Analysis of Abdominal Injuries Caused by the Submarining Phenomenon in the Rear Seat Occupants

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ABSTRACT

Improvements to vehicle safety have targeted mainly the front seating positions, where the rate of seat belt usage was high and there were many casualties. Recently, rear seat occupant protection become an important challenge, with an increase in usage of seat belts by rear occupants due to new regulations and new performance criteria defined by Japanese and European vehicle assessment programs for rear seating occupants. Some prior analyses of accident data indicate that rear seat belted occupants tend to be injured in abdominal regions by the seat belt in comparison with front seat occupants. Due to this, the need to study the cause of abdominal injuries and how to countermeasure it is becoming indispensable for improving the safety of the rear seat occupants. The following two phenomenons are considered as factors which great impacts on abdominal injuries due to the seat belt: the submarining phenomenon, lap belt sliding on iliac and intruding into abdominal region, and the incorrect routing of the belt, lap belt existing initially on abdominal region. However, the relationship between these probable causes and the abdominal injuries in the real world accident is not expressly described in prior studies. Therefore, first, the frequency of the abdominal injuries caused by the submarining phenomenon was estimated by micro analysis of the accident data. Second, the influence on abdominal internal organs, to which the lap belt load was applied, was analyzed using human body FE model THUMS. The results of this analysis indicated that serious effect might be applied to abdominal internal organs. As the routing of the lap belt on the pelvis was shown as being very important in this study, a parametric study using Madymo was conducted to determine additional factors that might influence the proper routing of the belt on pelvis. This study narrowed down the factors with big contribution and explains how they were determined.

INTRODUCTION

Crash data analysis clearly shows that seat belt use has an unquestionable advantage in occupant protection. A 2003 NHTSA research [1] estimated that 147,246 occupant lives were saved by seat belt use through 1975 to 2001 in US. The improvements of restraint systems such as seat belt and airbag has been pushed forward by changes in regulation and assessment programs mainly for the front seating positions, where the rate of seat belt usage was high and there were many casualties. In addition, requirements for improving the protection of rear seat occupants are increasing recently. In Japan New Car Assessment Program (JNCAP), off-set deformable barrier (ODB) test with Hybrid III 5%ile female dummy in rear seat started in 2009, while seat belt wearing for rear seat occupants was
made mandatory by regulation in 2008 and usage rate is rising. [2][3] In addition, European New Car Assessment Programme (ENCAP) is conducting frontal full width rigid barrier (FWRB) test with Hybrid III 5%ile female dummy in rear seat starting with January 2015. Due to the above, rear seat occupant protection is being regarded as more and more important.

R.Frampton et al. [4] reported that the abdominal injury risk for the rear seat occupants was higher than for front seat occupants based on UK accident data analysis. Additionally, the report on injuries to older passengers by the Institute for Traffic Accident Research and Data Analysis (ITARDA) [5] indicated that the rate of fatal and serious injuries for rear occupants involving elderly is higher due to abdominal injuries. The seat belt load is considered as one of the main causes for abdominal injuries. The following two phenomenons are considered as factors which greatly influence an abdominal injury due to the seat belt: 1) submarining phenomenon, lap belt sliding on iliac and intruding into abdominal region, 2) belt malpositioning, lap belt existing initially on abdominal region. In particular, JNCAP and ENCAP put focus on the submarining phenomenon; the score is reduced when submarining occurs in the above-mentioned ODB (JNCAP) or FWRB (ENCAP) evaluations.[6][7]

However the above-mentioned accident data analysis [4][5] does not clearly refer to the relation between abdominal injuries and submarining phenomenon by lap belt. An in-depth analysis about the abdominal organs injuries of belted / non-belted front occupants by K. Ono et al.[8] also did not report any relationship. Due to this, the first objective of this study is to estimate the occurrence frequency of the submarining phenomenon based on accident data analysis and to simulate the severity to the abdominal organs when the submarining phenomenon occurs using the human finite element model. Second objective is to examine the effective control scenario to reduce the submarining phenomenon, because of the unquestionable effect of the lap belt force on abdominal organs injuries.

METHODS

Crash Data Analysis

The National Automotive Sampling System-Crashworthiness Data System (NASS-CDS) database was used to estimate the frequency of the submarining phenomenon occurrence. In the analysis, 555 injury cases of belted rear seat occupants were extracted from 2007 to 2011 (PDOF 11-1 o’clock).

Analysis of the influence on abdominal organs

A series of simulations using 50%ile male human finite element model, THUMS Version3, was conducted to examine the influence to abdominal organs by lap belt force when the submarining phenomenon occurs. The severity of pressure to the abdominal organs due to the position of the lap belt, which assumed submarining phenomenon and belt malpositioning, was simulated and compared with the output of the lap belt worn appropriately. In this study, the geometry, seat belt parameters and deceleration of 56km/h full width rigid barrier test of actual vehicle were used. Abdominal organs deflection and deflection velocity were considered as characteristic indexes indicating the severity. However, the deflection mode and the injury mechanism of the abdominal organs were different because variation in structure (such as the solid and hollow). The characteristic value suitable for each internal organ was not expressly shown as various studies on those injury mechanism might be ongoing. Therefore, this study adopted both the deflection and the deflection velocity as the characteristic index.

Parametric Study for Submarining Performance

A parametric study using Madymo was conducted to narrow down the parameters with large impact to anti-submarining performance and to clarify their proper design values. The angle between the belt-to-pelvis (abbreviated BTP from now on) from the submarining phenomenon index used in the study of J. Horsch and W. Hering [9] was used in this analysis. (Figure 1)
The mechanism of the submarining phenomenon was based on the following hypothesis:  
“The pelvis receives the torso internal force $F_{\text{Chest}}$, the femur internal force $F_{\text{Femur}}$, the resistance force from the seat cushion $F_{\text{Seat}}$ as well as the lap belt force $F_{\text{Belt}}$ and moves forward with rotation. Due to this displacement of the pelvis, the lap belt direction for the iliac spine changes. As this displacement increases, the BTP increases counterclockwise and the force of the upper direction along the iliac spine increases. Due to the change in this load distribution, it becomes very likely that the submarining phenomenon will occur.”

According to this hypothesis, “continuously maintaining a small BTP” is important to control the submarining phenomenon. Therefore, the extraction of the factors that influence the change of lap belt angle $\theta_B$ and iliac perpendicular angle $\theta_P$, and the contribution degree of each factor were derived.

$$\Rightarrow \text{BTP} = \theta_P - \theta_B$$

RESULTS

Crash Data Analysis

According to the distribution ratio of the Abbreviated Injury Scale (AIS) from the extracted cases, 89% of the injuries in the rear seat were minor injuries with an AIS 1; the reduced severity of the injuries was attributed to the use of restraint system such as seat belt. This study researched the remaining 11% of AIS 2 and greater (AIS2+).

Injury source for the AIS2+ injuries in the rear seat (Figure 2a), injury description of the injuries by belt restraint system (Figure 2b), the rate of the injured abdominal organs (Figure 2c) are illustrated in below Figure 2. Results indicate that 28% of the injuries in the rear seat were caused by the belt restraint system, of which 34% were injuries to the abdominal region. In addition, 65% of the injured abdominal organs were distributed in the lower abdomen such as the intestine or the mesentery and it was confirmed that the main cause of the abdominal injuries were influenced by the lap belt force.

$$\begin{align*}
F_{\text{Belt}} & : \text{lap belt force} \\
F_{\text{Chest}} & : \text{torso internal force} \\
F_{\text{Femur}} & : \text{femur internal force} \\
F_{\text{Seat}} & : \text{resistance from seat cushion} \\
\theta_B & : \text{lap belt angle for reference plane} \\
\theta_P & : \text{iliac perpendicular angle for reference plane}
\end{align*}$$
Here, the micro-analysis of abdominal organs injury cases was researched to determine the relation between abdominal injuries and the submarining phenomenon. Results indicate that 30% of the abdominal organs injury cases were received abrasion and contusion to the hips by the belt restraint system. It was confirmed that the lap belt was fitted on the iliac spine at the start of the crash event and it became very likely for the submarining phenomenon to occur during the collision.

In four cases, rear seating occupants sustained the pelvic fractures with an AIS2 caused by the belt restraint system. After further analyzing these four cases, three were specified the fracture point and one of them was fracture of the anterior superior iliac spine on which the lap belt was fitted on; however, for this case, the cause of the fracture is believed to not be due to high load from the lap belt due to a low barrier equivalent speed of 45km/h. Two of the fractures were pelvic breaking away from iliac spine. From the collision velocity and involved region, it was assumed that the fractures were partly due to a contact with the buckle, the seat, another interior part or an unbalanced load by the vehicle behavior.

**Analysis of the influence on abdominal organs**

Vertical section views (in the initial stage of the restraint, in the maximum movement of the pelvis, and in the rebound of the pelvis) of the lap belt fitting on the iliac spine properly or the abdomen, maximum amount of abdomen deflection, and deflection velocity changes by time are illustrated in Figure 3. Here, the amount of abdomen deflection and deflection velocity were calculated using two places, top and bottom, that assume a division between spine and organs shown in vertical section views. This THUMS model used the abdominal organs model which is united; the upper part is equivalent to the upper abdomen including the liver and the spleen, and the lower part is equivalent to the lower abdomen including the intestine and the mesentery. Results indicate that the amount of abdomen deflection and deflection velocity in fitting of the lap belt on the abdomen were higher in the lower measurement part, equivalent to the lower abdomen, which has seen increased injury in the accident analysis compared to the upper part. This result matched the actual accident situation in the real-world. The maximum amount of abdomen deflection in fitting of the lap belt on the abdomen at the lower measurement part was increased by approximately 3.4 times compared with fitting of the lap belt on the iliac spine properly. Additionally, abdomen deflection velocity was increased by approximately 2.3 times. The above results showed that, when the lap belt was fitted on the abdomen, the characteristic value of abdominal injuries increased greatly. This confirms the uttermost importance of properly routing the lap belt on the iliac spine in order to restrain the occupants properly.

*Figure 3.* Vertical section views of fitting the lap belt on the iliac spine properly or the abdomen (Figure 3a), maximum amount of abdomen deflection (Figure 3b), time change of deflection velocity (Figure 3c).
Parameter Study of the Anti-Submarining phenomenon

The above analysis of abdominal injuries showed that the submarining phenomenon may cause a significant load to the abdomen and it was also confirmed, by simulation, the importance of fitting the lap belt on the iliac spine. Therefore, it is essential to improve the anti-submarining performance as much as possible.

At the beginning of the study, the factors impacting the lap belt angle $\theta_B$ and the iliac perpendicular angle $\theta_P$ were selected as shown in Table 1 - seat belt specification, inner anchor point, seat cushion characteristic and front seat position. In addition, the levels were set in realistic range applied to actual vehicle and provided in regulation. Hybrid III 5%ile female dummy model was used as this was considered to be worst case for the submarining phenomenon because of its small pelvis. The initial sitting position and posture were neutral and the lap belt was fitted on the iliac spine.

Figure 4 shows the contribution degree to BTP of each factor listed in Table 1. All factors had some contribution degree to BTP, and it was confirmed to reduce the submarining phenomenon by proper parameter settings. In particular, it was determined that the contribution degree of the inner anchor point (longitudinal) and the seat cushion characteristic (seat cushion airbag Yes / No) were particularly large.

Figure 5 shows the amount of BTP change ($\delta$BTP) during the crash event by changing the parameters of these two main factors. This shows that higher anti-submarining performance is achieved by following two items: ensure the initial relative angle by the front configuration of the inner anchor point, and reduce the relative angle change by the effect of the seat cushion airbag controlling the forward and downward pelvis displacement. In addition, the amount of BTP change ($\delta$BTP) was provided as the difference between the iliac perpendicular angle change ($\delta\theta_P$) and the lap belt angle change ($\delta\theta_B$); however, the change of $\delta\theta_P$ was small and the change in $\delta\theta_B$ was big. This confirmed that, in order to prevent the submarining phenomenon, the forward configuration of the inner anchor point, to make the initial BTP smaller, and more so the pelvis restraint countermeasure to reduce the forward and downward displacement during crash event were important.

As for the specific countermeasure to enable compatibility of the front configuration of the inner anchor point and the reduction of the pelvis displacement, the seat cushion airbag already mentioned above is one option, another option being a seat belt device to offer pre-tension to the lap belt effectively.

<table>
<thead>
<tr>
<th>Table1. Factor and Level</th>
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<tbody>
<tr>
<td><strong>Seat Resistance force</strong></td>
</tr>
<tr>
<td>A Seat cushion airbag Y/N</td>
</tr>
<tr>
<td>B Seat cushion airbag position</td>
</tr>
<tr>
<td><strong>Torso internal force</strong></td>
</tr>
<tr>
<td><strong>Femur Internal force</strong></td>
</tr>
<tr>
<td><strong>Lap belt force</strong></td>
</tr>
<tr>
<td>F Inner anchor point (crosswise)</td>
</tr>
<tr>
<td>G Inner length</td>
</tr>
<tr>
<td>H Lap force limiter</td>
</tr>
</tbody>
</table>

Figure 4. Cause and effect diagram
DISCUSSION

The accident analysis indicated that 65% of the injured abdominal organs were distributed in the lower abdomen such as the intestine or the mesentery. In addition, the micro-analysis showed that the submarining phenomenon might have occurred in about 30% of abdominal injured occupants. According to the simulation used FEM, it was confirmed that the lap belt positioned on the abdomen might be the cause for the large amount of deflection and deflection velocity, particularly in the lower abdomen and that, the main cause of abdominal injuries was influenced by the lap belt force. The conclusion, based on the above, is that it is important to hold the lap belt on the pelvis in order to reduce the load on the abdomen and prevent abdominal organs injuries. Although the combination of the anterior arrangement of inner anchor point and the measures for reducing the forward and downward movement of the pelvis described above are effective, the anterior arrangement is likely to increase the forward displacement of the pelvis by restraint performance degradation of the lap belt. Due to this, changes to the BTP should be considered but attention should be paid to the contradictory relationship with forward displacement suppression of the pelvis. However, in this analysis conditions, the effect of the initial BTP was larger than the BTP change during crash event and the most anterior arrangement, within the limit of regulation, was the optimum layout. Additionally, the decrease of pelvis restraint performance leads to the increase of the inner belt load and may have an influence to thoracic deflection. Therefore, it is necessary to design the proper location of the inner anchor in accordance with the geometry of vehicles or the crash characteristic.
LIMITATION

The above study of submarining phenomenon was intended for the occupant sitting in proper position and with proper posture, but there may be various sitting conditions in the real-world. Research of sitting position and posture in the rear seat of sedan and minivan was conducted using 50 subjects (age; 24-51, gender; 76% male / 24% female). The results showed that the hip of 46% subjects were positioned in excess of 20mm forward compared to proper position and 15% were positioned in excess of 60mm. This result suggests that the study of submarining phenomenon in consideration of real-world sitting position should be conducted as a future challenge. Therefore, not only improvement of the vehicle layout to facilitate placing the lap belt in the correct position on the pelvis are needed, but it is also necessary to promote education for understanding the importance of the correct use of seat belts.

As mentioned during the results of the accident analysis, the pelvic fractures are not only due to the high lap belt load, but also due to contact with the hard interior or unbalanced load by the vehicle behavior. However, the investigation of McCalden et al. [11] indicate that the breaking strain of the femur cortical bone decreases with age. Since the similar reduction can be considered against the pelvis, when implementing the measures to increase the pelvis restraint force to prevent the submarining phenomenon, it is necessary to control of the seat belt force in consideration of elderly.

CONCLUSIONS

The investigation into abdominal injuries due to the submarining phenomenon and the measurements from this study were carried out with the aim to further improve the safety of the rear seat occupant. The findings of this study show that:

1. Accident analysis indicated that 65% of the injured abdominal organs were distributed in the lower abdomen such as the intestine or the mesentery. In addition, the micro-analysis showed that the submarining phenomenon might have occurred in about 30% of abdominal injured occupants.

2. Simulation using human finite element model found that fitting the lap belt on the abdomen might cause serious abdominal injuries as the maximum amount of abdomen deflection was increased by approximately 3.4 times, and abdomen deflection velocity was increased approximately by 2.3 times relative to properly fitting the lap belt on the iliac spine. This confirmed that it is very important to route the lap belt on the pelvis in order to reduce the load on the abdomen and prevent abdominal organs injuries.

3. To prevent the submarining phenomenon, the forward configuration of the inner anchor point, to make the initial BTP smaller, and the countermeasure of increasing the restraint force of the pelvis to reduce the BTP change during crash event are valid.

4. The forward configuration of the inner anchor point can cause side-effects which impacts the lap belt angle during crash event or influence other injury values. Therefore, when designing the layout of the inner anchor point, the confirmation and the optimization based on the geometry and the crash characteristic of the target vehicle should be considered. And, in case of applying countermeasures to increase the restraint force of the pelvis, it is necessary to consider the pelvic tolerance of vulnerable population, such as elderly.

REFERENCES

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ADVANCED SEAT BELT REMINDER SYSTEM FOR REAR SEAT PASSENGERS

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ABSTRACT

Advanced seat belt reminder systems with audiovisual warnings have proven to be highly effective in increasing the belt wearing rates of a vehicle's front seat occupants. While the availability of such advanced SBR systems for the front seats is almost common in some markets and fast-growing in others, also thanks to NCAP incentives, the systems available on the rear seats have so far only offered a basic functionality. In 2014, an upgraded SBR function entered the mass market, and the world's first car with an advanced rear seat SBR system including occupant detection was launched on the Japanese market. This vehicle, the Subaru LEVORG, offers an advanced audiovisual SBR warning for the rear outboard seating positions. This advanced function is enabled by occupant detection sensors designed to detect human rear seat occupants, while being robust against the detection of child restraint systems (CRS) or other objects frequently transported on a vehicle's rear seats. The robustness of the occupant detection and the object non-detection has been tested extensively. Occupants shifted their position forward and laterally away from the nominal seating position. A multitude of CRSs and objects were tested to ensure that they do not trigger unnecessary warnings. Advanced rear seat SBR systems have the potential to significantly increase the belt wearing rates, especially as those tend to be much lower on the rear than on the front seats in almost all countries. As belt load limiters and belt tensioners are more and more available for the rear seats, the advanced SBR systems ensure that more rear seat occupants will benefit from the restraint system enhancements.

INTRODUCTION

Seat belts have proven to be highly effective in reducing the likelihood of severe or fatal occupant injuries in vehicle collisions. Additional technologies like seat belt tensioners and load limiters have helped to improve the seat belt effectiveness and to reduce belt induced injuries to the chest area. Many people, however, do not buckle up, for various reasons, often simply forgetting about it. Seat belt reminder (SBR) systems with audiovisual warnings have proven to be highly effective in increasing the seat belt use. The number of unbelted drivers is reduced by 80% in vehicles with advanced SBR systems meeting the Euro NCAP requirements [1]. For the front seat passengers the reminder effectiveness is comparable [2]. As seat belt reminders have such a significant impact on the belt wearing rates, the large majority of NCAP programs have decided to introduce incentives for front seat SBR systems into their rating. These incentives were very successful in motivating the vehicle manufacturers worldwide to fit SBRs in an increasing number of vehicle models [3]. In addition to the front seat SBR systems with audiovisual warnings, more simple systems had been developed for the rear seats, providing the driver with visual information on the buckle status on the rear seats. However, the effectiveness of those simple systems is limited as they are highly dependent on the driver response to the information. In 2014, a first car with an advanced seat belt reminder system also providing an audiovisual warning to the rear seat occupants entered the Japanese market. This paper describes the motivation behind this development, as well as the challenges that had to be solved with regards to occupant detection on the rear seats.

MOTIVATION FOR ADVANCED REAR SEAT SBR

Subaru's roots go back to an aircraft manufacturer, so safety is one of the company's core values. In the domain of active safety, Subaru has proven this philosophy with its award-winning EyeSight technology, which was the first system ever to use only stereo camera technology to support functionalities like Adaptive Cruise Control, Lane Departure Warning and Autonomous Emergency Braking. But also in the area of passive safety, Subaru identified additional road safety potential, aiming to reduce the number of vehicle occupant fatalities, namely by increasing the seat belt wearing rates on the rear seats. Although belt usage on rear seats has been mandatory since 2008, the rear belt wearing rates tend to be low in Japan, resulting in easily preventable occupant injuries and fatalities. Advanced seat belt reminder systems have
proven to be effective in raising the belt wearing rates on the front seats, but no such system had ever been implemented on a vehicle's rear seats. One key component for such a system, a rear seat occupant detection sensor simply did not yet exist.

