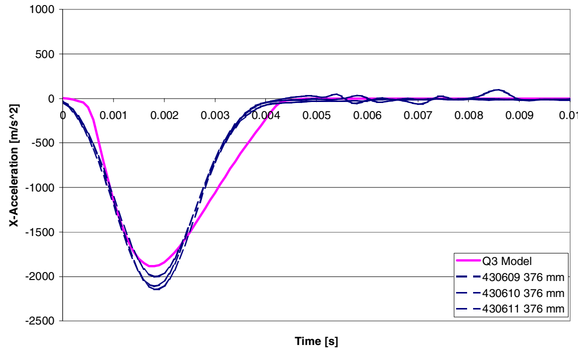

Figure 38. X-acceleration of Q3 head frontal drop test validation at 376 mm height.

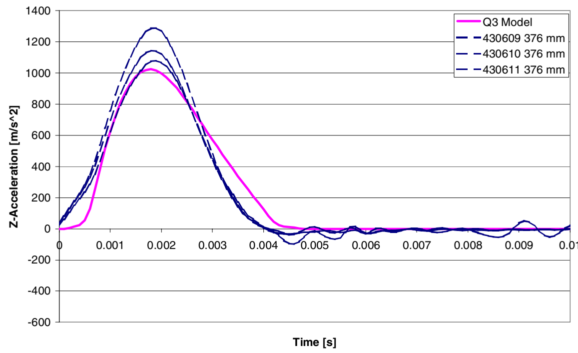

Figure 39. Z-acceleration of Q3 head frontal drop test validation at 376 mm height.

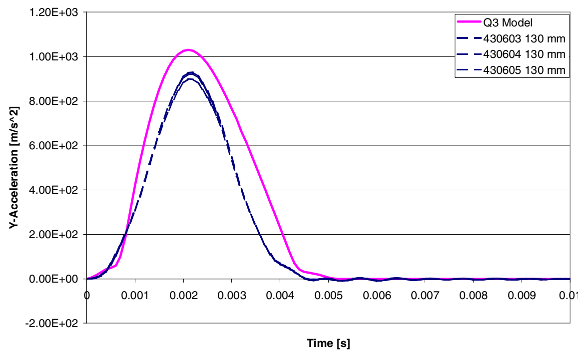

Figure 40. Y-acceleration of Q3 head lateral drop test validation at 130 mm height.

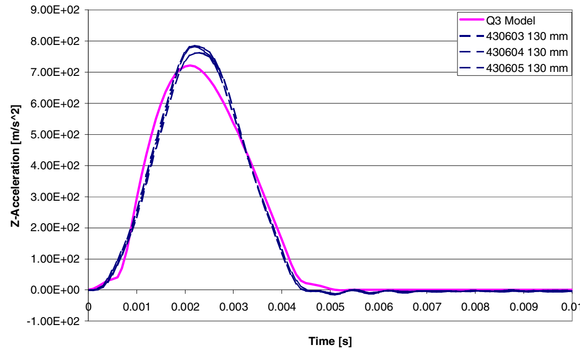

Figure 41. Z-acceleration of Q3 head lateral drop test validation at 130 mm height.

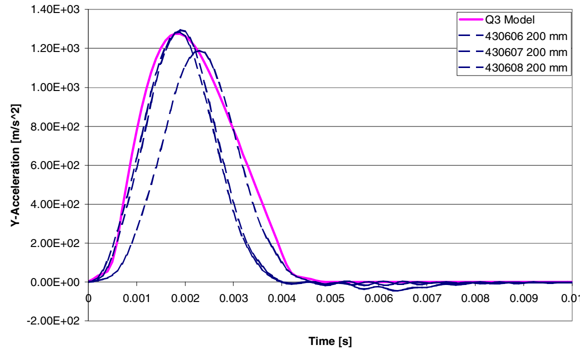

Figure 42. Y-acceleration of Q3 head lateral drop test validation at 200 mm height.

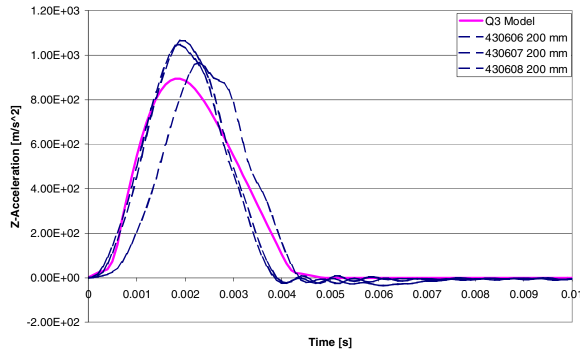

Figure 43. Z-acceleration of Q3 head lateral drop test validation at 200 mm height.

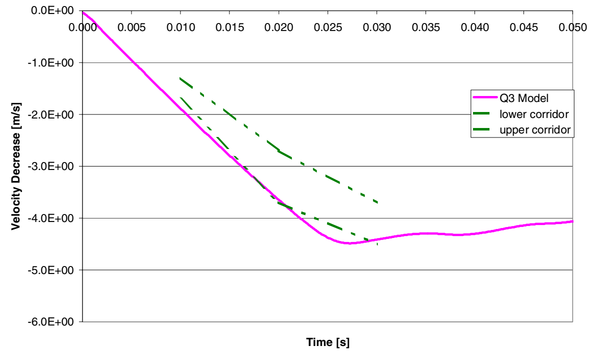

Figure 44. Velocity decrease of Q3 neck extension pendulum test validation at 3.9 m/s impact speed.

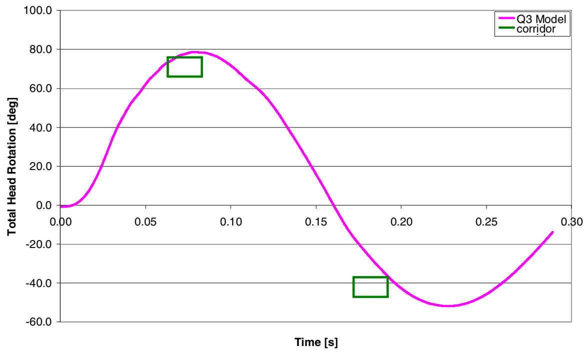

Figure 45. Total head rotation of Q3 neck extension pendulum test validation at 3.9 m/s impact speed.

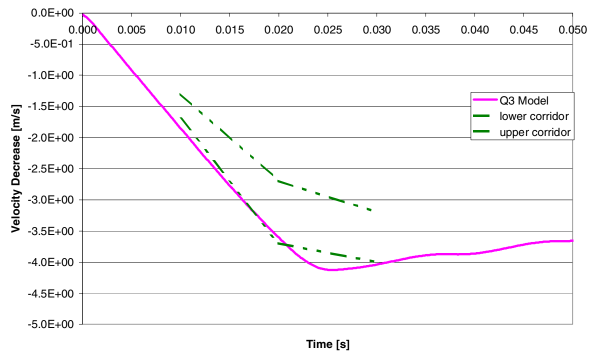

Figure 46. Velocity decrease of Q3 neck lateral flexion pendulum test validation at 3.9 m/s impact speed.

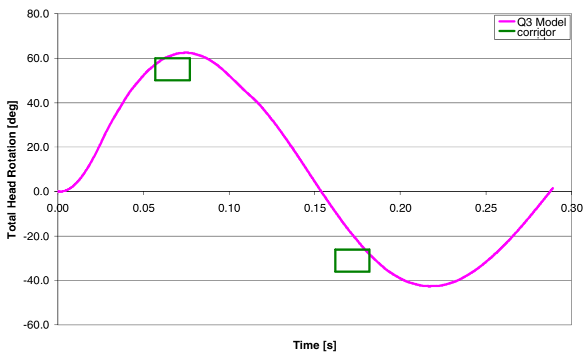

Figure 47. Total head rotation of Q3 neck lateral flexion pendulum test validation at 3.9 m/s impact speed.

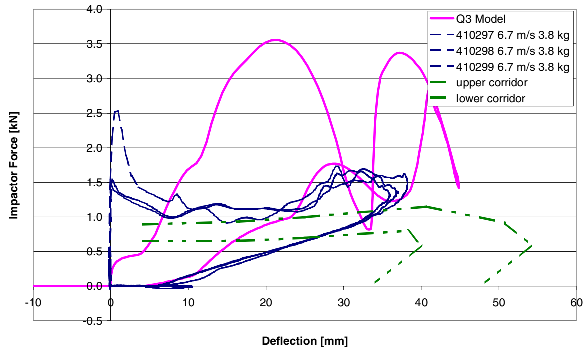

Figure 48. Force-deflection of Q3 thorax frontal impactor test at 6.7 m/s with 3.8 kg mass.

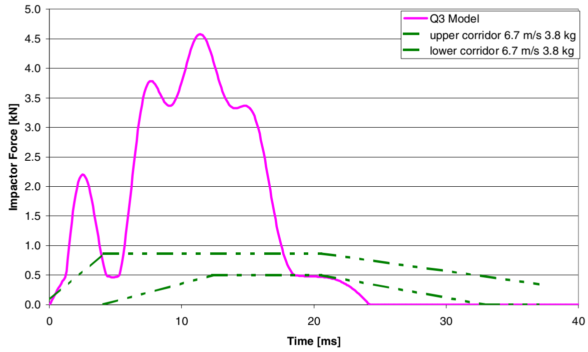

Figure 49. Force history of Q3 thorax lateral impactor test at 6.7 m/s with 3.8 kg mass.

APPENDIX 2: Q1.5 component validation

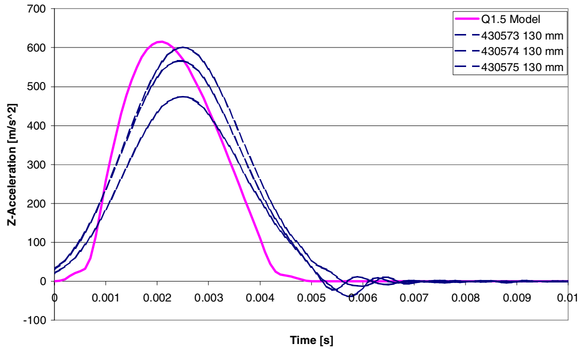

Figure 50. Z-acceleration of Q1.5 head frontal drop test validation at 130 mm height.

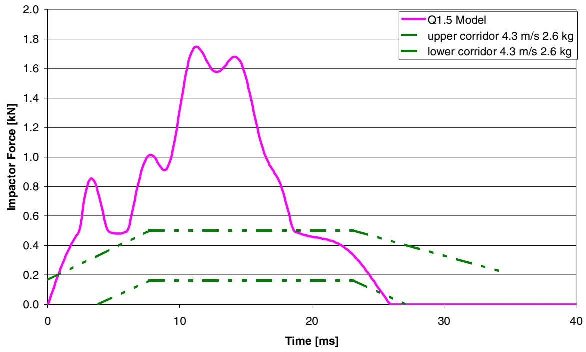

Figure 51. Force history of Q1.5 thorax lateral impactor test at 4.3 m/s with 2.6 kg mass.

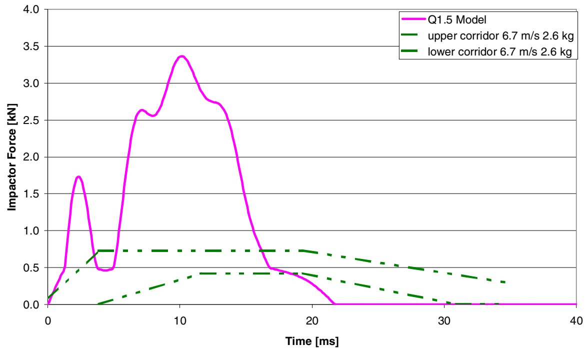

Figure 52. Force history of Q1.5 thorax lateral impactor test at 6.7 m/s with 2.6 kg mass.

THE EFFECT OF SWEDISH TETHERS ON THE PERFORMANCE OF REAR FACING CHILD RESTRAINTS IN FRONTAL CRASHES

Chris Sherwood
Yasmina Abdelilah
Jeff Crandall
Center for Applied Biomechanics
University of Virginia
USA
Paper Number 05-0346

ABSTRACT

Rear Facing child restraints (RFCR) have various component designs which can couple the restraint to the vehicle. Swedish tethers, which link the upper portion of the child restraint to the vehicle floor, prevent rearward rotation in rear impacts and during rebound in frontal crashes. They also simplify installation of restraints by allowing better control of the installation angle and removing the need of spacer devices. The objective of this study was to test the effect of Swedish tethers on RFCR in frontal crashes. The tethers reduced forward excursion and rotation, and had a positive but minor effect on injury values. The more secure attachment to the vehicle caused by the Swedish tether could also be beneficial in other crash types.

INTRODUCTION

The vehicle belt, lower LATCH belt, or ISOFix anchors serve as the primary components which attach Rear Facing Child Restraints (RFCR) to the vehicle. There are other devices, however, which can be used in addition to the primary components. Anti-rotation legs, Australian tethers, Swedish tethers, the ISOFix base, and anti-rebound bars are each designed to change the kinematics of the child restraint in different crash types.

Federal Motor Vehicle Safety Standard (FMVSS) No. 213, "Child restraint systems," requires RFCRs to meet the performance requirements of the standard when secured to the standard test seat assembly using (1) the lap belt only or (2) the lower LATCH (Lower Anchorages and Tethers for Children) anchorages only. NHTSA does not use a means supplemental to the lap belt/lower LATCH anchorages, such as a tether or a bar, of securing RFCR to the seat assembly in the agency's compliance test. In the past, NHTSA found that a very high percentage of parents did not use a supplemental tether strap to secure their child seats even when they knew the strap was needed to provide their child protection. The agency concluded that there was a strong likelihood that a tether or a bar would be misused with the seat, and that FMVSS No. 213 should thus require that child restraints must meet minimum requirements of the standard without supplemental tethers.

Swedish tethers prevent rear rotation in rear impacts and during rebound in frontal impacts [1]. They link the upper portion of the child restraint to the vehicle floor, and may also have benefits in non-frontal crash types by more rigidly attaching the RFCR to the vehicle [2,3]. They can be attached to built-in anchor points or to the front seat base structure. The tether may reduce excursion in side impacts (lateral) and rollovers (upward/rearward).

In addition to the effect it may have in vehicle crashes, the tether may also have benefits during installation. RFCRs have a recommended range of child restraint angles. The RFCR angle should be approximately 45 degrees (with respect to vertical), but no greater [3]. Since young children cannot hold their heads upright due to their weak neck musculature, the reclined angle prevents the head from flopping forward and cutting off the airway. At angles greater than 45 degrees, however, the child restraint provides less support for the head and neck. Variations in child restraint design, vehicle seat design, attachment equipment (LATCH, 3 pt belt), and the location of attachment anchors result in many RFCRs positioned at incorrect angles [4]. The most common method, which allows adjustment of the angle, is to place a spacer (typically rolled up towels or foam noodles) under the base of the RFCR near the seat back (Figure 1). Swedish tethers provide the opportunity to control and easily adjust this angle.

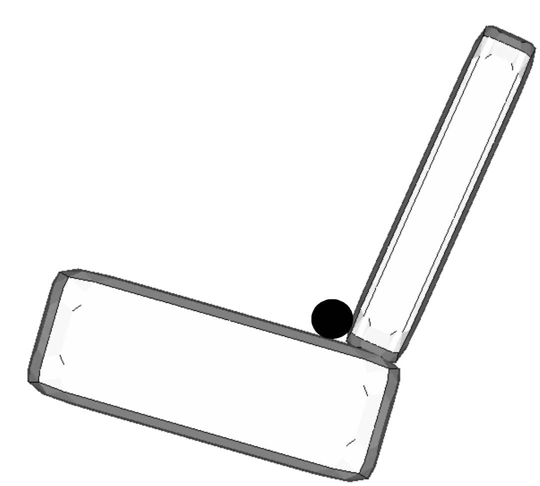

Figure 1. Schematic of RFCR with spacer placed under base to correct installation angle.

It is expected that Swedish tethers would have a minimal effect prior to rebound in frontal crashes because they are not rigid and would go into slack upon impact. However, the tension in the tether and the absence of the spacer may change the initial position of the child restraint and alter its interaction with the vehicle seat. The objective of this study was to test the effect of Swedish tethers on RFCR in frontal impacts.

METHODS

Six frontal sled tests were performed to measure the response of a restrained dummy in rear facing child restraints with and without a Swedish tether. All tests were conducted at a 49 km/h impact speed with an acceleration pulse similar to that specified in FMVSS 213 (23g peak, 90 ms duration). The tests were performed on a vehicle buck that represented a popular minivan. A third row bench seat and seat back were rigidly attached to the buck to create a durable, consistent seat system.

The CRABI 12 month old dummy was used to represent the child occupant. The dummy was equipped with head, chest, and pelvis accelerometers as well as upper and lower neck load cells. Electronic data were sampled at 10,000 Hz and were filtered per Society of Automotive Engineers (SAE) Recommended Practice J211. The tests were recorded at 1000 frames/sec with side and overhead digital video cameras.

Three convertible child restraint models were tested in the rear facing orientation: the Britax Roundabout, the Evenflo Comfort Touch, and the Safety First Comfort Ride. Each child restraint was restrained using the lower LATCH belt in two restraint conditions a) with and b) without a Swedish tether. The Britax Roundabout manual states that the upper tether, typically used as the tether when forward facing, can also be used as a Swedish tether when rear facing. The upper tether was used as a Swedish tether in the Evenflo and Safety First seats as well, although it was not instructed by the manual. All conditions had identical initial angles (40 ± 0.5 degrees, measured with respect to vertical at the child's back). The 40 degree value was chosen because children at 12 months of age (the size of the dummy used in these tests) can sit more upright than newborns.

Without the Swedish tether, the lower LATCH belt was tightened with the foam spacer in place until the appropriate restraint angle was reached. With the Swedish tether, the foam spacer was not used as it was not needed to provide the correct restraint angle. Positioning the child restraint was an iterative process in which the tensions of the lower LATCH belt and Swedish tether were adjusted until the correct angle was attained. The lack of the foam spacer changed the interaction between the restraint and the vehicle seat, but each restraint was installed with the purpose of a) providing a consistent angle and b) attaching the restraint to the vehicle seat as tightly as possible. Figures 2-4 show pre-crash side photographs of the tests, and data on initial positions and tether tensions are included in Table 1. The Head X position was measured with respect to an arbitrary reference point. In tests without the Swedish tether, a secondary tether was placed on the child restraint without any tension, and was only used to prevent the child restraint and dummy from striking the rigidized seat back during rebound.

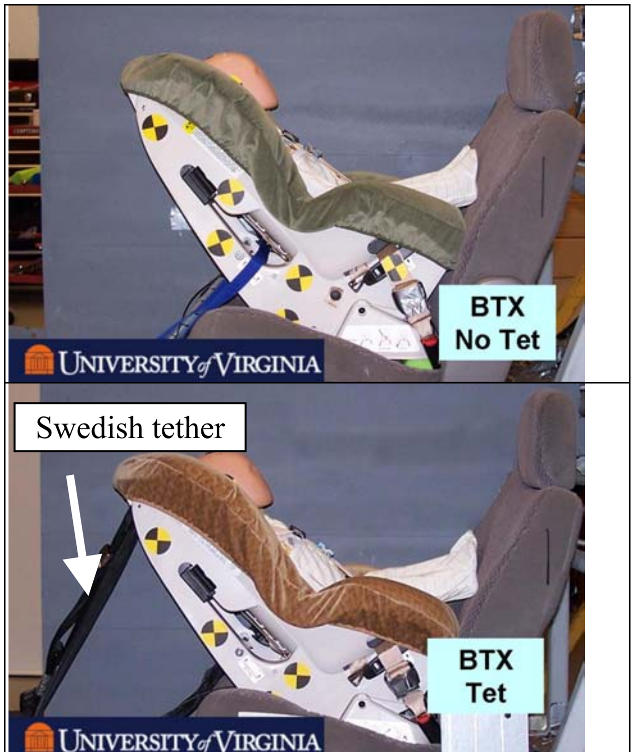

Figure 2. Pre crash photos of the Britax restraint without Swedish tether (Britax No Tet, top) and with Swedish tether (Britax Tet, bottom).