In a joint development effort, Subaru and sensing system specialist IEE created the world's first advanced rear seat SBR system for a production vehicle, the Subaru LEVORG, launched in 2014. The expectation is that the system will increase the belt wearing rates, thus reducing the number of injuries or fatalities in Subaru vehicles.

Rear Seat Belt Wearing Rates

Seat belt wearing rates on the rear seats are lower than those for the front seats in all countries for which data is available. The reasons for this difference in belt usage behaviour are manifold, possible contributing factors are:
- rear seat occupants feel safer because of the backrest in front of them
- belt usage on the rear seats was mandated much later than for the front seats, so fewer people have acquired the habit to use the seat belt on the rear bench
- a lower enforcement level by police, also because belt usage is more difficult to verify
- unavailable or less effective seat belt reminders

**Seat belt wearing data from Japan** for front and rear seat vehicle occupants is shown in Figure 1 for the time frame 2005 to 2014. It shows the data for public highways (cities and rural roads). Additional data had been collected for express highways [4]. The belt wearing rates are highest for the driver (driver SBR fitment has been mandatory in Japan since 2005), closely followed by the front passenger. Belt wearing rates for the rear seat occupants are much lower, only about 1/3 (35.1%) of the rear passengers buckle up on public highways. On express highways the belt usage increases to 70.3%, but is still far below the front seat usage rates. Seat belt usage on the rear seats was made mandatory in 2008, which explains the significant increase in the belt wearing rate for that year.

![Figure 1. Seat belt wearing rates in Japan on public highways.](image)

**Rear seat belt usage in the US** [5] is also lower than for the front seats, as shown in Figure 2. However, the difference is less important than in Japan. At 75%, the rear seat belt wearing rate in the US is only about 10% lower than the one for the front seats, while in Japan the rear seat usage rate is about 60% lower compared to the front seats. However, it should be noted that front seat belt usage in Japan (driver 98%, front passenger 94%) is about 10% higher than in the US (86%).

The US data also allows the analysis of rear seat belt usage by age group. The lowest belt wearing rate can be found for the age group teenagers and young adults (age 16 – 24), where only 67% buckle up, compared to the overall average of 75% belt users. The highest belt use can be found for children aged 8 to 15 (83%) and occupants aged 70 and higher (80%).
In Europe, large differences in rear seat belt usage can be found when comparing the different countries [6]. While the belt wearing rates of the rear seat passengers tend to be high with more than 80% for the Western and Northern European countries, much lower belt use is observed in most Eastern and Southern European countries.

**Table 1. Front and rear seat belt use rates for a selection of European countries.**

<table>
<thead>
<tr>
<th>Country</th>
<th>Belt use - front seat</th>
<th>Belt use - rear seat</th>
</tr>
</thead>
<tbody>
<tr>
<td>Austria</td>
<td>89%</td>
<td>75%</td>
</tr>
<tr>
<td>Belgium</td>
<td>86%</td>
<td>80%</td>
</tr>
<tr>
<td>Czech Republic</td>
<td>97%</td>
<td>66%</td>
</tr>
<tr>
<td>France</td>
<td>98%</td>
<td>84%</td>
</tr>
<tr>
<td>Germany</td>
<td>98%</td>
<td>98%</td>
</tr>
<tr>
<td>Greece</td>
<td>71%</td>
<td>21%</td>
</tr>
<tr>
<td>Italy</td>
<td>60%</td>
<td>50%</td>
</tr>
<tr>
<td>Poland</td>
<td>80%</td>
<td>43%</td>
</tr>
<tr>
<td>Spain</td>
<td>91%</td>
<td>81%</td>
</tr>
<tr>
<td>UK</td>
<td>95%</td>
<td>89%</td>
</tr>
</tbody>
</table>

In Korea, belt usage on the rear seats is significantly lower than on the front seats [7]. Only 19% of the rear seat occupants are belted, versus 84% of the front seat occupants.

**Rear Seat SBR Effectiveness**

The simple monitoring of the rear seat belt buckle status only allows for visual information to the driver and optionally the rear seat passengers at vehicle start. A brief audible warning can only be triggered if there is a "change of status", i.e. if a belted rear seat occupant unbuckles during the trip. The lack of a continuous audible alert limits the effectiveness of those simple systems.

Very little data is available on the effectiveness of such SBR systems. In a comment to NHTSA in 2010 [8], Volvo stated: "...Volvo surveyed Volvo owners in Sweden and Italy in 2005. The survey clearly demonstrated that the belt usage rate in the rear seat, with the monitoring system as compared to without belt reminders, had increased from around 60% to around 82%". This would correspond to a reminder effectiveness of approximately 50%.

A laboratory study was conducted in Japan in 2012 [9], comparing the effect of various optical and audible SBR warnings on the belt use of rear seat passengers. Table 2 summarises the most important study results. The initial belt wearing rate without SBR warning was 38%. When an optical warning was only presented to the driver, who then reminded the rear seat passengers, the belt use increased to 56%. When both, driver and rear seat passengers were presented with an optical warning, the usage rose to 72%. And when an audiovisual
warning was used, 97% of the rear seat passengers buckled-up. So audiovisual SBR warnings motivated up to 95% of the initially non-belted rear seat occupants to buckle up. For visual-only warnings the effectiveness was limited to 50% (in line with the Volvo data above).

### Table 2. Belt wearing rates for various SBR warning systems.

<table>
<thead>
<tr>
<th>Rear seat passenger information</th>
<th>No SBR information</th>
<th>Ceiling icon, blinking with frequency change, no audible signal</th>
<th>Ceiling icon, blinking with frequency change, audible signal with frequency change</th>
</tr>
</thead>
<tbody>
<tr>
<td>Driver information</td>
<td>No SBR information</td>
<td>38 %</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>Meter cluster icon, blinking with frequency change, no audible signal</td>
<td>56 %</td>
<td>72 %</td>
</tr>
<tr>
<td></td>
<td>Meter cluster icon, blinking with frequency change, audible signal with frequency change</td>
<td>-</td>
<td>97 %</td>
</tr>
</tbody>
</table>

As the first vehicle with an advanced rear seat SBR system only entered the market in Japan in June 2014, no field-data is available with regards to its effectiveness in increasing the belt wearing rates. But the laboratory study indicates a clear trend with regards to the effectiveness of various warning strategies.

### OCCUPANT DETECTION SENSOR DEVELOPMENT

Occupant detection on the rear seat can be achieved in principle in a similar way as on the front seat, a foil-based pressure sensitive sensor, integrated between seat foam and trim, is activated by the occupant's weight. However, some rear seat peculiarities have to be taken into consideration. The rear bench is often used to transport various objects, child restraint systems (CRS) are predominantly installed there, and the backrest can be folded down. For those scenarios sensor activation has to be prevented. In addition, the occupant himself often has a higher freedom of movement on the rear seat compared with the front seat, due to missing or less distinct side bolsters. Therefore sensor design and size have to be adapted to the specific rear bench needs.

**Figure 3. Top view onto rear bench with occupant detection sensors on outboard positions.**

A dedicated test matrix has been developed to ensure robust sensor performance for occupant detection and object non-detection. Typically occupant detection has to be guaranteed for a 5% female, but also smaller occupants like young teenagers can be taken into consideration. Occupancy detection tests are performed with occupants of the specified size and weight. In addition to the nominal seating position, testing includes some forward and lateral position shifts. Non-detection is among others tested with beverage packs, rice and potato bags and a multitude of child restraint systems. In particular ISOFIX CRS with an integrated harness should not actuate the sensor, as those don't require the 3-point seat belt of the car to fix the CRS or to secure the child. Another non-detection test puts some weight onto the folded backrest to simulate a heavy trunk load.
A rear seat specific sensor layout and an IEE patented interconnection of the sensor's pressure sensitive cells allows the differentiation of the pressure profiles typically generated by humans from those generated by CRS or other test matrix objects. Figure 4 shows pressure profiles of a 5% female and various CRS, recorded with a high resolution pressure sensitive mat on a front passenger seat. The pressure distribution looks similar on the outboard rear seats. Although CRS or other objects can also exercise some load on the area usually covered by a human buttock, a smart sensor design can almost entirely exclude unnecessary SBR warnings. For objects that are heavy enough to nevertheless activate the sensor, it is recommended to secure them with the belt or to load them into the trunk, as otherwise they are a potential danger for vehicle occupants if there is a crash.

Figure 4. High resolution pressure profiles of human and CRS on a vehicle seat.

The system integrated into the Subaru model "LEVORG" has occupant detection only on the outboard seating positions. A system covering three positions on the rear bench is under development in order to cover all seating positions with an advanced seat belt reminder function.

The current system has the sensors and buckles connected to the car's wire harness via cables and connectors. For vehicles with highly flexible seat configurations or removable seats, a wired system layout could be considered a limiting factor. Therefore a wireless prototype concept has been developed by IEE to address those concerns. It is based on the same communication technology as currently used by tire pressure monitoring or keyless-go systems. A serial feasibility evaluation for the wireless system, as well as other occupant detection technologies that could be used for rear seat passenger detection, is currently under investigation.

EXISTING AND FUTURE NCAP INCENTIVES

NCAP star ratings for a vehicle only have real-life relevance if occupants are belted during a collision. A five star car can only provide a "five star protection" if the occupants are buckled-up. That was the motivation for many NCAP programs to promote effective seat belt reminder systems, with a focus for the front seats. Several NCAP programs have now started to perform crash tests with adult dummies on the rear seats. One aim is to motivate the vehicle manufacturers to make restraint system technology that's widely available for the front seats, like belt tensioners and load limiters, also available on the rear seats in a larger number of vehicle models. However, as for the front seats, the rear seat occupants can only benefit from those improved belt systems if they are buckled up. Hence the NCAP programs have an increasing interest to promote more efficient SBR systems for the rear seat, especially taking into consideration the generally lower belt wearing rates on the rear compared to the front seats.

Japan NCAP

When Japan NCAP introduced an overall rating scheme in 2011, SBR points became part of the evaluation. Since then, the overall rating score has been based on the sum of three elements: occupant protection (up to 100 points), pedestrian protection (up to 100 points) and seat belt reminder (up to four points for the front passenger seat and up to four points for the rear seats) [10].

J-NCAP was the first NCAP program to create an incentive for advanced seat belt reminders on the rear seats. Simple buckle monitoring only systems limited to telltale/display-type information are awarded with a maximum of two points, with the score depending on display location and its visibility to the occupants. Two additional points can be scored if the rear SBR alert includes an audible warning of at least 30 seconds. Such a warning, however, can only be triggered if passenger presence information is available.

The Subaru LEVORG is the first car where such an advanced SBR functionality will be assessed for the rear outboard seating positions, and it is expected to score between 3.0 and 3.33 points for the rear SBR system (official results not yet published at paper deadline).
Euro NCAP
Euro NCAP was the first NCAP to introduce SBR bonus points in 2002. Their SBR protocols evolved over time, and currently two combined points are available for advanced SBR systems covering both front seats, and one point for the buckle monitoring variant on the rear seats. The Euro NCAP protocol recommends occupant detection on the rear seats, but does so far not require it. In its ”2020 Roadmap” [11] Euro NCAP announced to introduce incentives for advanced rear seat SBR systems in 2018. Out of 2 points available for rear seat SBR, 1.5 points will be available for the buckle monitoring function (all rear seats), and 0.5 point will be allocated to additional occupant detection covering the 2nd row outboard seating positions, enabling an advanced reminder function.

Australasia NCAP
Australasia NCAP has announced it will fully harmonise with the Euro NCAP rating from 2018 on, so advanced rear seat SBR systems will become rating relevant in Australasia NCAP too.

Other NCAPs
Some NCAPs are now about to introduce incentives for the simple rear SBR systems into their rating (Korea NCAP in 2015, ASEAN NCAP in 2017, Latin NCAP – year to be confirmed). It can be assumed that incentives for more advanced systems will follow a couple of years later.

CONCLUSIONS AND RECOMMENDATIONS

The relatively simple rear seat SBR systems so far used in cars, warning only via telltale or text message, have a limited effectiveness on increasing the belt wearing rate. Now the time has come to extend the concept of advanced SBRs to the rear seats and to address the issue of occupant detection in an environment with a higher variability than on the front seats.

Driven by Subaru’s safety strategy and Japan NCAP incentives, a first vehicle model with an advanced rear seat SBR system has entered the Japanese market. Occupant detection sensors, dealing with the specific needs of the rear seat environment have been developed by IEE.

Although field data on the effectiveness of an advanced rear seat SBR system is not yet available, a laboratory study on various rear seat SBR variants and the proven effectiveness for front seat occupants raise the expectation that rear seat belt wearing rates, typically much lower than those for the front seats, can be increased significantly.

And with NCAPs worldwide increasingly addressing the safety of rear seat occupants, it makes sense that they also create incentives for systems that ensure high belt wearing rates for those occupants. Euro NCAP and Australia NCAP will follow Japan NCAP, and start rewarding advanced rear seat SBR systems from 2018 on. By achieving higher belt wearing rates in combination with improved rear seat restraint systems one can expect to achieve additional road safety benefits in the future.

REFERENCES


INFLUENCE OF IMPACT TYPE AND RERAINT SYSTEM TRIGGERING TIME ON INJURY SEVERITY IN FRONTAL IMPACT CRASHES

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ABSTRACT

The safety performance of cars is evaluated using standard tests. These standard tests are normally performed with full overlap or 40% overlap from the corner in different speed ranges. Analysis of accident data indicated that the injury severity of car occupants that were involved in accidents that are different compared to the standard tests (e.g., central pole impact) is considerably higher than for those that are similar to the standard tests. One of the discussed possible reasons for this observation is that the restraint system triggering might not be appropriate for these situations.

The combination of NASS CDS data with the NASS EDR data allows to analyse the accident circumstances, the restraint system triggering times and the injury situation in frontal impact accidents. The result of this analysis is a grouping of accident situations with corresponding injury severities and restraint system triggering times. These groups are rechecked using the GIDAS data to confirm the influence of the accident circumstances on the injury severity, as restraint system triggering time is not available in the GIDAS data sample.

The restraint system trigger time depends on several factors (e.g., delta-v, impact configuration (e.g., involving both long members, only one long member or no long member), impact angle etc.). While most of the differences appear to be sensible for optimal protection (e.g., at higher delta-v the airbag is needed earlier) the differences for the different impact configuration appears to be critical with respect to injury severity levels of the frontal occupants.

The shown correlation between crash configuration, restraint system triggering time and injury severity does not necessarily mean that there is a causative relation between triggering delay and increased injury severity. However, it is likely that there is a causative relation.

INTRODUCTION

For the protection of car occupants the design of the vehicle structures, the restraint system that is composed of seat, bolster, belt system (including pretensioners and load limiters) and airbags as well as the appropriate timing of the active restraint system components (e.g., pretensioner and airbag) are important. The occupant protection system is assessed by standard tests (e.g., vehicle homologation and NCAP programmes). Although almost all cars are performing excellent in these standard tests, it might happen that individual cars are performing worse in real world conditions.

Reichert et al. [Reichert 2013] analysed thoracic and abdominal injury risks in NASS frontal impact accidents without front rail involvement (impact location between the rails to be more precise). While they observed an increased injury risk in the analysed accidents they were unable to replicate the increased injury risk in FE simulations. However, they used US NCAP trigger times for the restraint system triggering which might be the main cause for failing.

During the EU funded FIMCAR project airbag trigger times were analysed in order to define a frontal impact test procedure that is more representative of real world accidents than the Full Width Rigid Barrier (FWRB). For some cars considerably late restraint system triggering was observed in Full Width Deformable Barrier tests (FWDB) with important influence on the dummy readings [Johannsen 2012, Johannsen 2013a, Johannsen 2013b], see Figure 1 as an example.
Figure 1. (a) Late air bag deployment in 40 km/h FWDB test and its consequences for head acceleration. (b) At approximately 30 ms the air bag starts to deploy and (c) at approximately 50 ms the head contacts the deploying air bag [Johannsen 2013a].

In a physical reconstruction of a real world accident (approx. 35 km/h against a small object between the rails) the airbag was fired at approx. 90 ms by the ECU resulting in a contact between deploying airbag and the occupant in forward movement, see Figure 2 (note: in the original accident a rear facing CRS was positioned at the front passenger seat and the reconstruction was repeated to analyse the consequences for adults).

Figure 2. Front seat passenger airbag deployment in an accident reconstruction.

In accident studies it is normally not possible to assess the restraint system firing time due to missing data insight into the control units. So it is normally only possible to assess whether or not an active restraint system was fired at all. The US NASS EDR data (Event Data Recorder) [Da Silva 2008; Dalmotas 2009; Gabler 2008] allows to also consider the firing time as it is normally recorded in the data sets.
METHOD

For the EDR data analysis NASS CDS data for the accident years 2000 to 2012 were considered. Furthermore only MAIS 2+ injured belt restrained (according to NASS CDS data set) drivers that had a frontal impact and where EDR data are available were included. This selection resulted in 188 cases.

For some of the cases the restraint system trigger time was not recorded in the EDR spread sheet. These cases were excluded from the analysis. The same is true for most of the analyses for the cases without airbag deployment. These 19 cases were analysed separately. Following this selection 155 cases with available restraint system triggering time were included.

In a last step roll-over cases were excluded as well as cases without pictures of the case car. Finally 148 MAIS 2+ injured car drivers using the seat belt that were involved in a frontal impact accident in the years 2000 to 2012 without roll-over and available restraint system triggering time were included in the study.

Because the structural interaction appears to be an important influencing factor for the restraint system triggering, all cases were categorised as good structural interaction, under- / overrun, between rails or small overlap (impact zone outside the rails) respectively by checking the images of the deformed cars.

The delta-v of the case cars was retrieved from the EDR data in most of the cases except for cases where the delta-v was not available or wrong without doubts. The latter one was for example true for cases with interruption of the data storage process before the end of the impact phase. In some other cases the recorded delta-v does not fit with the vehicle deformation. For those cases the reconstructed delta-v was considered.

The restraint system triggering time is recorded in the EDR data set as the time between the algorhythm enable signal (normally a vehicle acceleration exceeding 2 g) and the driver airbag system triggering signal (in case of multiple stage airbags the trigger time of the first stage). In some cases the accident was recorded as two or more events in the EDR spread sheets. For those cases the time between preceeding non-deployment events were added to the triggering time of the deployment event.

For the GIDAS analysis frontal impact accidents of ECE R94 compliant cars (i.e., for this study year of first registration after 2003) with MAIS 2+ injured front seat occupants in cars with a front airbag at the relevant seating position were considered. The data set includes 91 occupants meeting the criteria. Structural interaction issues were identified using the CDC and the information in the database whether or not over- / underrun occurred. Delta-v information was obtained from the reconstruction.

DATA ANALYSIS

The data analysis is separated into the analysis of the NASS EDR data with airbag deployment, NASS EDR data without airbag deployment and the GIDAS data.

NASS EDR Data with Airbag Deployment

The focus of this study is the analysis of the restraint system trigger time and to evaluate the restraint system trigger time influencing factors. The distribution of the restraint system trigger time depending on the delta-v is shown in Figure 3. The trigger time varies between 2 and more than 600 ms. The average trigger time is 65 ms. When comparing this time with the average trigger time in FWRB tests of approx. 7 ms [Dalmotas 2009], it is obvious that the trigger time in real world accidents is normally later. However, in an accident with a lower crash severity (e.g., expressed by the delta-v) than in the test it might be sensible to trigger the restraint system later in order to provide optimal performance.
When dividing the accidents in groups with good structural interaction and those with poor structural interaction (i.e., small overlap, under-/ overrun or between rails impact) the average trigger times are 48 ms in the good performing group and 89 ms in the group with poor structural interaction, respectively, see Figure 4.

Although even in the accidents with good structural interactions relatively late timing of the airbag was observed there are more cases with long trigger times in the group with poor structural performance, Figure 4. The mean trigger time for the MAIS 2 cases with good structural interaction differs only slightly from the MAIS 3+ cases, 36 ms compared with 49 ms, respectively – see Figure 5.
Figure 5. Injury severity depending on delta-v and restraint system trigger time for good and poor structural interaction.

For the cases with poor structural interaction the mean trigger time is much longer for both MAIS 2 and MAIS 3+ injured drivers compared to the cases with good structural interaction. Furthermore there is a large difference between MAIS 2 and MAIS 3+ cases in the group with poor structural interaction, 60 ms compared to 119 ms, respectively.

Further separation of the group with poor structural interaction would lead to a too small sample size that would not allow further conclusions.