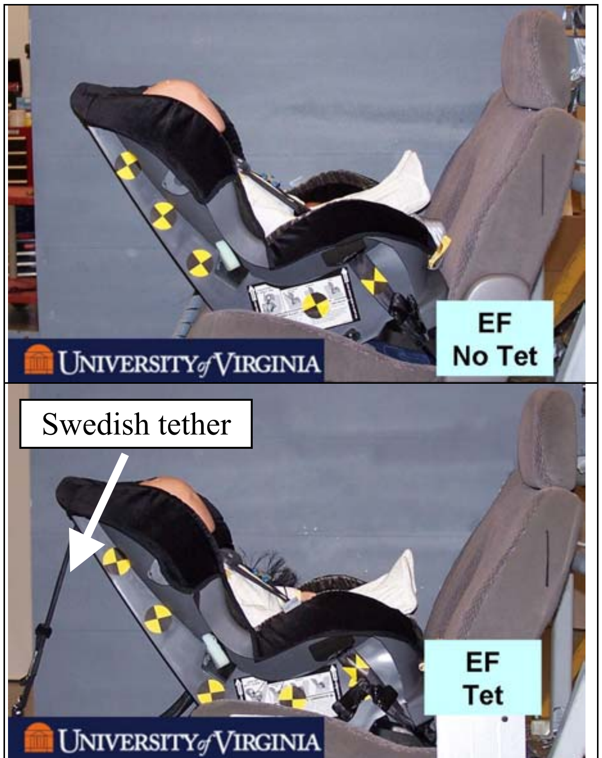

Figure 3. Pre crash photos of the Evenflo restraint without Swedish tether (EF No Tet, top) and with Swedish tether (EF Tet, bottom).

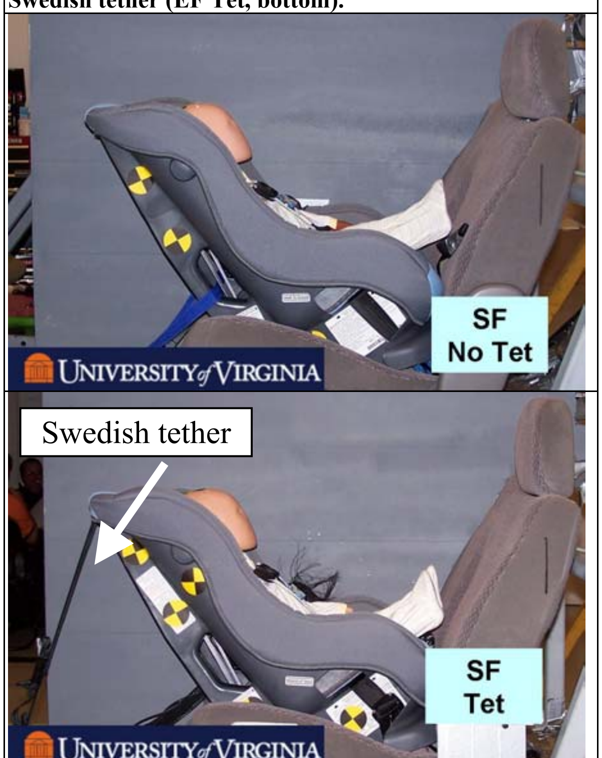

Figure 4. Pre crash photos of the Safety First restraint without Swedish tether (SF No Tet, top) and with Swedish tether (SF Tet, bottom).

Table 1.
Initial position data

	BTX No Tet	BTX Tet	EF No Tet	EF Tet	SF No Tet	SF Tet
Head X position (cm)	56.0	58.0	65.5	66.7	60.0	60.4
CR angle (deg)	40.3	39.5	40.2	39.7	39.9	40.0
Lower LATCH tension (N)	>90	No data	44	>90	>90	67
Swedish tether tension (N)	NA	31	NA	>90	NA	53

RESULTS

The Swedish tether changed the kinematics of each child restraint, but not by large amounts. Figures 5 and 6 show the kinematics of both conditions for the Evenflo restraint. For each restraint the tether reduced the maximum child restraint angle and horizontal excursion distance measured at a point near the child's head (Table 2). The average reduction in movement caused by the addition of the tether was 5.3 degrees and 1.8 cm.

Table 2.
Child restraint kinematic data

	BTX No Tet	BTX Tet	EF No Tet	EF Tet	SF No Tet	SF Tet
Max CR angle (deg)	33	25	18	13	25	22
Max Horiz Excursion (cm)	75.4	73.8	77.4	74.2	70.3	69.9

The sensor injury measurement values are provided in Table 3. The same data are shown graphically in Figure 7, when the percentage of change due to the addition of the Swedish tether is calculated for each injury measure and each restraint. The effect of the tether varied by injury measure and by child restraint. The tether caused an increase greater than 30% in only one instance (upper neck shear), while there were five instances of the tether causing a decrease in an injury measurement by more than 30% (HIC, lower neck shear, lower neck extension).

Figure 5. Evenflo restraint without Swedish tether at 0, 25, 50, 75, and 100 ms.

Figure 6. Evenflo restraint with Swedish tether at 0, 25, 50, 75, and 100 ms. (images flipped to allow easier comparison)

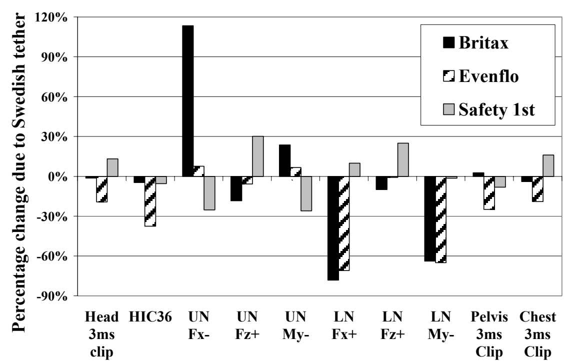

Figure 7. Graph of the effect of Swedish tether on each injury measure, for each restraint.

Table 3.
Dummy injury measurement peak values

	BTX No Tet	BTX Tet	EF No Tet	EF Tet	SF No Tet	SF Tet
Head 3ms clip	52.4	51.7	76.7	62	55.9	63.3
HIC_{36}	560	534	690	431	436	412
UN Fx	-220	-468	-372	-400	-409	-306
UN Fz	1395	1137	1332	1255	1190	1548
UN My	-13	-16	-12	-13	-13	-10
LN Fx	324	71	460	133	437	481
LN Fz	1482	1335	1469	1456	1299	1623
LN My	-11	-4	-14	-5	-14	-14
Chest 3ms clip	47.7	46	55.3	44.8	40	46.2
Pelvis 3ms clip	44.8	46	67.3	50.5	52.5	48.2

There were secondary peaks which occurred when the tether went into tension during rebound. However, these peaks never approached the peak values which occurred earlier in the test.

DISCUSSION

Each restraint was positioned on the vehicle seat with two primary objectives. The first was to position the restraint with consistent angles because installation angle is critical for young children, and because restraint angle significantly affects injury biomechanics. The second was to attach the child restraint to the vehicle as tightly as possible. The tension in the Swedish tether and the removal of the foam spacer changed the restraint's interaction with the vehicle seat, and resulted in different lower LATCH tensions. These varying tensions, however, are the real world by-product of the addition of the Swedish tether and represent a fundamental factor that should be included when comparing the two restraint conditions.

The addition of the tether had the practical benefit of allowing better control of the child restraint angle. Further studies are necessary, however, to ensure that the addition of the Swedish tether does not result in other misuse scenarios. Although the tether tension is minimal during installation and decreases to zero during the primary portion of the frontal crash, strength requirements of the anchor during rebound and in rear impacts must be analyzed.

The addition of a Swedish tether changed the kinematics of the child restraints, although the results varied between the child restraints tested. Rotations and excursion distances of the upper portion of the child restraint were reduced, which would reduce the chance of the child restraints striking vehicle structures such as front seats or the vehicle dash.

The effect of the Swedish tether on injury measures was less consistent. The addition of the tether generally caused an earlier onset of accelerations, but there was not a concomitant decrease in peak acceleration. The effects varied across injury measures and across child restraint model. Only six values (out of 30 calculated) changed by more than 30%. In five of these six instances, the tether resulted in reductions in injury measures. All but one of these instances occurred in the neck shear or moment measures, which are likely the least biofidelic sensors in the CRABI dummy. Thus, while the results varied, the overall effect of the Swedish tether was a negligible reduction in injury severity. Further testing on multiple vehicle seats would provide more support for these findings.

Although not measured as part of this study, the tether had significant effects on the lateral and vertical coupling of the child restraint. Although different coupling methods were tested, Kelly et al. (1995) showed that increased coupling of the child restraint to the vehicle improved test results in side impacts. The increased rigidity afforded by Swedish tethers would be expected to have benefits in side crashes and rollovers, but this area requires more research.

CONCLUSIONS

The results provide evidence that use of a Swedish tether causes a positive but small benefit on the injury risk to children in RFCRs in frontal crashes. The advantage of tethers during installation and possibly in other crash types (side impacts, rollovers) suggests that the use of Swedish tethers in RFCR could be beneficial. Further work is needed to consider issues such as misuse, tether anchors, and the effect in other crash modes.

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DEVELOPING AN EFFECTIVE AND ACCEPTABLE SAFETY BELT REMINDER SYSTEM

David W. Eby
Lisa J. Molnar
Lidia P. Kostyniuk
Jean T. Shope
University of Michigan Transportation Research Institute
United States
Paper Number 05-0171

ABSTRACT

Front-seat safety belt use in the United States (US) was 80 percent in June, 2004. This rate represents the highest ever for the US, but indicates that there is still a sizable minority of people who do not always use safety belts despite mandatory seat belt laws in all but one state. Changing the behavior of these people will require new and innovative countermeasures. Little research has systematically investigated the effectiveness, feasibility, and acceptance of vehicle-based countermeasures for promoting safety belt use. The purpose of this project was to promote safety belt use in the US by conducting research to develop an effective in-vehicle safety belt reminder system. Project activities included a nationwide survey of part-time safety belt users, development of potential safety belt reminder system ideas, and a series of focus groups with part-time safety belt users. The results indicated that the most effective and acceptable safety-belt reminder system concept was one that was adaptive; that is, one that changes its signal type and presentation modality depending on belt use behavior over some metric (time, distance, or speed). The study also assessed and developed an potential reminder system ideas for informing drivers about back-seat belt use.

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INTRODUCTION

The single most effective technology for reducing or preventing injuries from a motor vehicle crash is the safety belt restraint system. This system, however, is only effective if it is used. The most recent nationwide survey of safety belt use in the United States (US), the National Occupant Protection Use Survey (NOPUS), estimated that 80 percent of front-outboard motor vehicle occupants use their safety belt (Glassbrenner, 2004). While this is the highest rate ever in the US, the rate is lower than many other developed countries (e.g., Boase, Jonah, & Dawson, 2004) and shows that a significant portion of US travelers do not use safety belts, even though belt use is mandated in all but one state.