NASS EDR Data without Airbag Deployment

Looking at the cases without airbag deployment there are much more cases with poor structural interaction than with good structural interaction, see Figure 6. Furthermore 4 of the 5 non deployment cases with good structural interaction had a delta-v below 20 km/h, while this is true for only 3 of the 14 cases with poor structural interaction. In only one case with good structural interaction the non deployment resulted in MAIS 3 injuries while in 3 of the the cases with poor structural interaction the injury severity was MAIS 3, MAIS 4 and MAIS 5, respectively. The MAIS 3 case with good structural interaction was a multiple impact event – first with a small tree and approx. 30 m later with a wall – which could explain the MAIS 3 injuries.
The GIDAS data sample indicates that the injury severity for the head and chest is considerably higher in the cases with poor structural interaction. Especially for head, chest and abdomen it is expected that the injury risk is sensitive on the restraint system triggering time. For the arms and legs there appears to be no important difference between the accidents with and without good structural interaction. For the abdomen it is difficult to judge – the share of abdomen injuries is higher in the group with poor structural interaction while AIS 3+ injuries are only observed for the accidents with good structural interaction.

**GIDAS Data**

**Figure 6. Delta-v and injury severity in non deployment cases.**

**Figure 7. Injury severity depending on structural interaction assessment.**
DISCUSSION AND CONCLUSION

The analysed accident data indicates that restraint system triggering time is much later than in the standard tests (e.g., FMVSS 208 full frontal test) in a large number of cases. While triggering times just being later does not necessarily mean that they are too late. For example accidents with light accident severity require a late triggering of restraint systems. However, in a small number of crash tests it was proven that the late ECU airbag triggering results in considerably higher dummy readings.

The NASS EDR data with airbag deployment indicates that the restraint system trigger time tends to be later in cases with poor structural interaction (i.e., small overlap, impact between rails and under- / overrun scenarios). It is expected that the firing decision is more difficult to safeguard in these conditions compared for example with a FWRB test.

Furthermore the injury severity appears to be higher in cases with high accident severity and late restraint system trigger time. Furthermore the analysed GIDAS data shows that the injury severity for those body regions that are expected to be sensitive to the restraint system trigger time (i.e., head and thorax) is higher in the cases with poor structural interaction. This is interesting because those cases are expected to mainly increase intrusion related injury risks. However, for accidents with high accident severity and poor structural interaction it is also expected that the acceleration history would have a more injury causing shape (i.e., back loaded pulse).

Unfortunately for approx. half of the analysed NASS EDR data (accident years 2011 nd 2012) the individual injuries were not yet coded at the NASS CDS web site. Therefore it is not yet possible to confirm the observation from the GIDAS data sample with EDR data. The EDR data set mainly contains newer accident years due to the introduction phase of EDR.

As a next step it would be sensible to run accident reconstructions with observed and modified restraint system trigger time in order to verify whether or not the timing was wrong in individual cases.

ACKNOWLEDGEMENTS

The NASS EDR data and NASS CDS data analysed for this study was collected during a Bachelor Thesis of Iryna Shevchenko [Shevchenko 2014].

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ABSTRACT

ANCAP crash tests are conducted to well-established protocols and use driver and front passenger anthropomorphic test devices (ATDs) that represent 50 percentile (%ile) adult males. Most modern vehicles do well in these crash tests. However, concerns have been raised about the protection provided to smaller occupants. In 2013 ANCAP conducted a 64km/h frontal offset crash test of a Holden Commodore VF Ute (single-cab pick-up) with two Hybrid III 50%ile male ATDs (AM50). In 2014 the opportunity arose to conduct a further research crash test of a Commodore VF Ute using a small adult female driver ATD (5%ile adult female) and a 6 year old child ATD (Hybrid III 6) in a booster seat in the front passenger seating position.

The purpose of the research test was to determine whether the two occupants had an increased risk of injury, compared with the adult male ATDs.

The occupant injury measures for the smaller occupants were compared to the results of the previous vehicle crash test (with 50%ile adult male front occupants) in order to make comparisons between the level of protection offered to occupants of different sizes for this type of crash. Furthermore, the authors examined whether the restraint and airbag systems that perform well for 50%ile adult male occupants provide comparable protection for smaller occupants and whether there are any additional hazards for smaller occupants.

The outcome of the research was that for the case examined, with the available methods for assessing injury risk, smaller occupants appear to be offered comparative protection in a frontal offset impact for this particular vehicle model. It is apparent that the test vehicle manufacturer included consideration of smaller occupants in the design and development of this vehicle model.

INTRODUCTION AND BACKGROUND

In accordance with long-established protocols, ANCAP crash tests focus on average sized male occupants for the front seating positions (EuroNCAP 2011) Since 1993 Hybrid III 50th %ile adult male (AM50) ATDs have been used as driver and front passenger occupants for the ANCAP frontal offset crash test.

In 2014 a research test was conducted to examine the injury risk and occupant dynamics in a 64 km/h offset frontal crash, tested with a Hybrid III 5th %ile adult female driver (AF5) and a Hybrid III 6 year old (H6) child ATD restrained in a high back booster seat with a top-tether.

Front seat positions do not normally have an anchorage point for CRS because they are provided for rear seats, where available. This test was feasible because the Commodore Ute has an upper anchorage for the child restraint system (CRS) top-tether in this seat position and many Australian booster seats are provided with top-tethers. Where a top-tether anchorage is provided for a front passenger seat in a vehicle with one row of seats, a
child in a forward facing CRS or booster seat may occupy this position, even if a passenger frontal airbag is fitted. The Commodore Ute does not have provision for disabling the frontal passenger airbag when a CRS is installed in the front seat.

![Research test vehicle impact with deformable barrier.](image)

**Figure 1. Research test vehicle impact with deformable barrier.**

The possible safety concerns with a small driver are:

a) that they are close to the deploying airbag and might be at risk of injury from the airbag  
b) that because they are closer to the steering wheel there might be an increased risk of contact with the steering wheel  
c) that the restraint system (seat belt, airbag and seat) is not optimised for this size occupant and so does not provide the same protection as the 50%ile male occupant (in particular the potential for an increased risk of chest injury)  
d) that with the seat fully forward the knees are more likely to contact the instrument panel and/or steering column, with consequent extra risk of injury

In the case of a child in a booster seat, the possible concerns are:

e) interaction with the airbag might pose an injury risk  
f) the lap portion of the seat belt might ride up and penetrate into the abdomen, increasing injury risk - seat belt pretensioners have been observed contributing to this effect (in rear seats - Tylko 2012)  
g) the child starts to slide under the seat belt, particularly if there is excessive movement of the booster seat  
h) the sash portion of the belt may not retained on the ATD shoulder up to the peak of the crash

**SELECTION OF TEST VEHICLE**

A fairly unique feature of the Australian vehicle population is the utility, or "ute" - a small pick-up with a single row of seats. Many Australian pick-ups are based on popular sedan models where the rear row of seats and luggage area of the sedan are replaced by a utility tray. Over the years the Holden Commodore Ute and Ford Falcon Ute have been popular vehicles. To meet a demand from families both Holden and Ford have provided a top tether anchorage for the front passenger seat of these models. Manufacturers instructions state that this seating position must not be used for a rearward facing CRS (in effect an infant restraint) and to slide the seat back as far as possible for forward facing CRS and booster seats.

A 2014 model year Holden Commodore Ute (VF variant) was chosen for the research test, partly because data from an ANCAP crash test with AM50 ATDs were available for comparison.

**SELECTION OF DRIVER (AF5)**

Vehicle regulations and NCAP protocols generally use AM50 ATDs as front occupants in frontal crashes tests.

Vehicle manufacturers design vehicles to provide protection to a wide size range of occupants however information about how well smaller occupants are protected is not commonly available.

In 2015 EuroNCAP introduced a 50km/h full-width frontal crash test with AF5 ATDs in the driver and kerbside rear seat. Other crash-test organisations, such as Japan NCAP, and China NCAP, also use the AF5 ATD in some seating positions. It was therefore decided to use the AF5 in the research test.
The injury limits specified in the Euro NCAP adult occupant protection protocol (EuroNCAP, July 2014) were used for assessing the research test outcomes for the AF5.

**SELECTION OF FRONT PASSENGER (H6 restrained in booster seat)**

Australian Research in the late 1970s led to the development of an Australian Standard (AS/NZS 1754) for child restraints that required dedicated child restraint systems (CRS) to have a top tether to control pitch rotation of the CRS. Since then all Australian CRS with an in-built harness have had the top tether system and (under Australian Design Rule 34/02 - Child Restraint Anchorages and Child Restraint Anchor Fittings) all light passenger vehicles that have two or more rows of seats have to be fitted with top tether anchorages for some rear seats (Paine 2003). In 2004 the standard was updated to encourage all booster seats (that utilise the adult seat belt to restrain the occupant) to have a top tether (AS/NZS 1754:2004).

For more than a decade the Commodore Ute and Falcon Ute have had top tether anchorage available for the front passenger seat. This was mainly to meet a demand from consumers to carry young children in this seating position because no rear seat was available.

Since the 1980s Australian researchers have monitored injuries to children in car crashes and have not come across any cases of injuries to correctly restrained children in this front seating position due to airbag deployment (Brown 2013). Nevertheless there remain concerns within the community due to tragic experiences in the USA with children and airbags - due mainly to children being unrestrained, out of position, and/or early aggressive designs of airbags (Child Passenger Safety Committee on Injury, Violence, and Poison Prevention, 2011).

A Hybrid III 6 year old child ATD was selected to occupy the front passenger seating position using a booster seat with a top tether. Since the ATD is restrained by the adult seat belt and not a dedicated harness this was considered to be a worse case for issues such as restraint geometry, submarining and head excursion. A 6 year old child in a booster seat was considered the most likely configuration to reveal the above problems. A smaller child in a forward facing child seat may not have reveal seat belt-related problems since forward facing restraints for smaller children use an inbuilt harness, rather than the seat belt, to restrain the occupant. An infant in a rearward facing child seat is not recommended for the front passenger seating position, although Suratno and others (Suratno et al, 2009) concluded that there was unlikely to be additional injury risk with this configuration with modern designs of passenger airbags.

The H6 was restrained in a Hipod Boston Booster child seat. This booster performed well in the Australian Child Restraint Evaluation Program (CREP) and had a top tether.

**SET UP AND SEAT POSITIONING**

The vehicle was set up according to EuroNCAP frontal offset protocol version 5.1 (EuroNCAP, 2011) with the following changes:

- An AF5 replaced the AM50 in the driver position
- A H6 restrained in a booster seat replaced the AM50 in the front passenger position
- Driver seat and steering wheel were set to the following positions
  - The driver's seat was fully forward
  - The driver’s seat was set in the mid height setting.
  - The steering wheel was set in the most forward and mid height setting.
  - The driver’s head restraint was set in the lowest and fully forward tilt setting.
- The front passenger seat was set in the rearmost locking position
- The front passenger seat head restraint was removed as it obstructed installation of the booster seat.
- In addition to the standard ANCAP camera positions three cameras were mounted within the vehicle cabin. One roof camera was positioned to capture the left side of the driver ATD and another roof camera was positioned to capture the right side of the front passenger ATD. A third camera was mounted on the dashboard and was positioned to capture the right side of the front passenger ATD. These on-board cameras were positioned to capture views that would assist in the assessments, including lap belt performance and frontal airbag interaction. Footage was obtained from both roof mounted cameras but the dashboard mounted camera failed to record usable footage.
OBSERVATIONS AND TEST DATA

Injury scoring scales and limits for the EuroNCAP full-width frontal impact were used for the AF5 driver ATD. The AF5 scales are not necessarily the same as the AM50 scales because they reflect the risk of serious injury for a person of that size.

NHTSA has proposed Hybrid III 6 year old injury limits for US federal regulation FMVSS208 (Kleinberger, et al, 1998) and FMVSS 213 (NHTSA 2002). These injury limits were used for the H6 ATD because injury limits are not defined for this ATD in ANCAP and EuroNCAP protocols.

Injury data for the AF5 and H6 ATDs are shown below in Table 1 and Table 2 respectively.

Driver Outcomes

The calculated points score for the AF5 driver (13.42pts) was similar to the ANCAP points score for the AM50 driver (13.52pts) in the official ANCAP test.

The driver injury readings that exceeded the NCAP lower injury limits were the chest compression and tibia index. The AF5 chest compression of 20.7mm resulted in a score of 3.55 points compared to the official test (AM50) of 26.4mm for a score of 3.37 points indicating an ‘acceptable’ level of risk to the chest in both tests (using the NCAP injury risk scales - which are different for AM50 and AF5).

The AF5 tibia index of 0.88 returned a points score of 1.87 compared to the official test (AF50) tibia index of 0.82 for a score of 2.15 points. However the Euro NCAP full-width frontal crash test does not make use of the tibia readings for scoring due to uncertainty in the AF5 risk values for tibia and so caution should be exercised in interpreting the lower leg injury risk in the research test.

Although twisting of the driver neck was observed upon rebound from the airbag (a peak neck moment of 28.45Nm at 196ms) the twisting of the AF5 neck in this research test was not sufficient to warrant a modifier under the ANCAP protocols. No estimation of injury risk is carried out after the peak of the impact due to limited biofidelity of the H3 neck for this type of movement.

![Figure 2. Twisting of the driver neck.](image)

The driver airbag contacted the driver ATD’s face while the airbag was inflating (at approx. t= 35ms) and this may have contributed to the neck twisting. However the estimated airbag deployment speed at the time of contact was below the 90m/s limit used by Euro NCAP to determine whether the airbag deployment is potentially hazardous. It should be noted, however that although the hazardous airbag deployment modifier is used by EuroNCAP is it generally not applied by ANCAP. Airbag contact was judged to be stable and no undesirable interactions between the driver and airbag were observed.
Table 1. AF5 Driver ATD injury data

<table>
<thead>
<tr>
<th>Driver ATD (AF5)</th>
<th>Limits for scoring purposes*</th>
<th>Comment on injury outcomes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head G (Res) (g)</td>
<td>46.8</td>
<td>Good</td>
</tr>
<tr>
<td>3ms Head G (g)</td>
<td>42.8</td>
<td>Good</td>
</tr>
<tr>
<td>HIC (36ms)</td>
<td>243</td>
<td>Good</td>
</tr>
<tr>
<td>Neck Shear (kN)</td>
<td>0.30</td>
<td>Good</td>
</tr>
<tr>
<td>Neck Tension (kN)</td>
<td>1.15</td>
<td>Good</td>
</tr>
<tr>
<td>Neck Extension (Nm)</td>
<td>10.0</td>
<td>Good</td>
</tr>
<tr>
<td>Chest compression (mm)</td>
<td>20.7</td>
<td>Acceptable</td>
</tr>
<tr>
<td>Chest viscous criterion (m/s)</td>
<td>0.07</td>
<td>Good</td>
</tr>
<tr>
<td>Femur compression (kN) (L/R)</td>
<td>1.51/2.38</td>
<td>Good</td>
</tr>
<tr>
<td>Knee slider (mm) (L/R)</td>
<td>0.0/0.03</td>
<td>Good</td>
</tr>
<tr>
<td>Tibia index, upper (L/R)</td>
<td>0.56/0.47</td>
<td>Acceptable</td>
</tr>
<tr>
<td>Tibia index, lower (L/R)</td>
<td>0.88/0.37</td>
<td>Marginal</td>
</tr>
<tr>
<td>Tibia compression (L/R)</td>
<td>1.69/1.08</td>
<td>Good</td>
</tr>
</tbody>
</table>

* Zero if more than first value, maximum (4) if less than second value (based on Euro NCAP protocol for full frontal crash with AF5 driver)
# Not assessed by Euro NCAP for AF5. AM50 limit shown in this table

Figure 3 a-g. Sequence of snapshots from on-board camera - driver
Front Passenger Outcomes
The H6 passenger in the booster seat appeared to be well restrained with a relatively low likelihood of serious injury. The measured injury values were well below the limits for scoring purposes. The H6 injury data is provided in Table 2.

The passenger airbag fully inflated prior to contact with the H6 (contact occurs around $t = 74\text{ms}$ when the crown of the ATD head contacts the bag. After contact with the airbag (upon rebound) the top of the H6 head slides across the surface of the airbag.

The child ATD forward head excursion was estimated by analysing the test footage and observing the maximum forward position of the head against reference markings on the vehicle. The maximum forward head excursion for the H6 ATD was estimated to be $300\text{mm}$ and no hard contacts were observed.

<table>
<thead>
<tr>
<th>Table 2. H6 Passenger ATD injury data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Passenger ATD (H6) Injury value</td>
</tr>
<tr>
<td></td>
</tr>
<tr>
<td>$\text{HIC}_{16}$</td>
</tr>
<tr>
<td>$\text{HIC}_{15}$</td>
</tr>
<tr>
<td>$\text{Nij}$</td>
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<tr>
<td>Chest compression (mm)</td>
</tr>
<tr>
<td>Chest acceleration (3ms) (g)</td>
</tr>
<tr>
<td>Head excursion (mm)</td>
</tr>
</tbody>
</table>

* Proposed FMVSS 208 limit (Kleinberger, et al, 1998)
^ Proposed FMVSS 213 limit (NHTSA, 2002)
In addition to injury measurements, the ANCAP/EuroNCAP scoring system provides for "modifiers" that reduce the score if there is a further risk of injury that is not captured by the ATD injury measurements. These modifiers include steering column movement, structural integrity of the cabin, pedal movement, unstable head contact with the airbag, airbag bottoming out (resulting in head or chest contact with the steering column) and facial injury risk due to the manner in which the airbag deploys. The latter assessment is not currently applied by ANCAP.

In addition to these modifier assessments a preliminary assessment of the dynamic performance of the seat belts in accordance with observations by Suzanne Tylko from Transport Canada (Tylko & Bussières 2012) was carried out:

a) the sash portion should be retained on the shoulder up to the peak of the crash. It should not fall off the shoulder or load the neck

b) the lap portion should load the pelvis and not ride up and into the abdomen

c) there should be no tendency for submarining, where the ATD tends to slide down and forward

From analysis of the video footage there was no evidence of the lap portion of the seat belt riding up and loading the abdomen of the child ATD, nor was there evidence of the child ATD submarining. The sash portion of the seat belt appeared to be retained on the child ATD shoulder until after the peak of the crash (evidence of retention until 150ms) and there was no evidence that the seat belt loaded the neck of the ATD.

Pretensioners deployed for both occupant positions and no evidence of poor belt geometry caused by pretensioner deployment was observed.

<table>
<thead>
<tr>
<th>Research test Driver (AF5)</th>
<th>ANCAP test Driver (AM50)</th>
<th>Research test Passenger (H6)</th>
<th>ANCAP test Passenger (AM50)</th>
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<tr>
<td>Head G (Res) (g)</td>
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<td>35.8</td>
<td>45.0</td>
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<tr>
<td>3ms Head G (g)</td>
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<td>35.5</td>
<td>44.4</td>
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<tr>
<td>HIC (36ms)</td>
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<td>350</td>
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<tr>
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<td>0.46</td>
<td>*</td>
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<tr>
<td>Neck Tension (kN)</td>
<td>1.15</td>
<td>0.37</td>
<td>*</td>
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<tr>
<td>Neck Extension (Nm)</td>
<td>10.0</td>
<td>8.2</td>
<td>*</td>
</tr>
<tr>
<td>Chest compression (mm)</td>
<td>20.7</td>
<td>26.4</td>
<td>20.7</td>
</tr>
<tr>
<td>Chest viscous criterion (m/s)</td>
<td>0.07</td>
<td>0.07</td>
<td>*</td>
</tr>
<tr>
<td>Femur compression (kN) (L/R)</td>
<td>1.51/2.38</td>
<td>2.04/0.08</td>
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<tr>
<td>Knee slider (mm) (L/R)</td>
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<td>0.46/0.00</td>
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<tr>
<td>Tibia index, lower (L/R)</td>
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<td>0.82/0.31</td>
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<td>Tibia compression (L/R)</td>
<td>1.69/1.08</td>
<td>3.89/1.82</td>
<td>*</td>
</tr>
</tbody>
</table>

# Note: Different injury scales are applied to each ATD. Direct comparison of injury data between ATDs is not possible.

* Not recorded for this ATD

Restraint System Performance
In addition to injury measurements, the ANCAP/EuroNCAP scoring system provides for "modifiers" that reduce the score if there is a further risk of injury that is not captured by the ATD injury measurements. These modifiers include steering column movement, structural integrity of the cabin, pedal movement, unstable head contact with the airbag, airbag bottoming out (resulting in head or chest contact with the steering column) and facial injury risk due to the manner in which the airbag deploys. The latter assessment is not currently applied by ANCAP.