For nearly 30 years, the US federal government and vehicle manufacturers have developed and implemented numerous technologies for promoting safety belt use, with varying degrees of success. In the 1970s, the federal government mandated two vehicle-based safety belt use promotion technologies. The first required vehicles manufactured after 1971 to have a continuous buzzer-light safety belt reminder when safety belts were not used (vehicles equipped with air bags were excluded; Robertson, 1975). Analysis of belt use before and after the buzzer-light systems were installed showed no statistical increase in safety belt use (Robertson & Haddon, 1974). The federal government then mandated that all new vehicles sold after August 15, 1973 be equipped with a safety-belt-ignition-interlock system that prevented the vehicle from starting if the driver and front-right passenger were not using safety belts (Buckley, 1975). Despite the fact that these interlock systems increased safety belt use by as much as 30 percentage points (see e.g., Robertson,1975), public opposition to them led Congress to rescind the legislation in 1975. The three main reasons cited for opposition to safety-belt-interlock system were: 1) problems with proper functioning of the system when no front-right passenger was present; 2) safety concerns associated with preventing drivers from rapidly starting a vehicle in the event of an emergency; and 3) the relative ease of disabling the ignition interlocks.

After 1975, the US federal government turned its attention to legislating safety belt use. In the 1980s, the federal government began to urge states to pass legislation that required the use of safety belts, with New York passing the first mandatory safety belt use law in 1984. While these laws were initially unpopular in many states, every state except New Hampshire has now passed a safety belt use law. There is clear evidence that these laws have been successful in increasing safety belt use (see e.g., Eby, Molnar, & Olk, 2000; Reinfurt, Campbell, Stewart, & Stutts, 1990; Ulmer, Preusser, & Preusser, 1994).

In the 1980s, the federal government required that vehicles have passive occupant protection systems, and manufacturers responded by developing the automatic belt systems in which the shoulder belt automatically positions itself after the driver starts the vehicle. Research has shown that automatic belt systems do increase safety belt use (Streff & Molnar, 1991). However, these systems were judged as being less effective than the 3-point safety belt and were not well liked by consumers. When the federal government clarified its definition of “passive occupant protection” to encompass air bags, automatic belts were largely eliminated from newly manufactured vehicles.

Recent attention has turned to the development of new in-vehicle technologies for increasing belt use (NHTSA, 2003; Transportation Research Board, TRB, 2003). One promising technology is the safety belt reminder system. Since 1975, all new vehicles in the US have been required to display a 4-8 second signal if the driver does not use the safety belt after starting the vehicle. Once the belt is fastened, the signal stops. This relatively benign reminder system is easily ignored. Therefore, further research is needed to develop more effective and acceptable in-vehicle technologies to promote safety belt use, such as safety belt reminder systems.

The Project

The purpose of the project was to promote safety belt use in the US by gaining a better understanding of the effectiveness of current safety belt reminder systems as well as suggesting appropriate improvements. The project examined several aspects of vehicle-based safety belt use technologies. Two main research tasks were completed: a nationally-representative survey of part-time safety belt users and a series of focus groups with part-time safety belt users. A literature review was also performed. Results from this review appear throughout this document.

The project design was iterative in nature; that is, after each task, University of Michigan Transportation Research Institute (UMTRI) personnel met with sponsor representatives and we refined our thinking about the characteristics that would lead to effective and acceptable in-vehicle safety belt promotion technology. Combining information obtained from the literature review, UMTRI’s background in occupant protection research, and the sponsor’s expertise in developing in-vehicle safety technology, we developed a set of topics for the telephone survey and focus groups that we believed were relevant to the development of safety belt reminder systems. These topics included:

- The demographic trends of part-time safety belt users;
- Part-time safety belt users’ attitudes toward belt use;
- Reasons for part-time belt use by seating position;
- Which types of system were thought be effective and acceptable to part-time users.

After detailed discussion with all parties on the project, we realized that the number of potential systems we could investigate was vast. The decision was made, therefore, to investigate *features* of potential systems rather than example systems per se. These features were:

- The type of signal;
- The signal presentation method;
- The signal recipient.

In addition, safety-belt-interlock systems have the potential to be effective in-vehicle technologies for promoting safety belt use. As discussed previously, safety-belt-*ignition* interlocks were mandated in the US until public dissatisfaction led to their repeal. Other vehicle systems could be interlocked with safety belt use, such as the heating/cooling or entertainment systems. Therefore, we investigated features of this potential technology in the project.

METHODS

Nationwide Telephone Survey

The objective of the telephone survey was to gather information from a nationally representative sample of part-time safety belt users about their nonuse of safety belts, the reasons for this behavior, and what it would take to get them to use their safety belts. For the purpose of this survey, a part-time safety belt user was defined as a person who, by self-report, had not used a safety belt on at least one occasion in the last year either as a driver or passenger (front or back seat) in a private car that had safety belts available. This included not using a safety belt for some portion of the trip, other than a few moments at the very beginning or the very end of the trip.

A telephone survey instrument was developed with a screener to identify part-time safety belt users and to collect basic demographic information from those who did not qualify as part-time users. Once part-time users were identified, they were asked about their safety belt nonuse by seating position, reasons for safety-belt non-use, the perceived usefulness and acceptability of a set of system features of in-vehicle safety belt promotion technologies. The three system features investigated in the survey were: the signal type; the ways in which the signal could be delivered; and the target occupant(s) for the signal. We also investigated, to some extent, acceptability and effectiveness of these features for the driver when he or she is not belted (driver-driver), for the driver when a passenger is not belted (driver-passenger); and for the passenger when he or she is not belted (passenger-passenger). Other survey topics included:

- How often respondent was driver and/or passenger;
- Questions about the last time respondent did not use safety belt;
- Questions about respondent's general safety belt nonuse as driver and as passenger;
- Questions to driver about belt use of his/her passengers;
- Demographics.

The telephone survey utilized a nationally representative random-digit-dial (RDD) sample design of households. The telephone interviews were conducted by a professional survey research firm using Computer Assisted Telephone Interviewing (CATI) from April 21 to June 25, 2003. In all, there were 1,100 completed interviews from part-time safety belt users. The final sample was weighted to reflect regional population distributions of the US.

To obtain the final sample of 1,100 part-time safety belt users, 21,670 telephone numbers were used. If not answered, a telephone number was tried up to six times. Of the 21,670 telephone numbers called, 8,557 yielded persons eligible for an interview; 6,613 resulted in an ineligible classification (not part-time safety-belt users, not age 18 or older, disconnected number, fax or data line, business number); and 6,500 numbers resulted in an unknown classification (no answer, answering machine, scheduled for call-back). Using standard definitions for the final disposition of samples for RDD telephone surveys (American Association for Public Opinion Research, 1998), the minimum response rate for this survey was 7.3 percent and the maximum response rate 12.9 percent.

Focus Groups

Twelve focus groups were conducted in Michigan to collect qualitative data from part-time safety belt users on the potential effectiveness and public acceptance of various features of systems that could be placed in cars to remind or encourage people to buckle up. Discussions also focused on safety belt use in general, including reasons for using and not using belts. Six of the groups were conducted in Ann Arbor, an urban/suburban area, and six in Clare, a rural area of the state. Within each location, two groups each of 18-29 year olds, 30-64 year olds, and people 65 and older were conducted.

Part-time safety belt users (defined as those who reported nonuse at least some of the time) were recruited through advertisements in local newspapers, as well as postings at local businesses, academic institutions, and community organizations (e.g., senior centers). Potential participants were screened via telephone to ensure that they met eligibility criteria (age 18 and older, valid driver license, part-time safety belt user). Background information on participants was collected during the telephone screening process. Each selected participant was scheduled for a focus group session and sent written confirmation through regular mail or e-mail according to their preference. Reminder telephone calls were made the day before each session. A total of 97 participants were recruited, and 87 actually appeared at their session and participated in the focus group. Participants received an honorarium of $50 cash as an incentive to participate. Each session lasted about 2 hours.

Discussion during the groups was guided by a moderator using a uniform set of questions. Participants were also provided with worksheets on which to record some of their answers to facilitate discussion. During each session, focus group participants were shown a short computer demonstration of a sample safety belt reminder system and asked about their reactions. Participants were told that the system was made up of three levels, with each level being activated only when the driver or front seat passenger remained unbuckled. If someone were to unbuckle during the trip, the system would start over from the beginning.

- <u>Level 1</u> corresponded to the current US government requirement that cars display a 4 to 8 second signal if drivers do not put on their seat belt after starting the car. This is typically a flashing light on the dashboard with some type of sound signal. In the sample reminder system, it included a blinking light and a beeping signal that came on when the engine started and continued for 8 seconds.
- <u>Level 2</u> included a sound signal (delivered by a female voice, a male voice, a buzzer, or a beeping signal) that repeats three times with 8 seconds in between.
- <u>Level 3</u> included either a buzzer or beeping signal that stays on continuously for 45 seconds.

Each group was audio-taped and a project staff member was present at each session, in addition to the moderator, to take notes. After each group, a debriefing session was held to identify important themes that emerged from the discussion. Analysis of the focus group discussions was based on the debriefings of project staff conducted immediately after each focus group, a review of notes taken during the focus groups, and the audio tape recordings of the focus group sessions.

RESULTS

Nationwide Telephone Survey

<u>Respondents</u>

About 60 percent of respondents were female; education level was fairly well-distributed; a wide variety of ages was included; and about 40 percent of respondents had young children in their household. Approximately 84 percent of the part-time safety belt users drove a car almost every day, and almost all were passengers in a car at some time in the past year. Nearly 80 percent of respondents were passengers in the back seat at least a few times in the last year. Nearly 42 percent did not use a safety belt within the previous week. When asked about seating position the last time a belt was not used, about 40 percent reported being a driver, 21 percent were passengers in the front seat, and about 34 percent were passengers in the back seat.

<u>Reasons for Nonuse of Belts</u>

We analyzed the primary reasons people gave for part-time nonuse of safety belts. In the survey, people were asked to think back to the last they did not use a safety belt in the past year and report the main reason for their lack of use. Respondents gave a wide variety of responses to this open-ended question. We discovered, however, that all of the responses fell into six broad nonuse categories: *cognitive/personal* (e.g., forgetting or not in habit); *comfort* (e.g., too big for belt or belt does not fit correctly), *convenience* (e.g., belt hard to reach), *low perceived risk* (e.g., only driving a short distance or not driving on public road), *social* (e.g., others not wearing belt), and *vehicle* (e.g., no belt in vehicle).