In addition to these modifier assessments a preliminary assessment of the dynamic performance of the seat belts in accordance with observations by Suzanne Tylko from Transport Canada (Tylko & Bussières 2012) was carried out:

a) the sash portion should be retained on the shoulder up to the peak of the crash. It should not fall off the shoulder or load the neck

b) the lap portion should load the pelvis and not ride up and into the abdomen

c) there should be no tendency for submarining, where the ATD tends to slide down and forward

From analysis of the video footage there was no evidence of the lap portion of the seat belt riding up and loading the abdomen of the child ATD, nor was there evidence of the child ATD submarining. The sash portion of the seat belt appeared to be retained on the child ATD shoulder until after the peak of the crash (evidence of retention until 150ms) and there was no evidence that the seat belt loaded the neck of the ATD.

Pretensioners deployed for both occupant positions and no evidence of poor belt geometry caused by pretensioner deployment was observed.
LIMITATIONS OF THE RESEARCH TEST

The findings are based on one crash test with two occupants. However, the research draws on experience with a wide range of vehicles and compares directly with a crash test of the same vehicle model. Due to the different ATDs used, direct comparison between occupant injury readings is not possible but the scaled Euro NCAP scores do relate to the risk of serious injury for both types of adult ATDs.

Although tibia values are not used for the EuroNCAP full-width frontal test the values have been recorded here and compared to AM50 limits. Caution should be exercised in interpreting the lower leg injury risk in the research test. Direct comparison to AM50 values is not possible but the values provide an indication of injury risk for this body region.

The H6 ATD is only instrumented for head acceleration (3 axis), neck shear force and bending moment (3 axis) chest acceleration (3 axis) and displacement (x axis) and pelvis acceleration (3 axis). Injury outcomes for other body regions are not measured.

Child forward head excursion is approximate only, using external cameras and reference marks on the vehicle. This method is susceptible to parallax error. Strategically positioned on-board cameras would provide greater accuracy.

Additional on-board cameras could have provided an improved view of the seat belt buckles, retention of the sash portion of the seat belts on the ATD’s shoulders and interaction of the lap portion of the seat belts with occupant abdomens.

CONCLUSIONS

A successful research crash test was conducted to determine the crash protection provided to a small female driver and a 6 year old child restrained in a booster seat in the front seat of a utility style vehicle.

The 5%ile female driver ATD injury measurements indicated that the risk of serious injury was no worse than that of a 50%ile male driver in the same model vehicle and same type of crash test. The restraint system appeared to work well with no sign of submarining or lap-belt penetration into the abdomen. This test was conducted with the driver seat in its most forward position, which is considered to be a worst-case scenario for risk of airbag injury.

The child ATD injury measurements also indicated a low risk of serious injury. The restraint system appeared to work well with no sign of submarining or lap-belt penetration into the abdomen. It is likely that the head contact with the airbag would reduce head, neck and chest loads, compared with no airbag.

This test was conducted with the passenger seat in its rearmost position, as recommended by the manufacturer. Although this is considered to be the best-case scenario there is unlikely to be an elevated risk of serious injury from the airbag when the seat is located further forward.

It is concluded that, for the case examined, with the available methods for assessing injury risk, smaller occupants appear to be offered comparative protection in a frontal offset impact for this particular vehicle model. It is apparent that the vehicle manufacturer included consideration of smaller occupants in the design and development of this vehicle model.

REFERENCES


Child Passenger Safety Committee on Injury, Violence, and Poison Prevention Pediatrics 2011;127;e1050; originally published online March 21, 2011; DOI: 10.1542/peds.2011-0215


Innovative Seat Belt System for Reduced Chest Deflection

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Paper Number 15-0371

ABSTRACT

An innovative seat belt concept aimed at reducing chest injuries was evaluated by means of mechanical tests and mathematical modelling. The tools used were mechanical THOR dummy, mathematical THOR dummy model, THUMS human body model and (Post Mortem Human Surrogates) PMHS. The potential chest injury reducing benefits with an innovative seat belt concept relative to a state of the art belt system was evaluated in sled tests. The reference belt system was a state of the art belt system with a pretensioning of 2kN at the retractor, a force limiter of 4.5kN and an outer lap belt pretensioner with a pretensioning force of 3.5kN. The innovative seat belt concept was consisting of a retractor equipped with a 2kN pretensioner at the retractor, a force limiter of 6kN and two 3.5kN pretensioners at the buckle and outer lap belt anchorage. The belt was split at the buckle and the lower end of the diagonal belt was moved 50mm forward. With the altered belt geometry the load on the lower part of the chest was reduced and the peak chest deflection was reduced relative to a state of the art belt system. In mechanical sled tests with rigid seat and an impact velocity of 35 and 30kph with the THOR dummy peak chest deflection was reduced by 8.0mm compared to a state of the art belt system. In the corresponding sled model with the THOR dummy model peak chest deflection was reduced by 13mm. Head x-displacement was increased by 26mm for the mechanical THOR dummy and 24mm for the THOR dummy model. For the THUMS model peak chest deflection was reduced by 10mm with the split buckle system.

Generally for the mechanical THOR dummy, the THOR dummy model and the Autoliv THUMS model peak chest deflection was reduced by approximately 8-13mm with the split buckle belt system while only a minor increase in head x-displacement was observed relative to a state of the art belt system.

INTRODUCTION

Occupant fatalities and injuries in car crashes remains a global health issue. World Health Organization (WHO) found road injuries to be the 9th leading cause of death in the world after diseases such as stroke and heart disease [1]. Restraint systems have been developed and improved and have contributed to reduce the number of seriously and fatally injured occupants in vehicle crashes. However people are still being injured and killed in traffic. In Europe, frontal crashes still account 40% of all fatalities in car accidents [2]. In these accidents, injuries to the thorax are common and account for 13% of all moderate injuries and 29% of all severe injuries [3]. Statistics also show that the most frequent severe thoracic injuries are rib fractures [4]. Furthermore, the number of rib fractures is a good indicator of injuries to the thoracic and abdominal organs [5].

In testing with post mortem human subjects (PMHS) it was found that the injury threshold for chest deflection is strongly dependent upon the age of the subject [6]. This is true regardless of whether injury onset or severe injury is considered. A 30-year-old has a 50% risk of sustaining one rib fracture at a chest deflection level of 35% (of the
total width of the chest). A 70-year-old has a 50% risk of more than 6 rib fractures at 33% deflection. The age-fragility correlation is particularly important while in 2012 about 17% of all Europeans were aged 65 and older, the share of those above age 65 will rise to 28% in 2020 [7]. Due to their greater frailty and fragility, there is a need to develop restraint systems that reduces the load on the chest of the elderly occupants.

The use of mathematical human body models for restraint evaluation is increasing. One advantage with the human body models is that injuries can be assessed based on physical parameters such as strain for fracture analysis. A strain-based probabilistic method to predict rib fractures with finite element human body models was developed based on data from cortical bone coupon tests [8]. The method combined the results with collision exposure information to predict injury risk and potential intervention effectiveness in the field. An age-adjusted ultimate strain distribution was used to estimate local rib fracture probabilities within an FE model. These local probabilities were combined to predict injury risk and severity within the whole ribcage.

The Autoliv THUMS model was derived from the THUMS model (Total Human Body Model for Safety, version 1.4). The THUMS model was updated with a number of in-house modifications to improve its biofidelity in frontal impacts using table top and sled PMHS tests [9, 10]. The rhomboids major and rhomboids minor muscles that connect the medial border of the scapula to the spine were missing in the original THUMS and were added to the Autoliv THUMS. The aim of the study is to evaluate the potential injury reducing benefits of an innovative belt system, which alters the load distribution of the seat belt on the chest, relative to a state of the art 3-point belt system.

METHOD

The evaluation of the innovative seat belt system was carried out by combining mathematical modelling with mechanical testing [11]. The sled test fixture (Gold Standard) consisted of a rigid metallic frame allowing complete visual access to the occupant while preserving the basic geometry of a standard seating position of a passenger car. This test fixture was used elsewhere as a reasonable approximation to the passenger posture in the study of ATD biofidelity and in the development of thoracic injury criteria [12, 13]. The 50%-ile THOR model (THOR-M version 0.6) was used to validate the test environment (seat and seatbelt) by matching predictions from the model to experimentally measured THOR test responses (Figure 1). The impact velocity was 35 km/h and peak acceleration 20g and a duration of approximately 80ms. The reference belt system was a state of the art belt system with a pretensioning of 2kN at the retractor, a force limiter of 4.5kN of the diagonal belt and a 3.5kN outboard lap belt pretensioner.
When satisfactorily agreement between mechanical test results and mathematical model predictions was obtained for the reference belt system, the model of the restraint system was considered to be validated. The reference belt system was replaced with an innovative belt system (split buckle). The split buckle system consisted of a retractor equipped with a 3kN pretensioner at the retractor, a force limiter of 6kN at the diagonal belt and two 3.5kN pretensioners, one at the buckle and one at the outer lap belt anchorage. The belt was split at the buckle and the lower end of the diagonal belt was moved 50mm forward (Figure 2). THOR simulations were carried out with the advanced belt system, and thereafter corresponding mechanical tests with the THOR dummy were run to confirm the model predictions.

For both belt systems, standard and split buckle, chest loading in terms of mutlipoint deflection and strain were evaluated. The deflection measurement in the THOR consists of four 3D IR-Traccs (3D Infra-Red Telescoping Rod for the Assessment of Chest Compression) and their location was replicated in the THUMS model by deflection measurement at the level of the 4th rib for upper thorax, and between rib 6 and 7 for the lower thorax (Figure 3). In addition for THUMS rib strain was assessed. Head displacement (x-direction) and belt forces were also evaluated for the two systems.
The THOR dummy model was replaced with the human body model Autoliv THUMS. Both the reference belt system and the split buckle system were analyzed with the Autoliv THUMS model. Post Mortem Human Subject (PMHS) tests were previously carried out in the same test set-up using both the reference belt system and the split buckle system, and the results were to some extent compared to the Autoliv THUMS model results in this study [11]. The PMHS were of approximately the same age (reference: 42 YO, 60 kg, 159 cm; advanced: 39 YO, 62 kg; 175 cm). The predicted head x-displacement and belt forces were compared to the corresponding measurements in the PMHS tests, and the predicted risk to sustain rib fractures were compared to the number of fractured ribs in the post mortem human subject (PMHS) tests.

A parameter study, to evaluate the influence on chest deflection on diagonal belt lower attachment point, was carried out. The lower attachment point of the diagonal belt was moved in steps of 50mm horizontally (Figure 4). The buckle was moved 50mm rearwards, 50mm forward and 100mm forward relative to the test position.

**RESULTS**

**Model Benchmark**

The average peak chest x-deflection in the mechanical reference test with a state of the art belt system was for the mechanical THOR dummy 32mm in upper left IR-Tracc (Figure 5). The predicted peak chest deflection with THOR dummy model was 33mm in the upper left IR-Tracc. For the split buckle system the chest deflection in the mechanical THOR test was 25mm. For the model the predicted peak chest deflection was 20mm.

For the mechanical reference test with the THOR dummy head x-displacement was 397mm. The predicted head x-displacement was 343mm (Figure 5). For the split buckle system the x-displacement in the mechanical reference test was 405mm while for the model the head x-displacement was 367mm.
Peak belt force in the mechanical reference test was 5582N (Figure 5). While the predicted force was 4770N. In the mechanical split buckle belt system test the average diagonal belt force was 6085N. The predicted force for the split buckle system was 5606N.

For the THUMS model the predicted chest deflection for the reference belt system was 30mm in the upper left transducer and for the split buckle system the predicted chest deflection was 20mm (Figure 6).

The max head x-displacement in the reference test for the PMHS was 189mm while for THUMS the predicted head x-displacement was 378mm (Figure 6). For the split buckle system the max head x-displacement for the PMHS was 306mm while for THUMS the predicted x-displacement was 409mm.

The diagonal belt force in the reference PMHS test was 4000N. The predicted force with THUMS was 3900N. For the split buckle system the diagonal belt force in the PMHS test was 6000N while the predicted force was 5500kN.
Split Buckle Parameter Study

In the parameter study carried out for the split buckle system the lower attachment point for the diagonal point was moved in steps of 50mm. The predicted peak chest deflection for THOR was obtained at the upper left IR-Tracc for all locations of the lower attachment point (Figure 7). By moving the attachment point 150mm forward in the vehicle the chest deflection was reduced from 22mm to 15mm while head x-displacement was increased by 41mm, from 357mm to 398mm.

For THUMS the peak chest deflection was for the upper left IR-Tracc (Figure 7). By moving the lower attachment point 150mm forward peak chest deflection was reduced from 24mm to 15mm while the head x-displacement was increased by 95mm, from 376mm to 471mm.

Peak belt forces for both the THOR dummy model and THUMS were 5500N. The belt forces were not altered when the attachment point was moved forward in the vehicle.
For the Autoliv THUMS 0 fractured ribs were predicted for 39 and 60 year old occupants for both the reference belt system and the split buckle system.

**DISCUSSION**

By splitting the belt system and moving the lower attachment point of the diagonal belt forward in the vehicle, increasing the pretensioning force and adding a lap belt pretensioner peak chest deflection was reduced. By moving the lower attachment point forward the load from the belt is reduced in the lower part of the chest. By increasing the level of the load limiter the excursion of the body was kept at the same level as for the reference belt system while the chest deflection was reduced. There is a potentially increased risk of clavicle injuries due to the increased force at the shoulder level. However, in the PMHS tests no clavicle injuries were observed [11]. In addition, the influence on pelvis kinematics of the split buckle system will be evaluated in future analysis.

A split buckle in vehicle system can consists of one buckle that splits in a crash (Figure 8). The occupant buckles up in the same way as a state-of-the-art belt system today. Both the diagonal belt and the lap belt are pretensioned by moving the buckle downwards. In a crash the lower point of the diagonal belt is moved forward while the lap portion of the belt remains at the initial location.
Generally there was agreement between the predictions from the model and the results from the mechanical sled tests. However, there were some discrepancies between predicted and measured chest deflections. However, for the reference test there was agreement for the two IR-Traccs with the greatest deflections (Figure 5). For the split buckle test there was agreement between the predicted and measured upper left and upper right deflections. The reason for the poor agreement for some of the chest deflection predictions can be that the model of the THOR dummy used was a beta version of the model and not fully validated. Due to the fact that the THOR model predicted the same reduction in chest deflection as was observed in the mechanical tests the model was considered sufficiently valid for the intentions of this study.

For the PMHS there was a significant (over 100mm) increase in head x-displacement with the split buckle system relative to the excursion with the reference belt system. Such increase was not observed for neither the mechanical THOR dummy, the mathematical THOR dummy model or for THUMS. For the mechanical THOR dummy there was an increase of 37mm in head x-displacement for the THOR dummy model and THUMS there were an increase in head x-displacement of 71mm and 31mm respectively. The reason for the significant increase in head x-displacement for the PMHS in the split buckle system compared to the reference system was an increased forward motion of the pelvis that resulted in reduced torso pitch. This finding is discussed in [11] in detail. However, it is important to point out that just one PMHS test is not enough to assess the performance of both systems. In future analyses, the reason for the increase for the PMHS will be investigated in more detail.

Figure 8. Split buckle activation

Generally there was agreement between the predictions from the model and the results from the mechanical sled tests. However, there were some discrepancies between predicted and measured chest deflections. However, for the reference test there was agreement for the two IR-Traccs with the greatest deflections (Figure 5). For the split buckle test there was agreement between the predicted and measured upper left and upper right deflections. The reason for the poor agreement for some of the chest deflection predictions can be that the model of the THOR dummy used was a beta version of the model and not fully validated. Due to the fact that the THOR model predicted the same reduction in chest deflection as was observed in the mechanical tests the model was considered sufficiently valid for the intentions of this study.

For the PMHS there was a significant (over 100mm) increase in head x-displacement with the split buckle system relative to the excursion with the reference belt system. Such increase was not observed for neither the mechanical THOR dummy, the mathematical THOR dummy model or for THUMS. For the mechanical THOR dummy there was an increase of 37mm in head x-displacement for the THOR dummy model and THUMS there were an increase in head x-displacement of 71mm and 31mm respectively. The reason for the significant increase in head x-displacement for the PMHS in the split buckle system compared to the reference system was an increased forward motion of the pelvis that resulted in reduced torso pitch. This finding is discussed in [11] in detail. However, it is important to point out that just one PMHS test is not enough to assess the performance of both systems. In future analyses, the reason for the increase for the PMHS will be investigated in more detail.
CONCLUSION

The split buckle system:

- can reduce chest deflection for a belted occupant
- can have a minor increase in head x-displacement
- can reduce the number of fractured ribs for an elderly occupant
- can be made with one buckle for identical buckle up procedure as today
REFERENCES

INNOVATIVE RESTRAINTS TO PREVENT CHEST INJURIES IN FRONTAL IMPACTS

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Paper Number 15-0381

ABSTRACT

By 2050, 21% of world population is expected to be older than 60 years. This age shift poses a serious challenge to the protection of car occupants, as fragility and frailty are associated to increasing age. Advanced restraint systems that aim to reduce chest loading by implementing load limiters or inflatable parts have been introduced in the market over the last years. This paper investigates the kinematics and dynamics of two surrogates (THOR dummy, Post Mortem Human Surrogates or PMHS) in 35 km/h impacts under the action of two different restraints: a pretensioning, force-limiting seat belt (PT+FL) and a concept design consisting of two separate shoulder and lap belt bands (split buckle system or SB). Three repeats per condition where done with the THOR dummy, while only one PMHS was tested per restraint system. With respect to the PT+FL, the results from the THOR tests showed that the SB seat belt decreased chest deflection significantly without a substantial increase of the forward displacement of the head. The PT+FL belt allowed the pelvis of the PMHS to move forward preventing the rotation of the torso and therefore reducing the forward excursion of the head. The PMHS test with the SB resulted in improved kinematics compared with the PT+FL. A complete understanding of the kinematics and dynamics induced by these restraints would require additional PMHS tests.

INTRODUCTION

Life expectancy in Europe rose by eight years between 1960 and 2006 [1]. Additionally, the number of births has declined continuously over the last decades, resulting in an increase of the proportion of elderly people in the European population. While in 2012 about 17% of all Europeans were aged 65 and older, the share of those over age 65 will rise to 28% in 2020 [2]. According to United Nations, the same trend is observed in the whole world although it is particularly important in Eastern Asia (Japan, China, Mongolia), Europe and North America. Interestingly, fast population ageing will take place mainly in the less developed regions. Globally, it is forecasted that world’s population age 60 years or over will increase over the following years to reach 21% in 2050 [3].

In 2010, 6,563 elderly people were killed in road traffic accidents in the 24 European Member States for which CARE data is available [4]. This constitutes 21.7% of fatalities of all ages in 2010. What is even more significant is that the proportion of elderly fatalities has been increasing steadily for the last 10 years. While this increase is undoubtedly related to a higher exposure of this age group, it is also true that increased frailty and fragility are associated to aging. Contemporary research has shown that should the injury risk of the elderly was similar to that of a 20-year-old car occupant, some 10,000 lives could be saved in the United States [5]. Therefore, given the existing trend of a growing proportion of elderly road users, it is mandatory to recognize their physiological differences and to incorporate their peculiarities into the design of more effective restraints. A review of AIS3+ injuries within NASS CDS shows that as age increases from 15 to 75+ years old, the incidence of thoracic injuries increases and becomes the most frequent serious injury among car occupants older than 46 years old, and accounting for more than 35% of all AIS3+ injuries in people aged 75 years and above [6].

Force-limiting seatbelts in combination with airbags have been shown to reduce significantly chest loading and consequently chest injuries, without increasing the risk of a head impact [7,8]. The combination of pretensioners and force limiters in rear seat restraints has been also shown to allow greater torso rotation of the occupant without increasing significantly the peak forward excursion of the head and therefore without increasing the risk of head contact [9,10]. In recent years, inflatable seatbelts have been discussed as a countermeasure that can reduce the risk of chest injuries even further by increasing the area of the torso on which the seatbelt force is applied and incorporating a damping effect in addition to the elastic component of conventional belt webbing [11-14]. These innovative restraints have been shown to modify substantially the kinematics of the occupants, challenging existing knowledge about the
optimal seatbelt geometry and position of belt anchorages [14]. With the continuous improvement of finite element human body models (HBM), parametric analyses constitute an effective methodology to optimize the design of innovative restraint systems. Real testing with Anthropomorphic Test Devices (ATD) also allows to confirm that existing testing tools are sensitive to newly designed restraint systems. However, it is also necessary that both surrogates (HBM and ATD) are able to capture the effects of these restraints on real human subjects. Thus, Post Mortem Human Surrogate (PMHS) testing is advisable to acquire a more accurate picture of the potential benefit that these system can bring to elderly car occupants.

The present study discusses the performance of the THOR ATD and the mechanical response of two PMHS exposed to a frontal impact and restrained by two different seatbelt solutions. The kinematics and dynamics of both surrogates will be compared with the goal of assessing if a new seatbelt concept is capable of reducing chest deflection without increasing other important performance indices.

METHODS

Eight sled tests in matching conditions were performed at the crash test facility of the Institute of Engineering Research (I3A) of the University of Zaragoza. Two Post Mortem Human Surrogates (PMHS) were exposed to a 35 km/h (nominally) frontal impact, using two different restraints. Then, the THOR dummy was exposed to nominally the same test conditions and using the same two types of restraints than in the PMHS tests. Three repeats were done per restraint type with the THOR dummy. The test matrix is shown in Table 1. The time history of sled deceleration is included in Figure A1 and Figure A2 in the Appendix.

<table>
<thead>
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<th>Occupant type</th>
<th># runs</th>
<th>Restraint</th>
<th>Impact speed (km/h)</th>
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<td>PMHS</td>
<td>1</td>
<td>SB</td>
<td>37.7</td>
</tr>
<tr>
<td>PMHS</td>
<td>1</td>
<td>PT+FL</td>
<td>34.6</td>
</tr>
<tr>
<td>THOR</td>
<td>3</td>
<td>SB</td>
<td>34.8±0.2</td>
</tr>
<tr>
<td>THOR</td>
<td>3</td>
<td>PT+FL</td>
<td>34.7±0.0</td>
</tr>
</tbody>
</table>

Test setup

The test fixture consisted of a rigid metallic frame allowing complete visual access to the occupant while preserving the basic geometry of a standard seating position of a passenger car. This test fixture has been used elsewhere as a reasonable approximation to the passenger posture in the study of ATD biofidelity and in the development of thoracic injury criteria [15,16]. The seat consisted of a flat steel plate with two pelvic supports at the rear. Forward motion of the occupant was restrained by two different seat belts: a retractor pretensioned (2 kN), force-limiting (4 kN) belt (PT+FL) and an innovative belt consisting on two independent shoulder and lap bands (SB). The SB shoulder belt band was retractor pretensioned (3 kN) and the lap belt band was pretensioned at 3.5 kN bilaterally. The position of the buckle of the lower shoulder belt band was forward of that of the inner buckle of the lap belt with the goal of unloading the lower part of the rib cage. As for the position of the D-ring, it was modified accordingly to the specific dimensions and anthropometry of the different surrogates to ensure that the upper shoulder belt angle measured between the shoulder of the occupant and the D-ring was the same. Belt tension was measured at several locations both on the shoulder and on the lap seat belts for each system. Sensor data in the ATD and PMHS tests were captured at 10,000 Hz with an external data acquisition system (PCI-6254, National Instruments; Austin, TX). The kinematics of the surrogates were recorded by a lateral and a frontal high-speed imager at 1,000 Hz.

THOR tests

The THOR ATD used in this study is the THOR-NT upgraded within the THORAX project (European 7th Framework Program) to improve biofidelity and injury assessment capabilities [17,18]. This THORAX THOR (further denoted as simply THOR) is similar to the U.S version THOR Mod Kit with SD3 shoulder.
ATD head was instrumented with a 6-degree-of-freedom (DOF) cube including three linear accelerometers and three angular rate sensors (ARS) oriented in perpendicular directions according to SAE J211 recommendations. Two 6-axis load cells were present at the upper and lower neck locations. Chest instrumentation included four 3D IR-TRACCs (Infra-Red Telescoping Rod for the Assessment of Chest Compression) and 72 strain gauges distributed from rib 2 to rib 7 bilaterally (equally spaced along the rib). The dummy was equipped with a 96-channel digital on-board data acquisition system to supplement the external one.

A high-speed motion capture system consisting of 10 cameras was used to track the position of retroreflective spherical markers within a calibrated 3D volumen at 1,000 Hz (Vicon, TS series, Oxford, UK). A calibration procedure, performed prior to testing, established the position and orientation of each camera in a reference coordinate system at a laboratory fix location. A local coordinate system moving with the test buck was defined so that the relative motion between the occupant and the buck could be resolved. The local X axis pointed forward in the direction of the sled motion and the local Z axis pointed upwards. Displacements were calculated with respect to the local coordinate system. Reflective markers were attached to several relevant anatomical locations in the dummy, although only the head ones (attached bilaterally and over the line passing through the center of gravity of the head) were used in this study.

PMHS tests

The two PMHS tests used in this study as comparison with the THOR dummy were carried out within the 7th Framework Program Marie Curie Action BIO-ADVANCE [19]. The procurement and handling of PMHS were done according to the internal procedures of the laboratory of the I3A (University of Zaragoza). These procedures were reviewed and approved by the laboratory Oversight Committee and by the Ethical Commission for Clinical Trials of Aragon (CEICA), which is the officially appointed regional committee that supervises the ethics of all clinical trials performed within the region. CEICA is totally independent from the University of Zaragoza and reports directly to the Health Commissioner of the Regional Government.

Triaxial accelerometers were rigidly attached to the head, T1, T8, L2 and the pelvis. ARS were added to the head and T1. Linear acceleration was measured at the location of the sternum body. Photo targets were used to identify relevant landmarks, including the location of the external auditory meatus bilaterally. These photo targets were used to track the displacement of the center of gravity of the head in the sagittal plane.

RESULTS

Upper shoulder belt force

Figure 1 shows the time history plot of the belt tension measured at the upper shoulder belt location in the THOR sled tests. Red solid lines correspond to the SB seatbelt while blue solid lines are the forces measured in the PT+FL belt tests. Apart from the greater pretensioner force of the SB seatbelt, two out of the three tests with the SB seatbelt resulted in a higher peak force than in the case of the PT+FL seatbelt (PT+FL: 5465.7 ± 212.7 N; SB: 6124.7 ± 706.6 N). Figure 1 also shows that the SB seatbelt induced a bimodal force curve on the ATD with the first and higher peak occurring at around 100 ms followed by a smaller peak at approximately 135 ms. The second maximum observed when the SB seatbelt was used contributed to the longer interaction between the occupant and the restraint: while the magnitudes of the force-time traces of the PT+FL seat belt are almost negligible at t=120 ms, the curves corresponding to the SB seatbelt indicate that the occupant was still being restrained by the seatbelt up to t=150 ms.

The time traces in Figure 2 show the time history of the upper shoulder belt forces in the PMHS tests. The peak force sustained by the occupant when the SB seat belt was used was higher than in the case of the PT+FL seat belt (6109.7 N vs. 4133.51 N), the longer interaction of the occupant with the restraint that was observed with THOR is also present in the PMHS tests, although there is no secondary peak in this case.
Figure 1 Time history of the upper shoulder belt force measured in the dummy tests

Figure 2. Time history of the upper shoulder belt force estimated in the PMHS tests.

Figure 3 THOR peak chest deflection in mm in the local x-direction (blue bars) and resultant (red bars) as measured by the IRTRACCs, with error bars corresponding to the standard deviation. Labels within the bars indicate the IRTRACC sensor recording the peak deflection.

THOR chest deflection

The peak chest deflection in the local x-direction (initially aligned with the motion of the sled, but moving with the ATD torso) was measured at the lower left IRTRACC when the PT+FL seatbelt was used while it occurred at the location of the upper left IRTRACC with the SB due to the higher forces applied to the shoulder of the occupant. In addition, the peak deflection was greater with the PT+FL seatbelt (35.9 ± 5.6 mm vs. 25.2 ± 0.8 mm). When the three components of the deflection were combined into the calculation of the peak resultant deflection, still the PT+FL seatbelt resulted in a greater magnitude (46.2 ± 3.4 mm vs. 32.7 ± 1.1 mm). However, the resultant peak deflection was measured at the lower left chest location regardless of the restraint used.

Head displacement in the sagittal plane

The displacement in the sagittal plane of the center of gravity of the head of THOR and the PMHS is shown in Figure 4 and Figure 5. Blue solid lines correspond to the PT+FL seatbelt and red solid lines to the SB seatbelt.
Peak forward displacements of the head center of gravity of THOR were not significantly different for the two restraints under analysis (PT+FL: 390.7 ± 11.1 mm; SB: 406.4 ± 6.7 mm). Figure 4 illustrates that the nature of the trajectory of the head center of gravity was similar under the action of the two different restraints. Other than slight differences in magnitude, it is not possible to identify different trends or behaviors. Regardless of the restraint, the head center of gravity moved straight forward and slightly downwards up to t=50 ms and then it started to describe a curvilinear trajectory.

As the statures of the two PMHS were different, trajectories were length-scaled to those of a 50th male percentile (nominally, 175 cm). Figure 5 shows the comparison of the scaled trajectories followed by the center of gravity of the PMHS head depending on the seatbelt used. The use of the SB resulted in a greater forward excursion (306.1 mm vs. 188.8 mm). The maximum vertical excursion of the two seatbelts were not substantially different. The shape of the trajectories differed importantly. While the use of the SB seatbelt resulted into an initial forward displacement of the center of gravity of the head that occurred parallel to the local X axis for about 70 ms, the use of the PT+FL seatbelt caused the PMHS head to describe a curvilinear trajectory from the first instants of the deceleration. The analysis of the high-speed video images indicated that while the SB seatbelt allowed the spine of the dummy to pitch forward and rotate, the PT+FL seatbelt impeded the flexion of the PMHS spine resulting into the curvilinear motion of the head center of gravity in which the lower neck acted as fulcrum of the rotation. There are two reasons that contribute to the differences observed in the kinematics of the spine of the PMHS. The first one is that the higher lap belt forces of the SB seatbelt facilitated that the pelvis of the PMHS almost did not move forward during the impact, while the pelvis of the PMHS that was restrained with the PT+FL slid during the deceleration over the flat surface of the seat. The forward pelvic motion caused the occupant to adopt a slouched position and prevented the torso from pitching forward as it would have been desirable. The second reason is that the analyses of the lower shoulder belt forces show that even if the peak force is much higher when the SB seatbelt was used (4729.0 N vs. 3461.6 N, see Table 2), the PT+FL lower shoulder belt force was higher up to t=70 ms and therefore the torso of the occupant was subjected to a higher force that could have impeded the rotation during the first stages of the deceleration. In fact it is around t=70 ms that the torso of the PMHS restrained with the SB belt stopped its rotation and the head started to describe a curvilinear trajectory.

**DISCUSSION**

The concept design of the split buckle restraint system (SB) intended to increase the loading on anatomical parts that are potentially more likely to bear higher loads (i.e. clavicle, pelvis) so that the amount of restraint loading on the chest could be lowered without modifying significantly the forward displacement of the head. The assessment of the system was done using two different occupant surrogates: the THOR dummy and PMHS in sled frontal impacts with a delta-v = 35 km/h (nominally). A pretensioned, force limiting seatbelt (PT+FL) was
used as reference for the assessment. The data generated in these tests can be used to benchmark numerical models, so that parametric studies can be done to find the optimal geometry of the split buckle seat belt [20].

The tests with the THOR dummy showed that the use of the SB belt system resulted into slightly higher upper shoulder belt forces, considerably higher lower shoulder forces and higher bilateral lap belt forces as well. However, resultant chest deflection and local x-axis chest deflection as measured by the IRTRACCs of the dummy were reduced for all measurement locations with the exception of the upper left IRTRACC, in which the magnitudes were not significantly different. Comparable results regarding belt forces were observed also in the PMHS tests, confirming the predictions given by THOR. Despite the associated higher SB shoulder belt, the duration of the engagement between belt and occupant was longer than in the PT+FL case, which resulted in smaller sternal deceleration of the occupant (see Table 2).

As for the displacement of the head in the sagittal plane, Figure 4 illustrates that THOR did not capture significant differences in nature between the two belt systems. In addition, the peak forward displacement of the center of gravity was similar regardless of the belt used (PT+FL: 390.7±11.1 mm vs. SB: 406.4±6.7 mm). The tests with the PMHS showed very different results in this case. While the SB system resulted in a trajectory in which the head moved forward first parallel to the local X axis and then underwent a curvilinear translation, when the PT+FL system was used, there was almost no forward motion of the torso of the PMHS resulting into a curvilinear translation of the head from the beginning of the deceleration. Consequently, the magnitude of the peak forward displacement of the head was smaller in this case (PT+FL: 188.8 mm vs. SB: 306.1 mm). However, the analysis of the high-speed video images showed that while the SB lap belt prevented the forward displacement of the PMHS pelvis, the PT+FL belt allowed the pelvis of the occupant to move forward. The forward motion of the pelvis associated to the lack of torso pitch are indicative of poor occupant kinematics that could result in submarining. Of course, the flat design of the seat is playing an important role in the deficient control of the motion of the pelvis as indicated in previous work using the same test fixture [21]. These differences in the kinematics of the pelvis and the torso were not observed in the sled tests with the THOR dummy. Even if the pelvis of the dummy moves forward during the initial instants of the deceleration until it is arrested by the lap belt, the torso of the dummy rotates up to the moment of maximum shoulder belt tension in which the head starts to undergo flexion. Regardless of the seatbelt used, the analysis of the kinematics of THOR does not indicate that submarining occurred in the tests.

The test fixture, except for the restraint type used, is a replica of that used in previous studies aiming to assess the biofidelity of THOR and to develop a new chest injury criterion for this ATD [22]. In the present study, the knee bolster used in [22] was removed. The only other set of PMHS and dummy tests that were run with this fixture without knee bolster compared the kinematics of the Hybrid with the kinematics of PMHS in frontal impacts conducted at 9 km/h [23]. In this case, non-pretensioned, non force-limiting seat belts were used. Recorded peak upper shoulder belt forces were in the range between 1000 kN and 1250 kN, and peak forward head displacement of the three tested PMHS ranged between 280 and 310 mm. Two remarks are relevant from this study. First, the nature of the motion of the head was similar to that observed in the present study for the SB seat belt: an initial trajectory parallel to the sled X-axis followed by a curvilinear trajectory once the upper shoulder force peaked. Secondly, while THOR predicted greater forward head excursion compared to the PMHS, the Hybrid III (178±2 mm) fell short to predict the displacement of the PMHS heads.

The magnitude of the upper and lower neck reactions (forces and moments) in THOR were similar regardless of the seat belt system used in the test. Measured values were considerably smaller than those proposed in existing IARV (Mx=143 Nm; My=190 Nm; Mz=96 Nm; Fz=4 kN; Fx=Fy=3.1 kN) [24]. Neck reaction forces were not calculated for the PMHS, but the comparison of the values measured for the acceleration and angular rate of the head indicated that the SB belt induced higher linear acceleration than the PT+FL seat belt, and consequently the upper neck loads would have been slightly greater.
Table 2.
Peak value comparison of selected parameters between the PT+FL and the SB belt.

<table>
<thead>
<tr>
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<th>PT+FL</th>
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<tr>
<td></td>
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<td>PMHS test</td>
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<tr>
<td>Head acceleration (g)</td>
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<tr>
<td>x</td>
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<tr>
<td>y</td>
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<td>PMHS test</td>
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<td>x</td>
<td>-22.7±1.4</td>
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<td>y</td>
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<tr>
<td>z</td>
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<tr>
<td>Head angular rate (deg/s)</td>
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<td>y</td>
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<tr>
<td>z</td>
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<tr>
<td>Fx (kN)</td>
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<td>-1.1±0.0</td>
</tr>
<tr>
<td>Fy (kN)</td>
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</tr>
<tr>
<td>Fz (kN)</td>
<td>1.4±0.1</td>
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<tr>
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<tr>
<td>My (Nm)</td>
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<tr>
<td>Mz (Nm)</td>
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<td>2.9±1.3</td>
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<tr>
<td>Lower neck</td>
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<td></td>
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<tr>
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<tr>
<td>Fy (kN)</td>
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<tr>
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<tr>
<td>Mz (Nm)</td>
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<td>Chest acceleration (g)</td>
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<tr>
<td>x</td>
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<td>y</td>
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<td>z</td>
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<td>Head maximum forward displacement (mm)</td>
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<td></td>
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<td>Upper Left</td>
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<td>-32.1±0.8</td>
</tr>
<tr>
<td>Lower Left</td>
<td>46.2±3.4</td>
<td>-32.4±1.0</td>
</tr>
<tr>
<td>Upper Right</td>
<td>31.7±2.9</td>
<td>-31.8±1.4</td>
</tr>
<tr>
<td>Lower Right</td>
<td>26.6±2.9</td>
<td>-21.4±1.0</td>
</tr>
<tr>
<td>IRTRACC (x direction, mm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Upper Left</td>
<td>24.2±0.5</td>
<td>-25.1±0.9</td>
</tr>
<tr>
<td>Lower Left</td>
<td>35.9±5.6</td>
<td>-24.5±1.8</td>
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<tr>
<td>Upper Right</td>
<td>17.0±14.3</td>
<td>-8.0±0.9</td>
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<tr>
<td>Lower Right</td>
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<tr>
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<td>Lower shoulder belt force (N)</td>
<td>3998.5±156.0</td>
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<td>Right lap belt force (N)</td>
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<td>Left lap belt force (N)</td>
<td>2059.3±542.6</td>
<td>3324.7</td>
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</table>

* Values for THOR shown as average ± standard deviation.
** Magnitude measured with respect to a local coordinate system with origin in the head attachment plate, not transferred to the head center of gravity.
Limitations
Even if the test fixture has been utilized before in the assessment of the biofidelity of ATD and in the development of thoracic ATD injury criteria, it is questionable its utility in the assessment of restraint systems. The simple geometry and the completely rigid structure of the seat facilitated the forward motion of the pelvis of the PMHS restrained with the PT+FL and the subsequent differences in torso pitch with respect to the other seatbelt solution. A regular production seat would have reduced the forward motion of the pelvis and contributed to increase the forward rotation of the torso.

Some channels of the external data acquisition system malfunctioned during the THOR tests. In particular, the upper shoulder belt load cell did not measure correctly the tension of the belt. To overcome this issue, a methodology was developed so that the magnitude of the belt tension at the upper shoulder location could be estimated using the measurement from the lower shoulder belt. The methodology is explained within the Appendix, and it was built using previously sled tests ran with both restraint systems. The correlation factor of the relationship found between the upper and lower belt tension magnitudes were $R^2 = 0.98$ (PT+FL) and $R^2 = 0.99$ (SB).

One additional limitation is the magnitude variability observed in the deceleration pulse between the two PMHS experiments that reached almost 4g at its peak difference. Although this difference would have hindered a detailed quantitative comparison between the performance of the two systems, the differences in the nature of the kinematics suggest that these differences were not related to just a change in the magnitude of the mechanical insult but to the way in which the restraints interacted with the PMHS. Interestingly, these differences were not observed when the surrogate chosen was THOR.

CONCLUSIONS
The present study compares the kinematics and dynamics of the THOR dummy and two PMHS in frontal impacts at 35 km/h using two different seat belts. One belt was a pretensioned force-limiting seatbelt that was used to benchmark a new concept consisting of two separate shoulder and lap bands, equipped with pretensioners at the shoulder belt retractor and at both lap belt anchorages. The aim of the SB model was to increase the load on the clavicle and pelvis to unload the chest region, without increasing significantly the forward displacement of the head comparing to the benchmark. The sled tests performed with THOR confirmed the intended performance of the concept belt and the chest deflection measured by the ITRACC was substantially lower in comparison with the PT+FL seatbelt. The two PMHS exhibited very different kinematics depending on the seatbelt used. While the PT+FL seatbelt allowed the pelvis to move forward, reducing torso pitch and therefore inducing a curvilinear motion of the head, the SB allowed substantial torso pitch resulting on increased forward excursion of the head that moved initially in a rectilinear fashion undergoing a curvilinear trajectory only once the torso motion was completely arrested. Given the limited number of PMHS tests, it is not possible to conclude if the particularities of the individual PMHS were the cause of the differences or if THOR failed to capture the kinematics of an actual subject under the action of these seat belts.

ACKNOWLEDGEMENTS
Autoliv Research collaborated in this study providing access to the THOR dummy and the restraint systems. It was partially funded by the Instituto Aragones de Fomento of Gobierno de Aragon via the “Collaborative agreement to foster research on impact biomechanics”, signed on Feb 11th, 2015. The PMHS tests were performed within the BIO-ADVANCE project (Marie Curie Actions, FP7/2007-2013, REA grant agreement no. 299298). More information about this project and its goals can be found in [19]. The analyses presented here were not generated during the project and do not represent necessarily the position of the researchers involved in BIO-ADVANCE. This study shows solely the interpretation of the authors and is not necessarily the view of Autoliv Research or the Instituto Aragones de Fomento.

REFERENCES


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APPENDIX

Sled deceleration

Figure A1. Time history of the deceleration of the sled. THOR tests.

Figure A2. Time history of the deceleration of the sled. PMHS test.