Figure 1 shows the percent of respondents in each category as a function of seating position. The most commonly cited reason for nonuse involved perceived risk, followed by cognitive/personal reasons. Comfort and convenience were also commonly-cited factors. Comparing reasons by seating position showed that risk was much more commonly cited by drivers than occupants in other seating positions; cognitive/personal reasons were more commonly cited for front-seat occupants than those in the back-seat; both comfort and convenience were more important for back-seat passengers than for the driver; and vehicle-based reasons were much more common for back-seat passengers.

Because so few respondents indicated that their lack of belt use resulted from social factors, this classification was excluded from further analyses. In addition, the vehicle-based reasons could not be addressed through any type in in-vehicle safety belt promotion technology; that is, if the belt is missing or the buckle is broken, a vehicle occupant cannot use

Figure 1: Main Reason for Not Wearing Belt Last Time by Seating Location

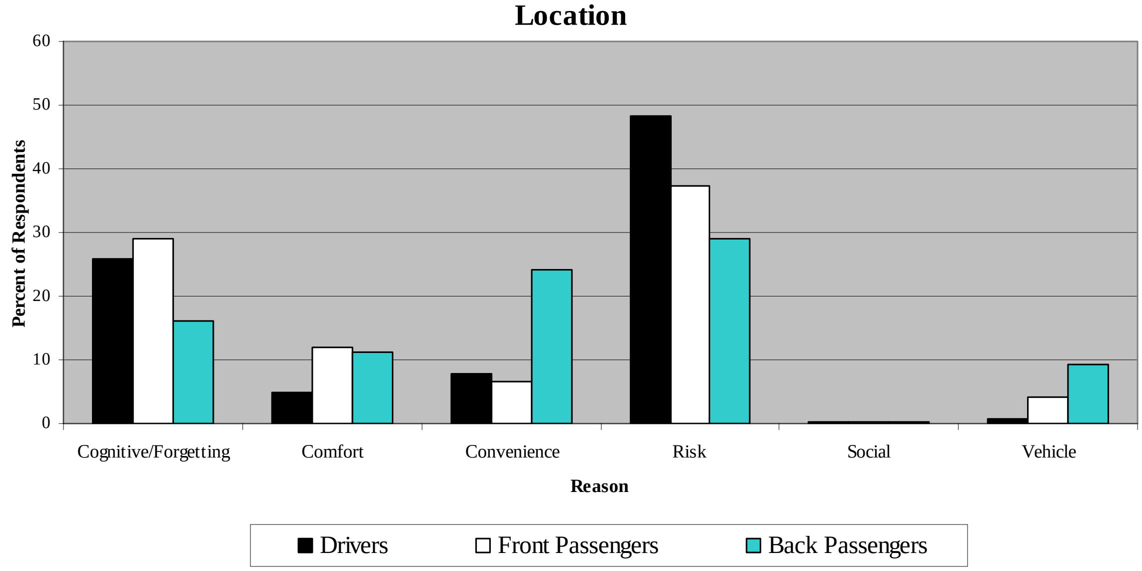
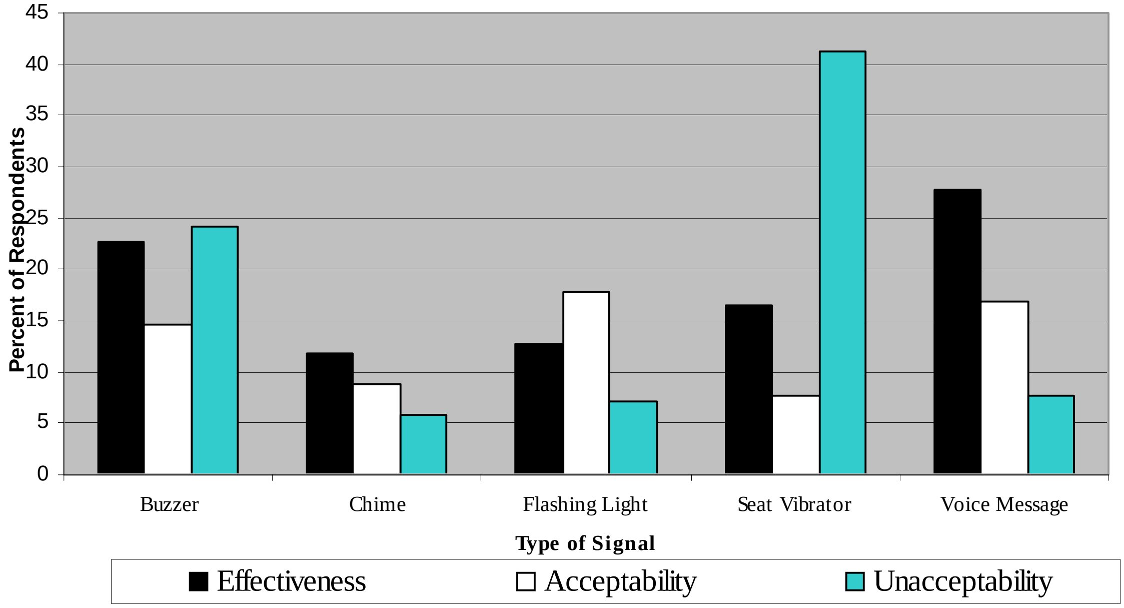


Figure 2: System Signal Preferences as a Driver Cognitive/Personal Group



the belt regardless of system effectiveness. Therefore, the vehicle-based classification was also excluded from further analyses. The classifications of comfort and convenience are not directly related to the development of effective in-vehicle belt promotion technologies as these factors are best addressed through human factors and ergonomic improvements to the vehicle interior. However, since these classifications were representative of many respondents and were of interest to the project team and sponsor, we combined them and addressed them separately from the in-vehicle belt promotion technology analyses.

Comfort and Convenience

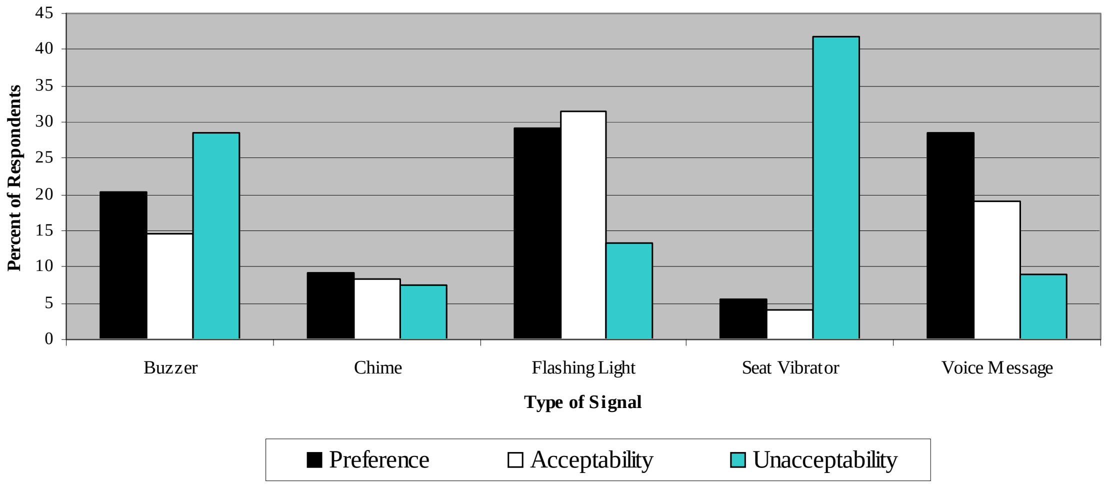
Survey results indicated that about 9 percent of respondents cited comfort and 13 percent cited convenience as the primary reason for nonuse of safety belts. As these classifications do not relate to the development of effective in-vehicle technology to promote belt use, the nationwide survey did not explore the dimensions of comfort and convenience in depth. A literature review on the topic, however, showed the following general results (Eby et al., 2004):

- Discomfort is a factor especially for shorter people (belt cuts into neck or clavicle);
- People who say they are not in the habit of buckling up are more likely to say belts are restricting and uncomfortable;
- Discomfort is more likely to be mentioned during winter and with heavier, bulkier clothing or coats;
- More complaints regarding comfort come from drivers over age 40;
- Women, overweight, and short drivers experience more problems with comfort/convenience;
- The most important convenience-related issues were:
  - Location and accessibility of buckle;
  - Levels of retraction force;
  - Perceptiveness to webbing extraction;
  - Susceptibility of webbing to tangling and twisting;
  - Belt buckle is too far back;
  - Belt trapped in door;
  - Awkward negotiating around clothes;
  - Belt twisting when getting it, when it retracts, and when adjusting it;
  - Belt locking up unexpectedly when leaning forward and when pulling belt;
  - Reaching for and gripping the belt buckle.

Cognitive/Personal

As mentioned previously, opinions about the type of signal, signal delivery method, and signal recipient (driver-driver; driver-passenger; and passenger-passenger) were examined separately for each of the nonuse classification groups of respondents. According to our survey, people who cite cognitive/personal reasons (usually forgetting) account for approximately 23 percent of part-time safety belt users nationwide.

Figure 3: System Signal Preferences as a Driver for an Unbuckled Passenger, Cognitive/Personal Group

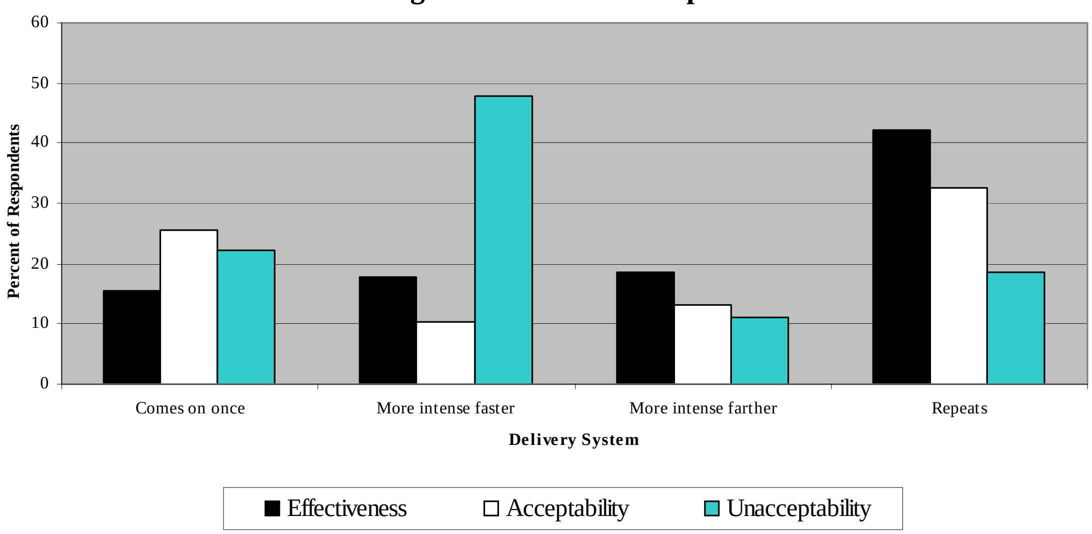


Type of Signal, Driver-Driver: Figure 2 shows the percent of cognitive/personal respondents who rated each type of signal on effectiveness, acceptability, and unacceptability as a driver. Unacceptability includes responses to the question: What signals would you definitely not want in your car? As can be seen in Figure 2, the voice message and buzzer scored the highest on perceived effectiveness. The voice message, flashing light, and buzzer also scored high on acceptability. The voice message, flashing light, and chime all scored low on unacceptability.

Type of Signal, Driver-Passenger: Figure 3 shows the percent of cognitive/personal respondents who rated each type of signal on their preference, acceptability, and unacceptability for a driver to be reminded that a *passenger* is not using a safety belt. Effectiveness in getting the passenger to buckle-up was not asked about for this situation because a respondent could not be expected to accurately predict the behavior of another vehicle occupant. As can be seen in Figure 3, the voice message, flashing light, and buzzer were selected most often as the preferred signal. The flashing light, voice message, and buzzer were also frequently cited as acceptable signals. The seat vibrator and buzzer were selected most frequently as unacceptable to drivers.

Type of Signal, Passenger-Passenger: The percentages of cognitive/personal respondents who selected each type of signal as the most effective for getting them to use a safety belt while they were

Figure 4: System Signal Delivery Preferences Cognitive/Personal Group



traveling in a vehicle as a *passenger* were the following: voice (24.7%); buzzer (22.1%); seat vibrator (15.6%); chime (12.8%); and flashing light (12.0%). We only asked about effectiveness, because passengers are not necessarily the owners of the vehicle in which they are traveling so acceptability/ unacceptability is not relevant.

Type of Signal Deliver, All Types of Systems: Figure 4 shows the percent of cognitive/personal respondents who rated each signal delivery method on effectiveness, acceptability, and unacceptability. The survey did not explore these questions as a function of seating position. As seen in Figure 4, repeating at a constant interval was the most frequently selected delivery system. Repeating, and a system that comes on once, were judged as the most acceptable overall. The most unacceptable system was one that became more intense the faster the vehicle travels.

Low-Risk

According to our survey, people who cite low risk as the reason for part-time belt use account for approximately 39 percent of part-time safety belt users nationwide. As with the cognitive/personal group, three system features were investigated in the survey: the signal type; the way in which the signal was delivered; and who received the signal. We also investigated, to some extent, acceptability and effectiveness of these features for the driver when he or she is not belted (driver-driver), for the driver when a passenger is not belted (driver-passenger); and for the passenger when he or she is not belted (passenger-passenger).

Type of Signal, Driver-Driver: Figure 5 shows the percent of low-risk-based respondents who rated each type of signal on effectiveness, acceptability, and unacceptability as a driver. As can be seen in Figure 5, the voice message and buzzer were selected most often as effective signals. The seat vibrator, chime, and voice message were found to be the least acceptable signals. The seat vibrator was selected by nearly half of this group as unacceptable, while nearly 25 percent thought the buzzer was unacceptable.

*Type of Signal, Driver-Passenger***:** The percentages of low-risk-based respondents who rated each type of signal on acceptability as a driver to be told that a passenger was unbelted were the following: flashing light (40.9%); buzzer (16.3%); voice (11.5%); chime (10.9%); and seat vibrator (2.1%).

Type of Signal, Passenger-Passenger: The percentages of low-risk-based respondents who selected each type of signal as the most effective for getting them to use a safety belt while they were traveling in a vehicle as a passenger were the following: seat vibrator (24.7%); buzzer (24.5%); voice (15.6%); flashing light (11.9%); and chime (9.5%). We only asked about effectiveness, because passengers are not necessarily the owners of the vehicle in which they are traveling; therefore acceptability is not an issue

Type of Signal Deliver, All Types of Systems: Figure 6 shows the percent of low-risk based

Figure 5: System Signal Preferences as a Driver Low Risk Group

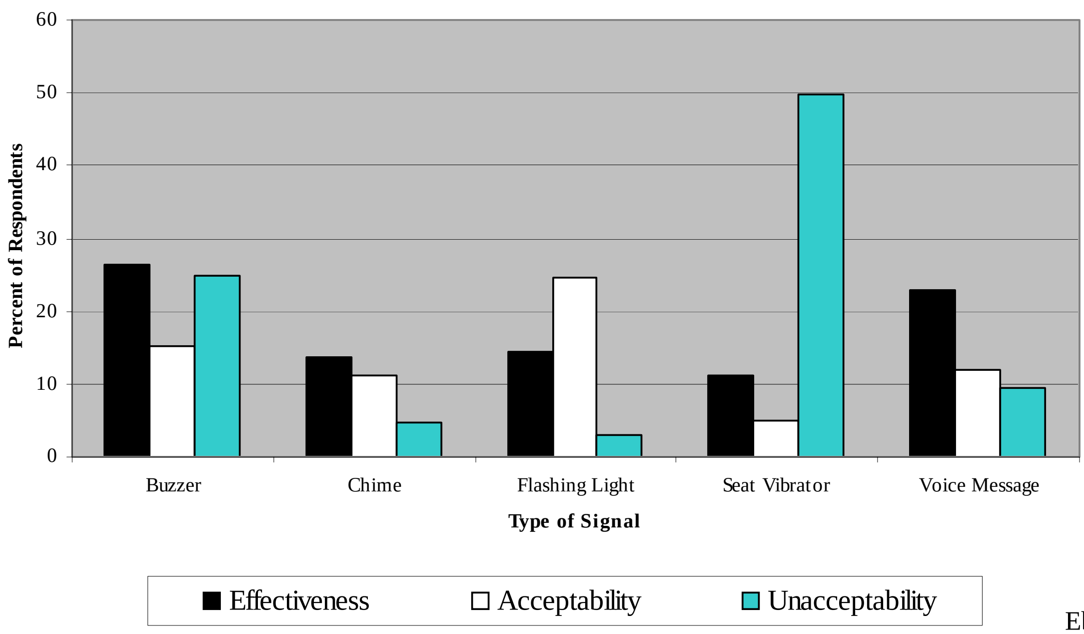
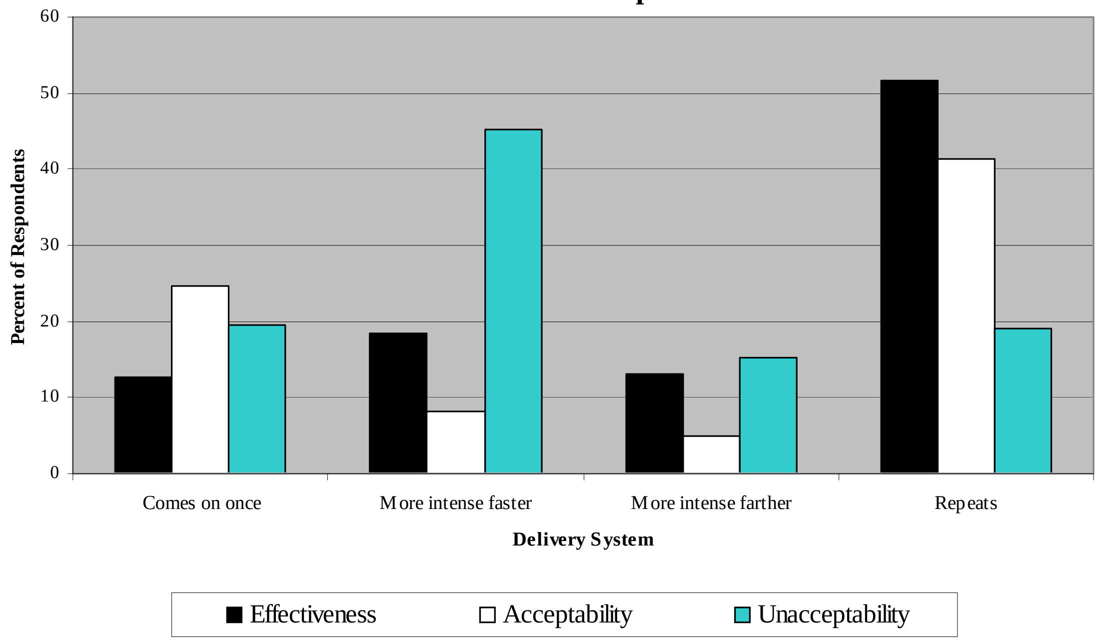


Figure 6: System Signal Delivery Preferences
Low Risk Group



respondents who selected each method for signal delivery on effectiveness, acceptability, and unacceptability. The survey did not explore this question as a function of seating position. As seen in Figure 6, repeating a signal at a constant interval was the most frequently selected delivery system for effectiveness, followed distantly by a signal that becomes more intense the faster the vehicle moves. The two least acceptable signal delivery methods were one in which the signal gets more intense the farther the vehicle travels and one in which the signal gets more intense the faster the vehicle travels. By far, the most unacceptable delivery method was one that gets more intense the faster the vehicle travels.