Estimation of upper shoulder belt forces (THOR tests)

The channel associated to the measurement of the tension at the upper shoulder belt location failed during the THOR tests. The input voltage fluctuated during the tests and, consequently, the measure of belt force was erroneous. To provide a reasoned estimation of the tension of the belt at this location, previous sled tests performed using the same deceleration pulse and the THOR dummy were used to find a correlation between the tension of the shoulder belt at the upper (close to the clavicle) and lower locations (close to the abdomen). It was found that a linear relationship could be established between these two measurements and therefore, the upper shoulder force could be estimated based on the measurement obtained at the lower location. The correlation factors obtained in the estimation of the upper shoulder belt force were $R^2 = 0.98$ (FL+PT belt) and $R^2 = 0.99$ (SB belt) (see Figures A3 and A4).

Figure A3. Linear relationship (blue solid line) existing between the shoulder belt forces measured at the upper and lower locations. FL+PT belt.

Figure A4. Linear relationship (blue solid line) existing between the shoulder belt forces measured at the upper and lower locations. SB belt.

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INVESTIGATION OF REAR SEAT OCCUPANT POTENTIAL INJURY RISK BASE ON SEAT BELT CONFIGURATIONS

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ABSTRACT

The occupants of all ages and sizes can be seated in the rear seats. But legal requirements regarding the qualification of the second seat row restraint system with anthropomorphic test devices (ATDs) currently do not exist. The protection of frontal seat passengers in both driver and front seated occupant has been more focused from the auto industries as well as regulatory bodies more than 40 years. Fortunately, their interests have been extended to rear seat occupants especially children and female occupants in recent years. However, the current available safety devices for the rear seat occupants are standard seat belt system only. Also, the majority of the rear seat occupant studies were focused to evaluate and protect child either CRS or using seat belt restrained in rear seat. The rear seat seemed to offer the greatest protection to children 0–12 years. Children seated in the rear seat had a lower risk of death compared with front seat passengers whether or not they were restrained. However, among adolescent and adult passengers, the rear seat offered less protection with increasing age and when restraints were used.

As a pilot project in Korea, total 452 accident cases have been collected and numbers of injured occupants (in patient) were 698. Drivers were 383 (54.9%), front passengers were 164, 2nd row left side seat were 47 (6.7%), 2nd row right side seat were 82 (11.6%), 2nd row middle seat were 15 (2.1%), and the remains are 3rd and 4th row seat occupants. Results from ISS injury severity analysis, the occupant of driver seating position has the highest ISS scores, 7.8±10.3, while front passenger (7.7±12.9), 2nd middle seat (6.3±7.7), 2nd left seat (6.1±9.3), 2nd right seat (6±11.7), 3rd left seat (6±0.0), 3rd middle seat (5±0.0), and 3rd left seat (2.8±1.7). Although the analysis was based on the limited numbers of data set, the safety of the rear seat adult occupants can’t be ignored. Especially, the majority of rear seat potential occupants may be vulnerable occupants such as child, female with children, pregnant woman, and elderly.

In this study, the rear seat belts anchorage locations of the current domestic passenger vehicles were investigated to evaluate the influence of rear seat belt anchorage geometrical configurations in terms of the rear seat passenger safety. The sled type simulation models are developed with three point belts are fitted on the Hybrid III 5th percentile dummy and Hybrid III 50th percentile dummy. The injury value, particularly HIC15 and Chest deflection were examined to evaluate the contribution of rear seat belt anchorage locations.

INTRODUCTION

Accident statistics over the last decades have shown a continuous reduction of killed and severely injured in the road traffic accidents. Historically the driver and front seat occupant have been the highest priority because these positions are occupied most often and therefore account for the greatest numbers of casualties. The introduction of airbags, seat belts and other advanced safety systems in front seating positions, along with vehicle structural improvements, has led to a significant reduction in the number of casualties and fatalities among vehicle occupants.

This development was driven forward by new legislative requirements and the introduction and continuous progress on worldwide consumer test programs like the Euro-NCAP. The user’s consciousness on safety is continually increasing due to publications and public discussion of road safety issues. Car manufacturers, in cooperation with suppliers, have taken massive action in order to achieve a top rating in consumer tests. The equipment rate of active and passive elements is steadily increasing and allows predicting further positive effect on road safety for the future.

In newer model vehicles, occupant protection is achieved through the utilisation of a ‘system’; a combination of vehicle crush characteristics, enhanced seat and seat belt technologies, such as seat belt pretensioners and load limiters, and airbags. This has largely been a result of consumer and regulatory assessments evaluating the performance of the system, rather than regulating the presence of any single component. Since these assessment
programs evaluate the protection provided in the front seat only, there has been less motivation for vehicle manufacturers to develop and implement these technologies in the rear seat, and the extent to which such technologies are available in the rear seat is not documented. Less attention has been paid to the rear seat. It was due to the rear seat has been considered safer than the front seat. Recently, however, a number of recent studies have been conducted showing instances of lower levels of protection for rear seat occupants compared to front seat occupants in the frontal crash test.

On the rear seat, the occupants of all ages and sizes can be seated. But the legal requirements regarding the vehicle safety issues of the second seat row restraint system with anthropomorphic test devices (ATDs) currently do not exist. Only consumer tests such as EuroNCAP, JNCAP and C-NCAP with respect to rear occupant protection systems have been adopted as a part of safety performance evaluations. Several recent publications discuss the passenger safety of the second row. Kuppa et al. indicates higher mechanical loads on rear seat passengers during a crash and deduce a higher injury risk compared to drivers and passengers in the first row. Restraint components like inflatable cushions (airbags) in order to protect the head and thorax or the lower extremities as well as pyrotechnical pre-tensioners partly with multi-stage load limiters for belt retractors are standard equipment in the front seat row. The next generation of advanced restraints system for first row occupants is already under development. Individualized restraint systems, like those providing adaptive pressure control of the airbag pressure and multi-stage belt force limitation concepts, are pending market introduction. These systems enable tailored restraint performance depending on crash severity and occupant size. In contrast to this, a 3-point belt retractor without pretension and force limitation is still the standard for the back seat passengers.

Consumer protection organizations incorporated adult passengers on the rear seat in their frontal test programs. The Hybrid III ATD with the 5th percentile is already an element of a test configuration for China-NCAP and Japan-NCAP. Euro-NCAP has been announced a follow up in 2015. From 2014 national survey of seat belt usages, the wearing rate of diver and front passenger was 78%, but only 22% of rear occupant was belted. It is significant improved from 2011’s 5% rate of the belted rear seat occupant. To further promote usage of rear seat belts, the traffic regulation for wearing all seated passenger is effected in 2015. According to the survey, the one of main reason for unbelted was inconvenience of fitting and geometrical difficulty of buckle-up in the rear seat. To protect the rear seated occupants, MLIT granted a research project as a part of ASV program. This research program is part of a project intended to understand the optimal rear seat environment for rear seated adults and old child passengers who use the vehicle belt for primary restraint. Therefore, the objectives of this study were development of test protocols for rear seat occupant protection in future KNCAP plan. The total domestic 39 passenger vehicles including SUVs of the second rows seat’s belt-anchorage geometrical configuration are investigated and compared with the KMVSS 102 (similar to FMVSS 210) criteria. To estimate injury risk of rear seat occupant, the sled type frontal crash simulations has been conducted with 5th female H3 and 50th male H3 dummies based on KNCAP crash pulse.

ACCIDENT DATA ANALYSIS

A numerous researchers have been studied to focus on the relationships between the safety of the front and rear seat passengers. The most of these studies were limited to evaluate and protect child either CRS or adult seat belt restrained in rear seat. On the basis of the analysis of injuries of the road accident victims (injured and killed), it has been estimated that the risk of death of a rear seat passenger was smaller by 26–41% compared to a front seat one, even without the seat belts. On the average, the risk of death was 21% lower among passengers in the rear seat compared with front seat passengers. This apparent protection varied with age, restraint use, and airbag presence. The rear seat seemed to offer the greatest protection to children 0–12 years. Children seated in the rear seat had a lower risk of death compared with front seat passengers whether or not they were restrained. Among adolescent and adult passengers, the rear seat offered less protection with increasing age and when restraints were used. Restraints offered more protection to both front and rear seat passengers in vehicles with a front passenger airbag. In addition, restraints offered greater protection to front seat passengers compared with rear seat passengers.

Improving road safety, the first step in the process is identifying significant safety enhancement areas and the mechanisms of accidents and/or injuries that govern the problem. As a pilot project, total 452 accident cases have been collected from 3 different regional hospitals during last 11months period of traffic accident investigations. The sizes of 3 cities are from 210,000 to 450,000 populations which are typical medium size city. The total collected numbers of injured occupants (in patient) were 698. Drivers were 383 (54.9%), front passengers were 164, 2nd row left side seat were 47 (6.7%), 2nd row right side seat were 82 (11.6%), 2nd row middle seat were 15 (2.1%), and the remains are 3rd and 4th row seat occupants. Results from ISS injury severity analysis, the occupant of driver seating position has the highest ISS scores, 7.8±10.3, while front passenger (7.7±12.9), 2nd middle seat (6.3±7.7), 2nd left
seat (6.1±9.3), 2nd right seat 6±11.7), 3rd left seat (6±0.0), 3rd middle seat (5±0.0), and 3rd left seat (2.8±1.7) as shown in Table 1. In this analysis, un-belted occupants were not discriminated. Although the analysis was based on the limited data set, safety of the rear seat occupants can’t be ignored. Therefore, the safety of rear seat occupant is equally important and must be seriously considered in terms of overall vehicle safety concepts from all related stock holders such as government body, auto makers, insurance agencies, and NGO. Especially, the majority of rear seat potential occupant may be vulnerable occupants such as child, female with children, pregnant woman, and elderly.

<table>
<thead>
<tr>
<th>(n, %)/Seating position</th>
<th>Driver</th>
<th>F. Pass</th>
<th>2nd row Left</th>
<th>2nd row Right</th>
<th>2nd row Middle</th>
<th>3rd row Left</th>
<th>3rd row Right</th>
<th>3rd row Middle</th>
<th>4th row Left</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total (698, 100)</td>
<td>383</td>
<td>(54.9)</td>
<td>164</td>
<td>47 (6.7)</td>
<td>81 (11.6)</td>
<td>15 (2.1)</td>
<td>1 (0.1)</td>
<td>4 (0.6)</td>
<td>1 (0.1)</td>
</tr>
<tr>
<td>ISS (mean±std)</td>
<td>7.8±10.3</td>
<td>7.7±12.9</td>
<td>6±11.7</td>
<td>6±9.3</td>
<td>6.3±7.7</td>
<td>6±0.0</td>
<td>2.8±1.7</td>
<td>5±0.0</td>
<td>1.5±0.7</td>
</tr>
<tr>
<td>ISS (median)</td>
<td>4</td>
<td>3</td>
<td>3</td>
<td>2</td>
<td>5</td>
<td>11</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>ISS (interquartile range)</td>
<td>2.98</td>
<td>2-12.3</td>
<td>2-10.3</td>
<td>1-3</td>
<td>2-11</td>
<td>3-22.8</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

LEGAL REQUIREMENTS FOR REAR SEAT BELT ANCHORAGES

Recent work at the University of Michigan Transportation Research Institute (UMTRI) has demonstrated that belt anchorage locations have a strong effect on occupant kinematics. More-rearward or higher lower anchorages (flatter lap belt angles) tend to reduce lower-body excursion, except when submarining occurs due to the belt slipping off the pelvis and into the abdomen. More-forward and higher upper anchorage locations tend to increase head excursion, except when submarining occurs.

In Korea, the locations of the anchorages of seat belt systems are regulated by Korea Motor Vehicle Safety Standard (KMVSS) 102 which is similar to FMVSS 210. Anchorage locations are referenced to the seating reference point (SgRP), which is measured using the SAE J826 H-point manikin. Lap belt anchorages must be positioned such that a vector from the anchorage to the SgRP in side view forms an angle of between 30 and 75 degrees with the horizontal. Upper anchorages must be located within a side-view zone defined with respect to the SAE J826 two-dimensional template with the template H-point aligned with the SgRP.
SIMULATION MODEL FOR REAR SEAT OCCUPANT

To assess safety of rear seat occupants, sled type simulation was conducted with calculated average value of 2010-2012 KNCAP 33 tested vehicles crash pulses to normalize crash severity as shown in Fig 4. The physical rear seat of each vehicle was manufactured differently due to different vehicle configuration as well as characteristic of vehicle model.

In this simulation, the rear seat was modeled according to UN R44 as a standard seat model in order to eliminate seat variation, only focus to observe influence of seat belt anchorage configurations. The standard rear seat and polyurethane foam were modeled as shown in Fig 5. The webbing of seat belt was modeled as having 7% elongation material property. The pre-tensioner and load limiter of rear seat belt were not available for most of current production vehicle. These devices were not modeled.

Also, in the simulation model, to examine kinematics of occupants and injury levels of the different types of dummies during the crash event, the front seat was modeled as shown in Fig 6. Injuries of 5th H3 and 50th H3 dummies were compared with different seat belt anchorage locations.
RESULTS

In this study, the total 39 passenger vehicles (sedan and SUV) from the 5 domestic car manufactures were examined. These vehicles were consisted in 3 sub-compacts, 3 compacts, 11 mediums, 9 large vehicles, and 12 SUV vehicles. In geometric data collection process, the coordinate of upper and lower anchorages were calculated from CAD data without any physical measurements due to availability of all vehicles. Fig. 1 shows the current KMVSS 102 requirements for the rear seat belt anchorages.

Table 2 and Fig 2 and 3 show the calculated seat belt anchorage locations relative to SgRP according to KMVSS 102 allowable zones. The red dots were average locations of examined vehicles. The lower anchorage locations scattered the entire allowable range of side-view angles. Upper anchorage locations also spanned a wide range, reflecting the wide variety of upper belt anchorage configurations, including those mounted in the C-pillar, package shelf, and integrated into the seat. But, the distributions of Z directional range were relatively narrowed due to accommodate from 5th H3 to 95th H3 dummies. Inboard, buckle-side lower anchorages had slightly steeper (higher) lap belt angles, and the inboard anchorage was generally closer to the occupant centerline than the outboard anchorage. Since the distributions of all 39 vehicle’s anchorages location were calculated only based on the CAD geometric data, it is not intended to verify fulfillment of requirements of regulation for out-ranged vehicle’s anchorage locations.

Table 2 Relative coordinates of rear seat belt anchorage points (reference point: SgRP in mm)

<table>
<thead>
<tr>
<th>vehicle Type</th>
<th>D-ring Coordinate</th>
<th>Lower Outboard</th>
<th>Lower Inboard</th>
<th>Seat Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X</td>
<td>Y</td>
<td>Z</td>
<td>X</td>
</tr>
<tr>
<td>01</td>
<td>Small</td>
<td>-679.4</td>
<td>-216.5</td>
<td>471.1</td>
</tr>
</tbody>
</table>
YOUN 6

Fig. 2 Side views (y direction) of rear seat belt upper anchorage locations
DISCUSSIONS

Geometrical distribution of seat belt anchorages

Normally, the belt geometry in rear seats is less favorable than in front seats. This is due on the one hand to vehicle geometry, which does not permit optimal belt-anchoring points, for example, the rear wheel house restricts the possibilities of fastening the anchor fitting. On the other hand, the rear seat running all the way through results in restrictions in fastening buckles. The seat position of the occupant in the rear seat is also different from in the front seat. Due to the restricted foot space extending to the front seat, the knees bend further causing the pelvis to tilt further backwards. The geometry of belts and the seat position result in the angle between the lap belt and the pelvis normally becoming comparatively small. As a consequence, the risk in a head-on collision of the lap belt slipping over the wings of ilium is evident, i.e., submarining can occur. The upper fastening point of the shoulder belt, which has frequently been positioned far to the rear, also promotes submarining.

In this study, the second-row belt anchorage locations for 39 domestic passenger vehicles using CAD geometric data were examined. The results indicate that seat belt anchorage locations in second rows vary widely. The range of upper anchorage locations were relatively widely spread in X direction compared with Y and Z directions as shown in Fig 7. In Z direction, the most of SUV type vehicles were in higher location of upper anchorage points. Due to absence of separated trunk wall in SUVs, the upper anchorage points were attached to the C-pillar. In general, upper anchorage locations may affect the risk of torso rollout in frontal crashes and also can contribute to discomfort due to contact between the belt and the occupant's neck if the anchorage is too high or too far from inboard. The fore-aft location of the anchorage can also affect head excursion and chest loading in frontal crashes.

![Fig. 7 Distribution of upper anchorage points in y-x and z-y planes](image.png)
Fig 8 shows the range of lower anchorages, which spans the allowable range, is of particular concern for belt performance in frontal impact. In general, flatter belt angles are associated with better pelvis restraint but a higher likelihood of submarining, although submarining is influenced by many other factors. Steeper lap belt angles are needed to prevent submarining for smaller occupants, particularly children seated on vehicle seats with long seat cushions that produce slouched postures.

![Outboard Anchorage Point (X-Y Plane)](image1)

![Outboard Anchorage Point (Y-Z Plane)](image2)

![Inboard Anchorage Point (X-Y Plane)](image3)

![Inboard Anchorage Point (Y-Z Plane)](image4)

Fig. 8 Distribution of lower anchorages points in x-y and z-y planes

**Dummy seating position with different seat belt fitting**

As shown in Fig. 9 and 10, the seated position of 5%tile and 50%tile dummies were based on case 28 configuration which is foremost X direction of upper anchorage. The shoulder belt can’t hold firmly upper torso of female dummy compared with same upper anchorage for 50%tile male dummy.

![5%tile female dummy Seated configuration of case 28](image5)

Fig. 9 5%tile female dummy Seated configuration of case 28
Injury assessment for rear seat belt anchorage location

Results from the sled simulation, injury values reveal that the most dominate factor is x directional location of upper anchorage point both in 5th H3 and 50th as shown in Table 3, Fig. 9 and 10. The more rearward locates upper anchorage point shows the less HIC values. Since regulation required at least suitable for 5th female dummy, less affecting coordinate in HIC is z directional point. For the chest deflection, also x coordinate of upper anchorage is influencing factor.

Table 3 Injury assessment for different rear seat belt anchorage points

<table>
<thead>
<tr>
<th>Type</th>
<th>Hybrid III 5%tile Female Dummy</th>
<th>Hybrid III 50%tile Male Dummy</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Head</td>
<td>Chest</td>
</tr>
<tr>
<td></td>
<td>HIC</td>
<td>Deflection (mm)</td>
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<td>02</td>
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</tr>
<tr>
<td>03</td>
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</tr>
<tr>
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<tr>
<td>05</td>
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<tr>
<td>22</td>
<td>Medium</td>
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<tr>
<td>23</td>
<td>Medium</td>
<td>667.0</td>
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<tr>
<td>24</td>
<td>Large</td>
<td>966.6</td>
</tr>
<tr>
<td>25</td>
<td>Large</td>
<td>644.1</td>
</tr>
<tr>
<td>26</td>
<td>Large</td>
<td>524.2</td>
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<td>27</td>
<td>SUV</td>
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<td>38</td>
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<td>880.7</td>
</tr>
<tr>
<td>39</td>
<td>SUV</td>
<td>952.5</td>
</tr>
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</table>

Fig. 9 Injury severity of 5%tile female dummy based on distribution of upper anchorage locations

Fig. 10 Injury severity of 50%tile male dummy based on distribution of upper anchorage locations
CONCLUSIONS

A significant increase has been observed in the number of installed restraint technologies available to front seat occupants as compared to rear seat technologies over the last couple of decades. With current research suggesting that the front seat is becoming relatively more effective at mitigating injuries [6-9], there is scope to improve the rear seat occupant protection system. The implementation of already available supplemental restraints available in the front seat and/or novel technologies, are potentially effective methods of improving safety in the rear seat.

In this study, the second-row belt anchorage geometric data were examined. Based on the geometric locations of rear seat belt anchorage, a standard frontal impact simulation was conducted to assess the influence of these belt anchorage points in rear seat occupant between 5%tile female H3 dummy and 50%tile H3 male dummy. From results from geometric calculation, the rear seat belt anchorages were widely scattered especially in X directional upper anchorage point. Some of vehicles were even out ranges of regulation requirements. The results indicate that seat belt anchorage locations in second rows vary widely. The range of upper anchorage locations were relatively wider spread in x direction compared with y and z directions. In injury assessments, the upper anchorage point strongly influences the motion of upper body. Results from the sled simulations, injury values reveal that the most dominate factor for HIC is x directional location of upper anchorage point both in 5th H3 and 50th. The more rearward location of upper anchorage point shows the lower HIC values. Since regulation required at least suitable for 5th female dummy, less affecting coordinate in HIC is z directional point. For the chest deflection, also x coordinate of upper anchorage is influencing factor.