<u>Interlock Systems</u>

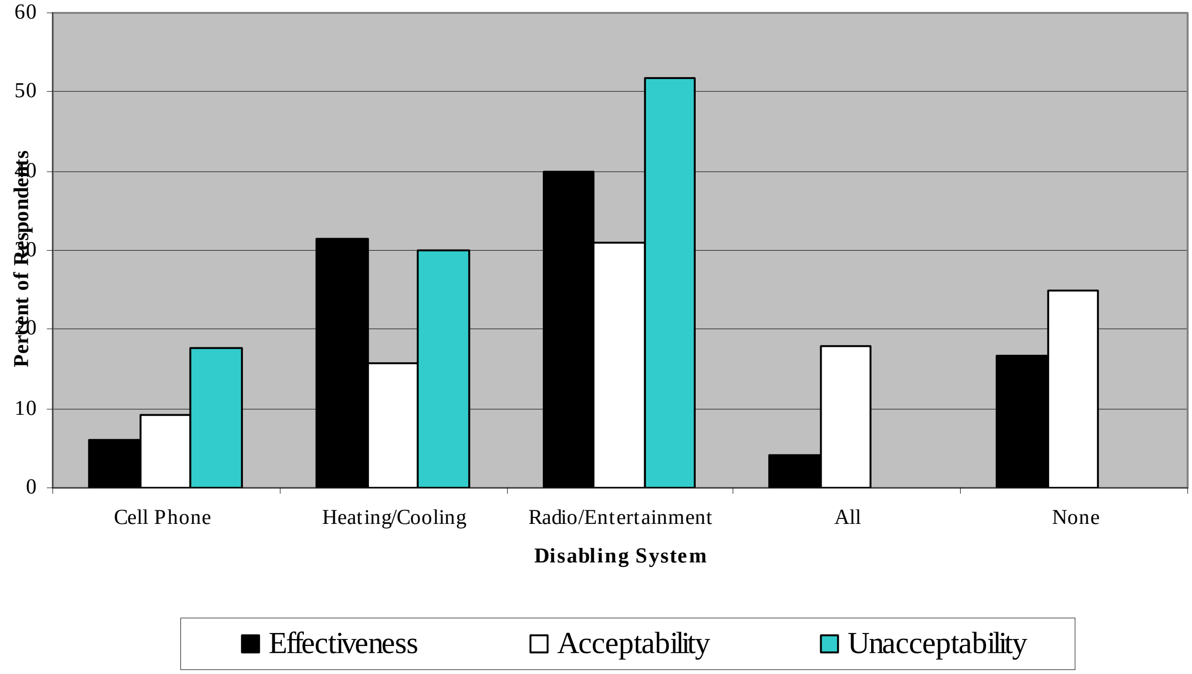
We investigated only interlocks that link to some vehicle feature other than the ignition. If a vehicle has an ignition interlock system, then no other system is necessary. The survey only considered interlock systems that would disable some system operating in the vehicle if anyone in the vehicle was not using a safety belt. Figure 7 shows the percent of respondents who selected each system to be interlocked with safety belt nonuse on effectiveness, acceptability, and unacceptability for all respondents in the survey. The survey clearly showed that disabling the radio/entertainment system was most often judged to be effective for promoting belt use and the most unacceptable system to have in the vehicle. Disabling the heating/cooling system was also judged to be fairly effective and unacceptable.

Focus Groups

Complete results of the focus groups, including illustrative quotes can be found elsewhere (Eby et al., 2004). Here we summarize the main findings.

- The main reasons cited for using a safety belt were: safety, Michigan's belt law, setting example for children in car, and belt use being a habit.
- The main reasons cited for not using a safety belt were: discomfort and inconvenience, lack of habit/forgetting, just driving short distance, and low perceived crash risk.
- The most commonly reported reasons for discomfort were: the safety belt cutting into neck, belt locking up or too tight across chest or body, roughness of belt material, tendency to wrinkle clothing, difficulty reaching buckle, and twisting of the belt.
- The following ideas for making belts more comfortable were cited: make belt out of softer material or soften belt edges and add padding to belt to cushion neck and shoulder.
- Nonuse of belts tends to be a deliberate decision rather than simply forgetting. The times when respondents were less likely to use belts were: short trips, a lack of police presence, lower

Figure 7: Disabling System Preferences
All Respondents



speeds, being in a hurry, and traveling in someone else's car or as a passenger.

- Responses about the point in the driving sequence when participants usually buckle indicate:
  - About half buckle up before starting to drive.
  - About half wait until they are actually driving to put on belt (half of this group wait until they are on patrolled roads).
  - Responses vary considerably across individuals and subgroups.
  - Participants buckle up earlier with passengers present, where there is police presence, on long trips or in unfamiliar areas, in public places with other cars, in inclement weather, and at night.
- Reactions to current US requirement (Level 1 of sample reminder system) were:
  - For most, it works only somewhat well or not at all well to get them to buckle up because of signal's short duration, ease with which it can be ignored, and low level of annoyance.
  - For majority, it is acceptable or very acceptable to have in their car.
- Reactions to Level 2 sound signals were:
  - For each signal - male voice, female voice, buzzer, and beeping signal – a majority thought it would work only somewhat well or not at all well.
  - There was a wide range of individual reactions to signals; similar reasons were often given for both liking and not liking signals.
  - The buzzer was reported to be least acceptable signal, with people voicing strong negative views.
  - The beeping signal was somewhat more acceptable than a male or female voice.
  - Acceptability was often linked to annoyance – the more annoying, the less acceptable.
  - For many, acceptability and effectiveness were inversely linked – the more acceptable, the less effective, and vice versa.
- Reactions to Level 3 sound signals were:
  - For most, the buzzer would work well or very well because of high level of annoyance associated with it.
  - The beeping signal was thought to be less effective because it was easier to ignore.
  - The majority reported that the buzzer would not be at all acceptable and the buzzer was associated with strong negative reactions.
  - The beeping signal was more acceptable than buzzer but was still thought to be only somewhat or not at all acceptable by the majority of participants.
- Reactions to a system that would alert the driver about back seat passengers' belt use were:
  - Opinions were mixed, with support generally limited to situation in which children are in the back seat.
  - The preferred signals were a flashing light and lighted diagram on dashboard to identify seating positions of unbuckled passengers.

- Reactions to a system that would alert back seat passengers directly about their own belt use were:
  - Opinions were mixed with the strongest support from the oldest age group.
  - There was a preference for the driver to remind passengers rather than to have a signal or to have diagram visible to passengers that shows the seating position of unbuckled passenger.
- Reactions to radio or entertainment center interlock system were:
  - There was general opposition to this system that was sometimes strong, with many finding the system unacceptable.
  - Concern was expressed that system would only work if people listened to the radio.
  - The oldest age group was somewhat more supportive of the system.
- Reactions to an ignition interlock system were:
  - Reactions were generally negative with many people stating that the system goes too far.
  - Concerns were expressed about how the system would work in emergency situations when driver might need to move quickly or in circumstances when belt could not be worn by someone in car.
  - Somewhat more favorable views were expressed from oldest age group, especially those who lived in an urban setting.

DISCUSSION AND CONCLUSIONS

This section contains a synthesis of the results from the literature review, telephone survey, and focus groups in order to provide guidelines for the development of an optimal in-vehicle safety belt promotion system.

Principles for Optimal System Design

Based upon previous work (Turnbell et al., 1996) and our own expertise, we derived seven principles for the development of an optimal safety belt reminder system:

1. The fulltime safety belt user should not notice the system.
2. It should be more difficult and cumbersome to cheat on the system than to use the safety belt.
3. Permanent disconnection of the system should be difficult.
4. The system should be reliable and have a long life.
5. Crash and injury risk should not be increased as a result of the system.
6. System design should be based on what is known about the effectiveness and acceptability of system types and elements.
7. System design should be compatible with the manufacturer's intended purpose/goals for the system.

Different Systems For Different Belt Users

Our results showed that the part-time belt users in the US fall into three broad, distinct categories when the reasons for part-time nonuse are considered: comfort/convenience, cognitive/personal, and low perceived risk. Full-time users, by virtue of their belt use pattern, form a fourth distinct group. Full-time nonusers, who are willing to face citations and higher injury levels in the event of a crash, form a distinct fifth belt use group. Thus, safety belt use behavior among people in different categories is motivated by different factors. We conclude, therefore, that optimal in-vehicle belt promotion technologies should target people in the different categories using different systems features and/or systems.

Level of Intrusiveness

In a recent publication by the Transportation Research Board (TRB, 2003), safety belt promotion technologies were described as varying along an intrusiveness dimension, with reminder systems at the low end of the intrusiveness scale and interlock systems at the high end of the scale. This concept,

Figure 8: Safety belt use groups aligned in order of the relative level of system intrusiveness that is most likely to change behavior.

Safety Belt Use Group	Part-time user: comfort/convenience	Full-time user	Part-time user: cognitive/personal	Part-time user: low perceived risk	Full-time nonuser
	Level of Intrusiveness	Low			High

combined with the conclusions that different users should be targeted with different features and/or systems, led us to the conclusion that the optimal in-vehicle technology should be adaptive in response to the type of belt user. A similar conclusion has been drawn by other researchers (TRB, 2003; Fildes, Fitzharris, Koppel, & Vulcan, 2002).

The conclusion that different belt use groups should be targeted with different features and/or systems and that the level of intrusiveness should be different depending upon the group, led to the development of Figure 8. The figure shows a continuum of intrusiveness, with low intrusiveness on the left and high on the right. We have placed each belt use group along the continuum, based on how we thought the intrusiveness of the system and/or features designed for each group would fall relative to each other. Note that the comfort/convenience part-time user group is not placed along the continuum. The most effective countermeasure for promoting belt use among this group is proper human factors and ergonomics research to enhance the comfort and convenience of safety belts. Low on the continuum are the full-time users, while high on the continuum are the full-time nonusers. In the middle part of the continuum, we have first placed the cognitive/personal part-time user group, followed by the low-perceived-risk group. Thus, we propose that cognitive/personal part-time users need a less intrusive system for the effective promotion of belt use than those in the low perceived risk group.

Effectiveness versus Acceptability

As previously discussed, the main thrust of the current research was to qualitatively determine which signals, signal presentation methods, and systems would be most likely to get a user to buckle up and would be acceptable to have in a vehicle. Effectiveness and acceptability, however, can be at odds with one another in belt promotion systems; that is, a highly intrusive system would be so unacceptable that even though the driver would be more likely use his or her belt to stop the annoyance, he or she would not want the system in the vehicle.

In order to maximize both effectiveness and acceptability, we developed effectiveness and acceptance criteria for each system feature and/or system to be targeted at each belt use group. These criteria are shown in Figure 9. Based upon Principle 1 for optimal system design, full time users, or those who use their belt at the start of trip, should not notice the system; that is, the system goal is that it is invisible to the full-time user. For the part-time belt users for cognitive/personal reasons, a more intrusive system is needed. The goals of this system are to maximize both user acceptance and effectiveness. Such a system corresponds to what is currently called a *safety belt reminder system*. The part-time users who cite low perceived risk as the reason for nonuse, do not need reminding, but instead need a system that provides a great enough annoyance to get people to use their belt. For lack of a better term, we have called this type of system an *annoyance system*. Because the system would be designed to be unpleasant, the system goal here is to maximize effectiveness and minimize acceptance. If this system was acceptable, then it would not be annoying enough to change behavior. Finally, we have the hard-core full-time nonusers. Despite the fact that safety belt nonuse can result in a citation and greater injury in the event of a crash, these people have made the conscious decision to not buckle up. Therefore, we believe that only the most intrusive system, an

Figure 9: Types of systems and system goals necessary for effective and acceptable in-vehicle safety belt promotion technology.

Intrusiveness →				
Safety Belt Use Group	**Full-time user**	**Part-time user: cognitive/personal**	**Part-time user: low perceived risk**	**Full-time nonuser**
System Goals	**System invisible to driver**	**Effectiveness and user acceptability are maximized**	**Effectiveness is maximized; acceptability is minimized**	**Acceptability is minimized**
Type of System Engaged	**No system engaged**	**Reminder system**	**Annoyance system**	**Interlock system**

interlock system, would be effective in getting these people to use a safety belt. As such, the system goal is simply to minimize acceptability.

Signal Type and Presentation Method

Following the framework depicted in Figure 9, the next step in developing an optimal in-vehicle belt promotion system was to determine which signals and signal presentation methods best met the system goals for each belt use group. According to the first system design principle discussed previously, if a driver uses his or her belt, the in-vehicle belt promotion technology should be invisible. Therefore, there should be no signal presented to this group. This recommendation suggests that the current 4-8 second signal that is required in US vehicles be removed.

For the cognitive/personal part-time belt use group, our survey suggested that the signals that maximized effectiveness and acceptability were a flashing light and a voice message. During the focus group discussions, however, where actual voice messages were presented, it was clear that there were strong preferences for certain voices and strong dislikes for others, and these preferences were not consistent. Having a single voice message, therefore, would be unacceptable for many users and would violate an important goal of the system for this belt use group. Many focus group participants suggested that they be allowed to input or select the voice used in this system. Since acceptance is an important criteria for this group, we extend this idea, and propose that the signal, whether it is a specific voice, light, buzzer, or chime, be selectable by the driver. The presentation method for the signal, on the other hand, must still maintain a moderate level of intrusiveness to be effective. An optimal delivery method would be selected most often by the cognitive/personal respondents as effective and acceptable, and least often as unacceptable. As seen in Figure 8, repeating at a constant interval scored high on both acceptability and effectiveness. Thus, based upon these results, we recommend that the signal delivery method for reminder systems should be one that repeats at a constant interval.

Moving along the intrusiveness continuum, the next system is the annoyance system targeted at those drivers who are part-time belt users due to low perceived risk. An optimal signal and delivery method for this group should optimize effectiveness and minimize acceptability. As shown in Figure 9, the buzzer scored fairly high on both effectiveness and unacceptability. The seat vibrator scored quite high on unacceptability but quite low on effectiveness. Based upon these survey results, the buzzer seems to be the best annoyance signal for getting a driver to buckle-up. Based on the finding in Figure 8, a signal that gets more intense the faster the vehicle travels scored high on both effectiveness and unacceptability. We conclude, therefore, that this would be the best signal delivery method for getting

Figure 10: Types of systems, system goals, signal, and signal presentation methods necessary for effective and acceptable in-vehicle safety belt promotion technology

Intrusiveness →				
Safety Belt Use Group	**Full-time user**	**Part-time user: cognitive/personal**	**Part-time user: low perceived risk**	**Full-time nonuser**
System Goals	**System invisible to driver**	**Effectiveness and user acceptability are maximized**	**Effectiveness is maximized; acceptability is minimized**	**Acceptability is minimized**
Type of System Engaged	**No system engaged**	**Reminder system**	**Annoyance system**	**Interlock system**
System Signal Type	**No signal**	**User selected**	**Buzzer**	**Shut off entertainment system**
Signal Presentation Method	**No signal**	**Repeats at a constant interval**	**Intensity increases the faster the vehicle moves**	**A warning signal prior to interlock**

the low-risk-based part-time belt user to buckle up. Note that we did not describe the characteristics of how the intensity of the signal changes. There are three options that are open for further research: increasing frequency (decreasing the inter-signal-interval); increasing volume, and increasing pitch.

The final group to target are the full-time nonusers. This group is targeted with the most intrusive system, the interlock. The system goals for the interlock, are simply to maximize unacceptability—drivers should not like having the system engage. Here we do not consider effectiveness, because these drivers will either buckle up or go to the extreme measure of disconnecting the system. Figure 7 shows that the most unacceptable vehicle system to interlock with belt use is the radio/entertainment system. This is also the system that our respondents thought would be most effective. One must be careful, however, to design this system so that the driver is not surprised and potentially distracted trying to figure out why the entertainment system is not operating. Such a situation could increase the driver's chance of crashing, violating system design Principle 5. Therefore, we propose that the optimal delivery system provide a warning signal (not determined in this study) prior to engaging the interlock, so that the driver is aware that the interlock has turned off the entertainment system. The recommended system features for all safety belt user groups are summarized in Figure 10.

An Integrated and Adaptive Reminder System

The final issue in the development of an optimal in-vehicle safety belt promotion system, is how to integrate the various systems we have discussed. We propose the adaptive system depicted in Figure 11. The figure depicts an adaptive system that changes its characteristics as the trip proceeds either in time, distance, vehicle operation, or some other metric. The figure also shows for each period of the trip, the safety belt nonuse group that is targeted by the system, that group's primary reasons for nonuse of safety belts, the system that is activated, and the important characteristics of the countermeasure. Once a trip begins, the system assumes that the driver is a full-time user and does nothing. Thus, if the driver uses his or her safety belt, then the system is invisible to them. If, however, belts are not used within some period of time or distance traveled (or other metric), then the system assumes that the unbelted driver has forgotten to use his or her safety belt. At this point, the reminder system is activated. As more time passes, or as a greater distance is traveled, if the driver still does not use his or her

Figure 11: Framework for an adaptive, integrated driver-driver in-vehicle belt promotion system

Example Metrics	Car not started 0 seconds Start of trip	Car started, not in gear < 10 mph; 4- 8 seconds	Car starts moving 11- 25 mph 2- 3 minutes	Car on patrolled roadways > 25 mph 5 minutes
Safety Belt Use Group	Full-time user	Part-time user: cognitive/personal	Part-time user: low perceived risk	Full-time nonuser
System Goals	System invisible to driver	Effectiveness and user acceptability are maximized	Effectiveness is minimized; acceptability is minimized	Acceptability is minimized
Type of System Engaged	No system engaged	Reminder system	Annoyance system	Interlock system
System Signal Type	No signal	User selected	Buzzer	Shut off entertainment system
Signal Presentation Method	No signal	Repeats at a constant interval	Intensity increases the faster the vehicle moves	A warning signal prior to interlock

safety belt, then the system assumes that the driver has chosen not to use a belt because of a low perceived risk of a crash or citation. At this point, the annoyance system is activated. Again, as more time or distance passes without the driver using his or her belt, at some point the system assumes that the driver is a full-time nonuser and an interlock system is activated, shutting off the entertainment system following the warning signal. If at any time during the trip, the buckled driver removes his or her belt, the sequence of events begins again.

The Choice of a Metric: The project did not gather definitive information about which metric is optimal or at which point along the metric the various systems should engage. We have provided three examples, based on our best judgment, the literature review, and comments from the focus group participants. In particular, during the focus groups, we discussed when during an average trip people buckle up. We developed the first metric based on how people answered this question. When choosing a metric, it is important to keep in mind the principles of optimal system development, in particular the principle that states that safety should not be compromised. The most appropriate metric or combination of metrics should be the topic of further research.

Other Reminder System Recommendations

The previous system design recommendations refer to a system designed to promote driver safety belt use (called driver-driver systems). This project, however, also investigated (in less detail) features of systems to inform the driver that a passenger is not using a safety belt (called driver-passenger systems) and to inform a passenger that he or she is not buckled (called passenger-passenger systems).

Driver-Passenger Systems: The intent of this system is to let the driver know that a passenger is not using a safety belt. In most US jurisdictions, adult passengers in a vehicle are responsible for their own belt use and will receive the citation for nonuse. Non-adult passengers, on the other hand, are the responsibility of the driver who can be cited for violating the child passenger safety law, if a non-adult does not use a proper restraint. As such, the goal of a driver-passenger system is to inform the driver of passenger nonuse of belts, so that he or she can require and monitor passenger belt use. Because the driver may not have perceived authority over an adult passenger, we conclude that a driver-passenger system should include the reminder and interlock components, but not the annoyance component of the system described in Figure 11. The signal type indicated for driver-passenger systems in the survey that maximized effectiveness and acceptability was a flashing light on the dashboard. In the focus groups, however, many participants suggested that the driver should be presented with a pictograph that shows the seating positions where passengers are not buckled. Combining these two ideas, we propose that the best signal and signal presentation method for a driver-passenger system is a seating-position pictograph that flashes at a constant interval.

Passenger-Passenger Systems: This type of system is designed to let passengers know that they are unbelted and encourages them to use their belt. As with driver-passenger systems, the passenger may be a child or adult. The large majority of focus group participants did not favor such a system, preferring that the driver tell the passenger. Therefore, as with the previous system, the annoyance system component should be omitted from a passenger-passenger system. Survey results showed that respondents thought the most effective signal for the reminder component of a passenger-passenger system would be either a buzzer or a voice message. In the focus groups, however, these signals were strongly opposed in favor of either a flashing light or no signal at all. The survey did not investigate acceptability of various passenger-passenger system components, but the focus group results suggested that the buzzer or voice would not be well received by vehicle owners. We propose, therefore, that the best signal and signal presentation method for a passenger-passenger system is a light or “unbelted” pictograph that flashes at a constant interval.

A Fully Integrated System

We have discussed three potential systems to promote safety belt use. These systems, however, would be most effective if they were integrated. Figure 12, shows the framework for a fully integrated system. This figure shows the sequence of signals, how they should be presented, and to whom, as the trip progresses. If the driver puts on his or her belt, then the sequence for the driver stops. If the passenger puts on his or her belt, then the sequence for the passenger stops. If either the driver or passenger unbuckles after having used the belt, the sequence will begin again for the person who unbuckles.

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Figure 12: Framework for a fully-integrated, adaptive in-vehicle safety belt promotion system

Example Metrics	Car not started 0 seconds Start of trip	Car started, not in gear < 10 mph; 4- 8 seconds	Car starts moving 11- 25 mph 2- 3 minutes	Car on patrolled roadways > 25 mph 5 minutes
Type of System	No system engaged	Reminder system	Annoyance system	Interlock system
Driver	No signal	If driver not belted: user-selected signal that repeats at constant interval. If passenger not belted: flashing pictograph showing seat location	If driver not belted: buzzer that increases in intensity the faster the vehicle moves. If passenger not belted: flashing pictograph showing seat location	If driver not belted: a warning signal, then entertainment interlock. If passenger not belted: flashing pictograph showing seat location
Passenger	No signal	Light or "unbelted" pictograph" that flashes at a constant interval		A warning signal followed by entertainment system interlock.

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