ACKNOWLEDGEMENTS

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[4] Stegmeier M., Bausch H. Rear Seat Protection for Occupants Sizes from Q6 to HIII 95%ile in Frontal Crashes 9th International Conference Protection of Children in Cars, 2011
Preliminary Study of Roof Airbag Protecting Rear-Seat Occupants in Frontal Impact

Chun-Tao, Wu
Kai, Zhang
Great Wall Motors Co., Ltd.
People’s Republic of China

Paper Number 15-0130

ABSTRACT

A kind of innovative Airbag, Roof Airbag (RAB), with external straps is studied via virtual engineering, which provide protection to rear-seat occupants in frontal impact. Frontal sled model with AF05 female dummy is generated in LS-DYNA, which is well correlated with full vehicle frontal crash tests in terms of the kinetics and injuries of the dummy. In addition, it is adapted for AF50 and AF95. Based on above models, different configurations of restraint system are studied, different levels of load limiter, belt with or w/o pretensioner, with or w/o RAB for instance. It can be summarized that roof airbag with low level of belt load limiter and pretensioner could provide protection to most size of rear-seat adult occupants as good as that of front-seat occupants in frontal impact.

0 INTRODUCTIONS

Frontal Airbag, Driver Airbag (DAB) and Passenger Airbag (PAB), has been fitted in most of the vehicles as standard safety restraints, which provide inevitable protection to drivers and passengers in frontal accidents. This is benefitted from the requirements of regulations and consumer metric protocols, such as NCAP. It is reported that 40% fatal and serious injuries is caused by frontal accidents for rear-seat occupants[4]. However, for rear seat occupants, such helpful restraint equipments are not fitted in any commercial vehicle, yet. Anyway, the protection to rear seat occupants in frontal impact has been paid more attentions since last decay. NHTSA funded Rear Seat Occupant Protection Research Program. In 2005, Kuppa[1] and etc. reported that protection performance of rear-seat occupants is related with the age of the occupants in frontal impact. For the children and young adult people, age below 50, it is safer seated in rear seats than in the frontal-row seats, which is opposite for older people. It may be caused by misuse of frontal airbag. In 2007, Richard Kent[2] and etc. revealed that newer vehicle models, after 1999, with better restraints has improve the benefit to occupants younger than 50 seated in frontal seats compared with vehicle models before 1999, which is similar to those seated in rear seats. Meanwhile, Koji Mizuno[3] and etc. carried out couples of full vehicle frontal impact tests at 50km/h, which emphasis the necessaries of using seatbelt to avoiding fatal injuries during frontal impact for adults and children. Lotta Jakobsson and her partners have introduced integrated booster cushion with progressive torsion bar of seatbelt to provide protection for growing Children in frontal accidents[5]. Katarina Bohman and etc. has studied the rear seat child protection due to different type of seatbelt retractors, which shown that seatbelt with load limiter and pretensioner can not only constrained the occupants well to avoiding secondary impact but also reduce the injuries by 10% to 40%[6]. Above development and research work has consolidated current vehicle fitments for protection to rear seat occupants in frontal impact.

1.0 ROOF AIRBAG WITH EXTERNAL STRAPS

Seiji Aduma[7] and etc. has studied airbag integrated with seat belt via virtual simulation and sled test, which would provide improved protection performance in terms of head and neck, fatal injuries reduced by 30% appropriately. For chest injuries, the protection of the airbag is not mentioned in the paper, which takes 35% of the fatal locations of human body. Seatbelt Airbag has advantages in no requirements of mounting or supporting system in front of occupants, which is different from traditional airbag system, shown as Figure 1. Unfortunately, it has not been implemented to automotive industry to improve frontal protection to rear seat occupants, yet.
1.1 DESIGN IDEA

According to the Rating results released by EuroNCAP, some vehicles could achieve merely maximum score, 98.8%, in frontal impact, which could be considered as perfect protection. Frontal airbag, belt load limiter, belt pretensioner and even knee airbag could be used to presuming better protection for front-seat occupants. Due to lacking head restrained, for the rear-seat occupants low level of belt load limiter is not recommended in risk of secondary contact to frontal seats. To absorb energy of the head of occupants, the airbag for rear-seat occupants should be constrained in the front of the occupants. Generally, there are frontal seats for outboard occupants. It has been designed as container of rear-row airbag. The position of frontal seats are not fixed, which cause challenges to restraint system as smart as possible. The system needs to recognize the position of the frontal seats and accordingly the airbag can be scaled to interact with the occupants well. Here, a roof airbag with external straps is studied, shown in figure 2. The housing of airbag is fixed in frontal roof cross member and the cushion attaches to external straps. The rear end of straps is fixed at rear roof cross member. So, the external straps can constrain the airbag when it interacts with rear-seat occupants in frontal impact. The airbag does not need any external support, supported by frontal seat back for instance. It covers the shortage of the head restrained in frontal impact.
The influence by add-in the roof airbag to occupant protection is discussed in the following paragraphs via virtual engineering.

2.0 SLED MODEL GENERATION

To carry out virtual engineering of restraint system development, the models should be correlated overall, from materials to parts, subsystem and full models[8]. Regarding rear occupants protection, the sub system including compartment, seats, seat belts, occupants and the roof airbag, shown as figure 3, which is generated within LS-DYNA.

![Rear seat occupant protection model for frontal impact](image)

Figure 3 Rear seat occupant protection model for frontal impact

2.1 Full Vehicle Frontal Impact Tests with HIII 5% Female Dummy

According to the protocol of C-NCAP[9], full-wrap frontal barrier impact at 50km/h, called FFB, and 40% overlap frontal deformable barrier impact at 64km/h, called ODB, are carried out with one under development vehicle model. HIII 5% Female dummy is seated at rear seats in both tests, shown as Figure 4. The Seatbelt is normal emergency locking retractor with extra high level of belt load limiter, about 5kN.

![Full vehicle test with AF05 female dummy](image)

Figure 4 Full vehicle test with AF05 female dummy

The injury possibility of dummy in both tests is shown as figure 5. Here, the injury possibility of individual location is assessed by the method as the same as that of US-NCAP[10]. The joint injury possibility is about 41.5% in FFB at 50km/h and 31.2% in ODB at 64km/h respectively. It is reasonably that the injury possibility of lower limbs is neglect-able, 0.0%, resulted from merely no hazardous deformation to rear occupants in frontal impact.
2.2 Correlation of Sled model with HIII 5% Female Dummy

Based on the data of accelerometers from the tests, the motion of the vehicle is assigned to the virtual model, shown as figure 3. Correspondingly, the seatbelt and position of dummy is adjusted as that of tests. The results of simulation are compared with tests are shown as Figure 6, which shown good correlation level between simulation and tests in FFB.

![Graph showing comparison between sled model and test results for different body locations.]

**Figure 6 Injury and kinematics of AF05 predicted by sled model compared with that of test in FFB**

The comparison of injury possibility between sled model and test are shown in figure 7. In FFB, the error of body locations predicted by sled model is less than 4%, which resulted in good joint injury possibility, 46.1%, error about 3.6%. In ODB, the injury possibility predicted by sled model is about 26.1%, 6.1% less than that of test. The injury possibility of chest predicted by sled model is 8% lower than that of test. Except for the chest, the injury possibility of other locations has good accuracy, error less than 2.4%. The error of injury possibility predicted by sled models is shown in table 1. The accuracy of sled models is acceptable to study parameters of restraint system.
Figure 7 Comparison injury possibility of rear-seat AF05 in frontal impact between sled models and tests

Table 1. Error of ASI 3+ possibility predicted by sled models compared with tests

<table>
<thead>
<tr>
<th>Cases</th>
<th>Head</th>
<th>Neck</th>
<th>Chest</th>
<th>Femur</th>
<th>Joint</th>
</tr>
</thead>
<tbody>
<tr>
<td>FFB</td>
<td>-0.1%</td>
<td>4.0%</td>
<td>1.1%</td>
<td>0.1%</td>
<td>3.6%</td>
</tr>
<tr>
<td>ODB</td>
<td>-1.3%</td>
<td>2.4%</td>
<td>-8.9%</td>
<td>0.0%</td>
<td>-6.1%</td>
</tr>
</tbody>
</table>

2.3 Adapted Sled models with HIII 50% and 95% Dummies

For the rear occupants, not only female dummy seated, normal gentle man and bigger size of occupants seat there in real life. Respectively, AF50 and AF95 dummy are adapted to the correlated sled model. The injury possibility of AF50 predicted by sled models is about 24.5% in FFB and 20.9% in ODB respectively. That of AF95 is about 18.8% in FFB and 14.0 in ODB respectively. Here, the injury limits of AF50 are used for assessment of AF95. The injury possibility of AF50 and AF95 is shown as Figure 8.

Figure 8 Injury possibility of AF 50 and AF95 in frontal impact at rear seat predicted by adapted sled models.

As the dummy size increased, the trend of injury possibility shows reduction, which is appropriate to common sense, shown as figure 9. How to provide good protection to smaller female occupants will be challenge for current restraint system. FFB load cased and AF05 will be focused in following paragraphs to improve performance of protection to rear-seat occupants.
3.0 RESTRAINT SYSTEM STUDY

In 2013, Jeongkeun Khim[11] and etc. studied protection performance of different kind of seatbelt to rear occupant, AF 05 Female, via correlated sled model within LS-DYNA, which prepared for EuroNCAP FFB implemented in 2015[12]. It is reported that the pretensioner with constant load limiter will improve the dummy injuries in reduction of HIC15, Neck tension and Chest compression. As the load limiter level reduce, the chest displacement will be reduced as well, which are coincident between simulation and tests.

3.1 Affect of Load Limiter Levels

Based on correlated sled model with AF05, different levels of load limiter are studied. The injury possibility is shown as figure 8. The trend of injury possibility shows reduction overly, including head, neck and chest, as the level of load limiter decreased. The joint injury possibility ranges from 32.5% to 50.2%.

![Figure 10 ASI 3+ possibility of rear-seat AF05 in FFB predicted by sled models regarding different level of load limiter.](image)

However, the joint injury possibility is still very high, more than 30%, even for vehicles fitted with low level of belt load limiter of emergency retractor.

3.2 Affects of Retractor with or w/o Pretensioner

Pretensioner is common for front seats, which can constrained the occupants earlier during impact compared with emergency seat belt. Different levels of load limiter with pretensioner are studied in the sled models. As load limiter level reduced, the trend of injury possibility shows reduction overly, which is similar to the trend of emergency seatbelt, shown as Figure 11. The injury possibility can be reduced to about 20% for rear seat AF05 using low level of load limiter seatbelt with pretensioner. In addition, adding pretensioner can reduced the injury possibility of AF05 by 10% to 20% at least comparing with corresponding load-limiter level of emergency seatbelt, shown as Figure 12.
However, the lower level of load-limiter, the dummies will move more displacement. The bigger size of occupant, the dummies will move more displacement as well. For AF05, the low level of load limiter seatbelt can be fine, there is no secondary contact. According to the results of simulation, the secondary contact can be observed for AF50 and AF95 with low level of load limiter seatbelt, shown as Figure 13. The joint injury possibility is increased by 4.6% from 14.3% to 18.9% for AF50 and by 11.2% from 12.9% to 24.0% for AF95 respectively due to secondary impact, shown as figure 14. The fatal injury comes from head and neck. For AF50, the injury possibility is increased by 3.7% for head and by 2.6% for neck respectively. For AF95, bigger size body, the injury possibility of head is increased much more than that of AF50, mediate body, 11.1% increased. For bigger size of occupants, the injury may severer using seatbelt of low-level load limiter and pretensioner in risks of secondary contact.
3.3 Affects of Roof Airbag

As discussed above, the belt with low level of load limiter and pretensioner could provide better protection to rear seat occupants if there was no secondary contact, which is not suitable for most of current commercial vehicle due to constraint of the space. The roof airbag can protect the head from secondary contact even working with low level of belt load limiter. Interaction between roof airbag and different size of dummy is illustrated in Figure 15. There is no secondary contact observed. The ASI 3+ possibility of rear-seat occupants is reduced to 14.5% for AF05, 12.5% for AF50 and 10.1% for AF95 respectively, shown as Figure 16.

Regarding risks of secondary contact, compared with that of w/o roof airbag, the joint injury possibility of occupants protected with Roof Airbag is reduced by 6.3% for AF05, 6.4% for AF50 and up to 13.9% for AF95.
respectively, shown as figure 17. Restraint system having low-level belt load limiter, belt pretensioner and roof airbag can provide better protection to rear-seat occupants in frontal impact.

![Graph showing benefits to ASI 3+ possibility of rear-seat AF05, AF50 and AF95 due to Roof Airbag in frontal impact predicted by sled models.]

**Figure 17 Benefit to ASI 3+ possibility of rear-seat AF05, AF50 and AF95 due to Roof Airbag in frontal impact predicted by sled models.**

### 3.4 Engineering Approach of Roof Airbag

#### 3.4.1 Interaction Stability

In ODB impact, the benefit of protection to rear-seat occupants maybe not as good as that in FFB impact. Due to the rotation of vehicle, the head may rotate and pass by the side of the Roof airbag, shown as figure 18. The rotation of head may result in severer neck injury. It could be solved by modifying the shape of the airbag.

![Images showing head interactions with ODB airbags: (a) Head of AF05 passes by roof airbag, (b) Head of AF05 is restrained by roof airbag.]

**Figure 18 Head of AF05 interacts with roof airbag in ODB impact.**

#### 3.4.2 Customer Survey

The idea has been shown on The 3rd Technology Day of Great Wall Motors at first time, which was held in October of 2014, shown as figure 19.

![Image of roof airbag sample shown at the 3rd Tech. Day of Great Wall Motors.]

**Figure 19 Sample of Roof Airbag shown on the 3rd Tech. Day of Great Wall Motors**
Meanwhile, customer survey has been conducted on the show. The questionnaire includes seven closed questions, covering users’ safety consciousness, recognition of design purpose, protection effects, demanding, numbers, impact by sunroof and price acceptance, and one open question for suggestions. 300 pieces of sheet were finished during the events. According to the investigation data, the typical number of Roof Airbag is about 2.5, which depends on the vehicle seats. As concerned, the demand of sunroof and roof airbag is about 40% and 60% respectively, shown in figure 20. More surprising, a reasonable price is appropriately accepted by customers to fit Roof Airbag, which reduces the pressure of cost due to low volume at early stage.

![Sunroof vs. Roof Airbag](image)

**Figure 20 Demand of Roof Airbag vs. Sunroof**

### 3.4.3 Package, Transportation and Hazardous deployment

The airbag should cover most sizes of occupants. The shown sample can covers typical short female adult, nominal 50% and 95% male adults, shown as Figure 21.

The external straps need to be hidden inside of roof interior to minus impact on styling. Additional effort is needed to package the module during transportation.

In real life, people may sit forward, which is out of position. In this case the airbag deployment will be hazardous to occupants. Seatbelt reminder will remind people to be belted, when vehicle drive forward. Intellectual system is needed to suppress the roof airbag in case the deployment being hazardous to occupants.

![Deployment of roof airbag](image)

**a) Anologous 5% female adult b) Anologous 50% male adult c) Anologous 95% male adult**

**Figure 21 Deployment of roof airbag covers most of general rest-seat occupants**

### 4.0 DISCUSSION

Through out the world, the regulations and mandatory standards require vehicle providing protection to front-seat occupants in high speed frontal impacts. Lots of attentions has been paid to frontal-row occupants protection in frontal impact. However, the requirements of protection to rear-seat occupants are required mostly by cosumer protocols. In 2004, EuroNCAP started to assess vehicle performance of child protection in frontal ODB impact. In 2012, C-NCAP started to have AF05 in rear seat in frontal impact and the head, neck and chest injury is assessed. In 2015, FFB is integrated to EuroNCAP protocol, which focus on protection to AF05. The head, neck, chest and femur will be assessed and the limits is more serious than that of AF50. According to the data released by Euro-
NCAP from 2009 to mid of March 2015, about 7%, 17 out of 242, vehicle models are fitted with belt load limiter and pretensioner for rear seats. The trend of fitment ratio of belt load limiter and pretensioner for rear seats is shown as Figure 22. So far, 3 vehicle ratings is released and 1 of them is fitted with rear-seat belt load limiter and pretensioner. The fitment ratio soaring to about 33%. The new FFB of EuroNCAP has pushed the evolution of fitments for rear-seat occupants protection forward.

![Figure 22 Fitment ratio of rear-seat belt load limiter and pretensioner by years, up to March 2015.](image)

It can be estimated that if new protocol assessing more size of occupants will encourage the implement of roof airbag, which provide better protection to rear-seat occupants with belt load limiter and pretensioner.

5.0 CONCLUSIONS

Well correlated sled model is developed to study rear-seat occupants protection due to belt load limiter, pretensioner and roof airbag. The followings could be summarized:

1. The smaller size of occupants, the severer of injury possibility for specific restraint system.
2. For emergency belt, the lower level of load limiter, the better of protection to rear-seat AF05. The joint injury possibility of AF05 ranges from 32.5% to 50.2%.
3. Adding belt pretensioner, the joint ASI 3+ possibility of AF05 will be reduced by 10% to 20% compared with corresponding level of load limiter of emergency seatbelt. However, the ASI 3+ possibility of bigger rear-seat occupants may be increased by 6.4% for AF50 and by 13.9 for AF95 respectively in risk of secondary contact.
4. Roof airbag with low level of load limiter and pretensioner can protect all size of rear-seat occupants from secondary contact and provide better protection in frontal impact. The ASI 3+ possibility of rear-seat occupants can be reduced to 14.5% for AF05, 12.5% for AF50 and 10.1% for AF95, respectively.
5. Consumer protocols persuming better occupant protection has benefitted to rear-seat protection in frontal impact.

Protection to rear-seat occupants in frontal impact has come to vision of safety engineers. More efforts will be made to improve the performance. It will and must be stimulated by new protocol or standards requiring assessing more sizes of rear-seat occupants in frontal impact. The roof airbag or similar equipment will come true.

ACKNOWLEDGEMENTS

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REFERENCES

GENDER DIFFERENCES IN OCCUPANT POSTURE AND MUSCLE ACTIVITY WITH MOTORIZED SEAT BELTS

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Paper Number 15-0384

ABSTRACT

The aim of this study was to assess gender differences in the posture and muscular activity of occupants in response to pretension from motorized seatbelts. Male and female vehicle occupants were tested in both front seat positions during normal driving and autonomous braking. This data is useful for the development of human body models (HBM), and increases the understanding of the effects of motorized belts.

Kinematics and electromyography (EMG) were analyzed for 18 volunteers (9 male, 9 female) subjected to autonomous braking (11 m/s^2 deceleration) during real driving on rural roads. Two restraint configurations were tested: a standard belt and a motorized belt, activated 240 ms before the initiation of braking. Statistical comparison of volunteers' posture and normalized EMG amplitudes was performed to understand differences incurred by the motorized belts, as well as to compare response across gender and role (occupant position within the vehicle). Data was analyzed both prior to and at vehicle deceleration, which occurred 240 ms after motorized belt onset.

Motorized belts significantly affected all postural metrics, and significantly elevated the activity of all muscles compared to typical riding. Though increases in muscle activity were small at deceleration onset compared with typical riding for male occupants and female passengers, female drivers demonstrated significantly larger increases in muscular activity: between 5 and 13% of the maximum voluntary contraction (MVC). At deceleration onset, standard belts showed little change in posture or muscle activation, with the median changes being well within the ranges exhibited during typical riding for all groups (i.e. not distinguishable from typical riding). Typical riding postures of males and females were similar, as were muscular activation levels—generally less than 5% of the MVC. However, drivers exhibited significantly higher muscular activity in the arm and shoulder muscles than passengers.

Limitations include the repeated nature of the testing, as prior work has shown that habituation across trials alters occupant response compared to that of unaware occupants. However, randomization of the trial order helped mitigate potential habituation effects. Another limitation is the sample size of 18 volunteers.

An important finding of this study is that the increase in occupant muscular activation seen with motorized belts was gender-specific: at deceleration, the change in activation of most muscles was significantly different across gender and belt type, with female drivers exhibiting larger increases in muscular activation than male drivers or passengers of either gender, particularly in the arm muscles. These activations appeared to be startle responses, and may have implications for interactions with the steering wheel and motion during a braking or crash event. This warrants further studies and stresses the importance of quantifying male and female subjects separately in future studies of pre-crash systems.

Keywords: Gender; driver, passenger; reversible, motorized seat belt; braking; EMG
INTRODUCTION

Vehicle safety has improved significantly since the 1950s. Early interventions were in-crash systems; for example, improved vehicle structures and occupant restraints. The three-point seat belt was first introduced in 1957, and reported injury reduction was 40–90% (Bohlin 1967) and more than 35% (Norin et al. 1984). Later, Cummings et al. (2003) estimated a 61% lower risk of death for belted front seat occupants compared to unbelted occupants, based on accident data from 1986–1998. More recently, accident avoidance and pre-crash systems that avoid or mitigate the severity of accidents have been implemented on a large scale. Autonomous braking systems have been introduced (Coelingh et al. 2007; Distner et al. 2009; Schittenhelm 2009). Autonomous braking is beneficial by reducing the vehicle’s kinetic energy. However, for impacts with equivalent speed, pre-crash braking increases chest deflection and belt forces (Antona et al. 2010), indicating that injury prevention can be improved with integrated safety systems that reposition the occupant before the impact. One example is autonomous braking in combination with reversible seat belt pretension, which removes belt slack and secures the occupant prior to an impact (Schöneburg et al. 2011). Motorized seat belts provide reversible pretension before the crash and have the potential to reposition the occupant to an optimal pre-crash position.

Reversible seat belt pretension has been studied in combination with driver emergency braking (Tobata et al. 2003), lateral maneuvers (Mages et al. 2011), pre-impact braking (Woitsch and Sinz 2014), and stationary conditions (Good et al. 2008a; Good et al. 2008b; Develet et al. 2013). In these studies, volunteers or anthropomorphic test devices (ATDs) were used. ATDs have severe limitations as they are too stiff to represent relaxed occupants in low loading conditions (Beeman et al. 2012). Increasingly, computational human body models (HBMs) are used in vehicle safety assessment. Currently available HBMs have mainly been developed and validated in the crash loading regime. Active HBMs, which include occupant muscle response and are validated for braking and similar scenarios (Östh et al. 2015), have the potential to become strong tools for the development and assessment of integrated safety systems. However, the active HBM of Östh et al. (2015) represents the 50th percentile male and there are no active HBMs representing female occupants. Therefore, to study gender differences in occupant response with integrated safety systems, volunteers are the best option to date. With increased knowledge from volunteer experiments, future HBMs can be developed to represent gender differences and study integrated safety systems for the full sequence from pre-crash to crash.

The aim of this study was to analyze how pretension with motorized seatbelts changed occupant posture and muscle activity, and to assess differences in response based on gender and occupant position in the vehicle. This was done with statistical analyses of volunteer data from drivers and front seat passengers with and without motorized seat belt pretension prior to autonomous braking interventions. The results have the potential to enhance traffic safety for all occupants, males and females, by providing an increased understanding of how motorized belts affect occupants and by providing data for enhancement of simulation tools, such as HBMs representing both genders.

METHODS

This study analyzed data from volunteer experiments, approved by the Ethical Review Board at the University of Gothenburg, Sweden, where 20 volunteers were exposed to 29 braking interventions (11 m/s2) as drivers (Östh et al. 2013) and passengers (Ólafsdóttir et al. 2013). Volunteers with incomplete data sets were excluded, and therefore 9 females and 9 males were included in this study. For each volunteer, 18 braking interventions were analyzed (12 driver, 6 passenger). The tests were conducted in a passenger car equipped with a motorized belt. Interventions were performed in a randomized order with two seat belt retractor configurations: a standard configuration that locked at 4 m/s2 vehicle deceleration or when the belt pay out acceleration was 15 m/s2 (subsequently denoted “standard”), and a reversible configuration where the electrical motor provided 170 N of belt pretension force at an approximate maximum retraction speed of 300 mm/s (subsequently denoted “motorized”). Each braking intervention was triggered without prior notification to the volunteer. Data were analyzed for two different time periods, before triggering the intervention (termed “typical riding”) and at the onset of vehicle deceleration (termed “initial braking”). Vehicle deceleration occurred on average 350 ms after triggering; for trials with the motorized belt, pretension occurred on average after 110 ms.
Prior to testing, volunteers’ anthropometry was measured (Table 1). Sitting height was the distance between the superior aspect of the head and the seated surface in the mid-sagittal plane, with volunteers sitting on a stool (Schneider et al. 1983). Surface electromyography (EMG) electrodes were applied bilaterally to the volunteers (Figure 1). Volunteers were positioned in a Maximum Voluntary Contraction (MVC) rig, designed to provide a posture that resembled the driving position, and three repetitions of MVC were performed for each tested muscle (Östh et al. 2013). Within the test vehicle, volunteers could partially adjust the driver seat and steering wheel to find a comfortable driving position and were told to keep their hands symmetrically on the steering wheel. Allowed adjustments were translation of the seat, change of the inclination angle of the seat back, and steering wheel position and angle. The passenger seat was fixed (in the mid fore/aft position with a seat back angle of 22°), and volunteers were instructed to keep their feet symmetrical to the midline of the footwell and rest their hands on their lap.

Surface EMG was recorded with a sampling rate of 2048 Hz using a Compumedics Grael (Compumedics, Abbotsford, Australia) for eight muscle groups: sternocleidomastoid, cervical paravertebrals, rectus abdominis, lumbar paravertebrals, biceps brachii, triceps brachii, anterior deltoid, and posterior deltoid (Figure 1). EMG data were normalized using MVC. In typical riding, EMG data were averaged over a one-second interval, from 1.5 to 0.5 s before trigger. In initial braking, EMG data were averaged over a 20 ms interval starting at deceleration onset. The change in activation in initial braking compared to typical riding was investigated per trial as the absolute increase or decrease in activation. Then, the median change across trials of a given condition was used to represent each volunteer’s response.

Kinematic data was acquired at 50 Hz through film analysis (TEMA Automotive, Image Systems, Linköping, Sweden). Posture was measured with video tracking of markers on the volunteer’s head and chest. In the present study, the head center of gravity (CG) position was calculated from film markers close to the ear and eye (Östh et al. 2013). Kinematic posture data were collected in a vehicle-fixed coordinate system, with positive X forward and positive Z upward (Figure 2). Head rotation was the angle between the horizontal plane and the Frankfort plane, in the sagittal plane, with positive rotation representing extension. Relative head-to-sternum distance was the difference between the calculated head CG and the chest marker, effectively measuring the posture of the neck: an individual with a larger head-to-sternum X in one position compared to another would have a more retracted posture in the former. Head-to-head restraint distance was the difference between the head CG and the mid-point on the anterior surface of the head restraint, in the mid-sagittal plane. For typical riding, kinematic data was analyzed and defined as the average value over the first 100 ms after trigger. The median across all trials of a given condition was used to represent each volunteer’s response. For initial braking, the postural metrics were collected at deceleration onset. For each trial, the volunteer’s change in metric was taken as the difference between the response at deceleration onset and at typical riding, with the median change across all trials of a given condition representing each volunteer’s response.

Figure 1. EMG electrode placement (anterior locations on left, posterior locations on right). SCM: sternocleidomastoid; CPVM: cervical paravertebrals; ADELT: anterior deltoid; PDELT: posterior deltoid; BIC: biceps brachii; TRIC: triceps brachii; LPVM: lumbar paravertebrals; RA: rectus abdominis; REF: reference electrode. Adapted from Östh et al. (2013) by permission of The Stapp Association.
Statistical analysis included a normality and homoscedasticity assessment and a general test for group differences: either a parametric repeated-measures ANCOVA, with sitting height as a covariate, or a nonparametric Friedman test (when data did not meet normality or homoscedasticity assumptions). A 5% significance level was used. For typical riding, two-way repeated measures designs were used, with factors being role (driver or passenger) and gender. For initial braking, three-way repeated measures designs were used, with belt type (motorized or standard) as an additional repeated measure. To compensate for the seat belt loading asymmetry, outboard muscles of the drivers and passengers were compared, and likewise for inboard muscles. Data processing was performed in MATLAB (Version 8.0.0, MathWorks, Natick, MA), while statistical analyses were implemented in SAS (Version 9.3, SAS Institute Inc., Cary, NC).

RESULTS

The volunteer sitting height was normally distributed (919 ± 39 mm). A post-hoc t-test indicated a significant difference (p < 0.05) in male and female sitting height, with females (894 ± 33 mm) being shorter than males (945 ± 27 mm). Table 1 lists the average anthropometric measurements for females and males.

Table 1. Volunteer anthropometric data. SD: Standard deviation.

<table>
<thead>
<tr>
<th>Volunteers</th>
<th>Age (years)</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>Sitting Height (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male Mean (SD)</td>
<td>34.1 (12.7)</td>
<td>178.1 (4.8)</td>
<td>77.2 (5.8)</td>
<td>945 (27)</td>
</tr>
<tr>
<td>Female Mean (SD)</td>
<td>28.8 (5.9)</td>
<td>166.6 (5.0)</td>
<td>59.4 (5.2)</td>
<td>894 (33)</td>
</tr>
</tbody>
</table>

Motorized belts significantly affected all postural metrics (Figure 3). The occupant position at initial braking with motorized belts is illustrated in Figure 4 for one male and one female volunteer. Notably, the median sternal X location was shifted posteriorly by more than 10 mm (depending on gender and role) with motorized belts, and the median head-to-sternum X distance was reduced by 12 to 13 mm for females (depending on role) and by 6 to 7 mm for males. The lower values for males may be due to the higher inertia (mass) of male volunteers compared to female volunteers. The range (difference between the 75th and 25th percentiles) of change in postural metrics seen with standard belts during initial braking was well within the range exhibited during typical riding, for all comparable metrics, indicating that there were no other factors except the belt influencing the occupant response. The typical riding position did not seem to depend on gender, as the only significant effect of gender was found for the head-to-sternum X, which also showed significant covariance with sitting height. Role had a significant effect on the head-to-sternum X, indicating differences in neck curvature. Passengers had more than 20 mm larger median head-to-sternum X distances than drivers.
Figure 3. Median change in postural metrics with motorized belts for initial braking compared to typical riding (left) and median postural metrics during typical riding (right). The interquartile ranges are indicated with boxes and outliers by circles. All head metrics are defined at the head center of gravity.

FD: female drivers; FP: female passengers; MD: male drivers; MP: male passengers.

*=significant main effect of gender, +=significant main effect of role, #=significant main effect of belt type, ^=significant covariance with sitting height.
Gender differences were not found for the muscle activity during typical riding (Table 2). Muscle activity was not normally distributed, and was analyzed nonparametrically. Drivers exhibited significantly higher activity in the arm and shoulder muscles than passengers. Though nonsignificant, this trend was evident for most neck and trunk muscles. Notably, while muscles generally displayed interquartile ranges of less than 5% MVC, the lumbar paravertebral muscles had muscle activity up to 20% MVC. With standard belts, the median changes in muscle activity during initial braking were well within the ranges during typical riding for all groups (typically <1% MVC, LPVM <5% MVC), in line with the postural metrics.

Motorized belts significantly increased muscle activity for all muscles (Figures 5 and 6, Table 3). Gender generally displayed significant effects on the arm, shoulder, and trunk muscles, with females showing larger changes than males. The increase in muscle activity was generally small (less than 5% MVC), except for female drivers. In the group of female drivers, for the measured muscles activity increased by 5-13% MVC, and inboard muscles typically displayed slightly larger increases in activity than outboard muscles. With motorized belts, a trend of drivers displaying larger median changes in muscle activity than passengers was seen for most muscles. The lumbar paravertebral muscle had higher increases in activity with motorized belts compared to the other muscles: 9-11% MVC for female drivers, 7-11% for female passengers, 6% for male drivers, and 2-5% for male passengers (Figure 6). Role significantly affected the outboard and inboard cervical paravertebral muscles, with drivers showing larger changes than passengers (Figure 6). Role also significantly affected the outboard rectus abdominis and inboard anterior deltoid.
Figure 5. Median change in arm and shoulder muscle activity with motorized belts for initial braking compared to typical riding (left) and median muscle activity during typical riding (right). The interquartile ranges are indicated with boxes (white for outboard muscles, grey for inboard muscles) and outliers by circles.

FD: female drivers; FP: female passengers; MD: male drivers; MP: male passengers.

*=significant main effect of gender, +=significant main effect of role, #=significant main effect of belt type.
Figure 6. Median change in neck and trunk muscle activity with motorized belts for initial braking compared to typical riding (left) and median muscle activity during typical riding (right). The interquartile ranges are indicated with boxes (white for outboard muscles, grey for inboard muscles) and outliers by circles.

FD: female drivers; FP: female passengers; MD: male drivers; MP: male passengers.

* = significant main effect of gender, += significant main effect of role, # = significant main effect of belt type.
### Table 2.
Median (25th, 75th percentile) Muscle Activity (%MVC) in Typical Riding.
* significant main effect of gender, + significant main effect of role.

<table>
<thead>
<tr>
<th>Group</th>
<th>SCM</th>
<th>CPVM</th>
<th>RA</th>
<th>LPVM</th>
<th>BIC</th>
<th>TRIC</th>
<th>ADELT</th>
<th>PDELT</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Outboard muscles</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Female Drivers</td>
<td>1.6</td>
<td>3.4</td>
<td>1.4</td>
<td>3.8</td>
<td>1.0</td>
<td>1.2</td>
<td>1.7</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>(1.3, 2.3)</td>
<td>(3.0, 4.5)</td>
<td>(1.2, 3.3)</td>
<td>(2.5, 4.2)</td>
<td>(0.7, 1.4)</td>
<td>(1.0, 1.9)</td>
<td>(1.4, 4.3)</td>
<td>(1.0, 1.9)</td>
</tr>
<tr>
<td>Female Passengers</td>
<td>1.1</td>
<td>3.1</td>
<td>1.0</td>
<td>4.7</td>
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<td>0.4</td>
<td>0.2</td>
<td>0.7</td>
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<tr>
<td></td>
<td>(0.9, 2.1)</td>
<td>(2.7, 3.6)</td>
<td>(0.8, 1.7)</td>
<td>(3.6, 6.4)</td>
<td>(0.2, 0.5)</td>
<td>(0.3, 0.5)</td>
<td>(0.1, 0.2)</td>
<td>(0.5, 1.0)</td>
</tr>
<tr>
<td>Male Drivers</td>
<td>1.1</td>
<td>4.8</td>
<td>1.4</td>
<td>4.4</td>
<td>0.7</td>
<td>2.7</td>
<td>4.0</td>
<td>1.7</td>
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### Table 3.
Median (25th, 75th percentile) Change of Muscle Activity (%MVC) with Motorized Belts.
* significant main effect of gender, + significant main effect of role, # significant main effect of belt type.

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<th>LPVM</th>
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</table>

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LIMITATIONS

There are limitations inherent in the experimental procedure, such as the use of a specific test vehicle or the level of autonomous deceleration applied. With respect to the current investigation, all braking events analyzed were autonomous events with volunteers who were unaware of the impending deceleration. As prior work has shown that habituation across trials alters occupant response compared to that of unaware occupants (Blouin et al. 2003a, Siegmund et al. 2003b), the repeated nature of the testing introduces some limitations. However, randomization of the trial order helped mitigate potential habituation effects. As the passenger seat was fixed, the typical passenger riding posture does not take into account potential gender differences in seat adjustment. However, it is plausible that passengers adjust the seats to a lesser extent than drivers during real driving. On the other hand, the driver data capture differences in seat adjustment and are more representative of real life driving postures. Another limitation is the sample size of 18 volunteers.

The change in occupant metrics due to the motorized seat belt was evaluated approximately 250 ms after the start of pretension. At this time, the deceleration had started to build up and was a few percent of the maximum value. It was assumed that the deceleration during initial braking would not influence the occupant response. To ensure that the influence of factors other than the belt pretension was negligible, braking interventions with a standard belt were analyzed at initial braking and compared to the typical riding metrics. The small changes (often less than 1%MVC) in muscular activation from typical riding values seen with standard belts are consistent with the studies by Ejima et al. (2007, 2008, and 2009), which did not find major muscle activity between 0 and 100 ms after deceleration initiation for the muscles tested. They are also in line with trends seen in frontal perturbations, in which normalized integrated EMG values from the sternocleidomastoid and cervical paravertebral muscles were less than 5% and 15%MVC, respectively, for the first 25 ms after deceleration initiation (Blouin et al. 2003b). The reported median changes were within the 25th-75th percentiles of values for the typical riding (i.e. not distinguishable from typical riding) for comparable metrics. Hence, the change in muscle activity compared to typical riding can be considered an effect of motorized belt pretension, and not of deceleration onset.

DISCUSSION

During typical riding, posture and muscular activity was similar across gender. Differences were seen between drivers and passengers. Drivers displayed significantly smaller (by over 20 mm) head-to-sternum X values than passengers, meaning that drivers adopted a more protracted head posture than passengers. The typical riding postures found here are in line with other studies. For instance, Carlsson and Davidsson (2011) reported an average head-to-sternum X distance of approximately 82 mm in a similar study, which is slightly below our median values of 86 - 118 mm depending on group. Likewise, a significant effect of role was seen for muscles in the arm, which is expected due to driver interactions with the steering wheel. However, though significant differences in activation across role were present, median activation levels during typical riding were low: all median values were below 6% MVC and many were below 2% MVC. These values are in line with the maximum pre-impact activation levels of 1-5% MVC reported in the rear impact tests of Szabo and Welcher (1996) for the muscles investigated (sternocleidomastoid, suboccipital cervical extensors, superior trapezius, and paralumbar muscles). They are also in line with reported median muscle activation levels prior to sled perturbation of between 0.6 and 3.7% MVC (Ólafsdóttir et al. 2014), for several cervical muscles (sternocleidomastoid, trapezius, levator scapulae, splenius capitis, semispinalis capitis, semispinalis cervicis, and multifidus).

Motorized belts significantly altered all postural metrics. Though the median changes seen with motorized belts were small for metrics associated with the head, they were slightly larger for metrics associated with the sternum. Indeed, the sternum marker moved typically more than 10 mm posteriorly with the motorized belts, likely because of the direct interaction between the belt and the sternum. These differences were present throughout the braking event: occupants with motorized belts displayed significantly less forward displacement than occupants with standard belts (Ólafsdóttir et al. 2013, Óst et al. 2013). Similarly, Schöneburg et al. (2011) reported that reversible belt tension reduced the median peak forward chest and neck displacements of volunteers with 42% and 34%, respectively, in braking tests with reversible belt tension compared to tests without.
Likewise, motorized belts significantly increased muscle activity. The changes were interesting as they occurred early in time relative to the total braking event, as steady-state deceleration levels were not reached until 0.8 seconds post-trigger (Ólafsdóttir et al. 2013, Östh et al. 2013). Since motorized belt pretension fires before deceleration starts, it is hypothesized that these muscular reactions occur as part of a startle or reflex response to the belt (Ólafsdóttir et al. 2013, Östh et al. 2013). Such startle contractions would result in higher muscle activity at deceleration onset with motorized belts compared to standard belts, which is an important consideration for human body modeling. To the best of the authors’ knowledge, there are no comparable studies investigating muscular effects of motorized belts.

Furthermore, the effect of motorized belts appeared to be gender-specific. The gender difference in the current study was largely driven by female drivers, whose changes in muscle activity ranged between 5 and 13% MVC. As passengers, their changes were less than 5% MVC for all muscles except the lumbar paravertebrals. Also, male changes in muscle activity were typically less than 5% MVC, both as drivers and passengers. However, gender differences in reflex time and activation onset may be contributing to the differences observed in this study. Females have faster stretch reflex times than males for neck flexor (sternocleidomastoid) and extensor (semispinalis capitis, splenius capitis) muscles (Foust et al. 1973). Furthermore, Siegmund et al. (2003a) found significantly different muscle onset times between males and females in a series of frontal sled tests, with female activation occurring 5 and 3 ms before male activation for the sternocleidomastoid and cervical paraspinal muscles, respectively. Taken together, these effects could contribute to the gender differences in activation observed in the current study.

Though the presence of startle was not rigorously investigated in this study, the results presented here indicate that the initial muscular activity provoked by motorized belt pretension is different between males and females. These differences are consistent with differences found later in the braking event, where average activation levels during steady-state braking were higher for females than males for drivers (Östh et al. 2013) and passengers (Ólafsdóttir et al. 2013). Hence, we recommend further investigation of female drivers, exploring the potential startle effect that motorized belt pretension seems to induce for this group of occupants.

CONCLUSIONS

Motorized belts significantly changed the occupant posture, especially with respect to the chest, and significantly increased muscle activity. Gender did not seem to influence the typical riding postural metrics or muscle activation levels. In contrast, drivers and passengers had significantly different metrics for posture and muscle activity in typical riding. The effect of motorized belt pretension was gender-specific. When belt pretension was applied, though changes in postural metrics were similar for males and females, significant differences in activation were observed across gender. This gender difference at initial braking was driven by high changes in activation for female drivers. Therefore, further studies with a focus on female drivers are needed to explore the startle effect that motorized belt pretension seems to induce for this group of occupants.

ACKNOWLEDGEMENTS

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REFERENCES


