SIMULATION ASSESSMENT OF INJURY TRENDS FOR 50TH PERCENTILE MALES USING POTENTIAL SEATING CONFIGURATIONS OF FUTURE AUTOMATED DRIVING SYSTEM (ADS) EQUIPPED VEHICLES

Vikas Hasija
Rohit Kelkar
Bowhead (Systems & Technology Group)
USA

Erik G. Takhounts
National Highway Traffic Safety Administration
USA

Paper Number 19-000345

ABSTRACT

Occupants in vehicles equipped with automated driving systems (ADS) may sit in various seating positions (e.g. forward facing, rear facing, oblique facing) and at different seatback recline angles. Since forward facing impacts have been studied in detail, the objective of this study was to: A) analyze rear impacts using finite element (FE) human and seat models; B) update the seat model based on lessons learned from part A and then analyze the injury metrics trend for the 50th percentile male occupant: i) at various seatback recline angles, and ii) in different carriage style seating configurations; and C) investigate a potential countermeasure for reducing injury metric values. For analyzing rear impacts, the Total Human Model for Safety (THUMS) FE model along with a FE driver seat model of a Toyota Yaris was used. Simulations were carried out at both low (Delta-V=15 mph) and high (Delta-V=35 mph) speeds to understand the effect of seat hinge stiffness on THUMS kinematics at both speeds. Other design changes such as integrated seat belts and active head restraints were also evaluated. Then injury metrics were analyzed for the 50th percentile male occupant at various seatback recline angles and in different carriage style seating configurations. For this part of the study, Global Human Body Models Consortium (GHBMC) 50th percentile simplified (M50-OS) male FE model was used along with a Honda Accord FE driver seat model. Head Injury Criterion (HIC) and Brain Injury Criterion (BrIC) were used as injury metrics for the head/brain, while max chest deflection was used as the chest injury metric. A potential countermeasure for reducing BrIC was investigated for the seating configuration with the highest BrIC value. From the rear impact study, it was found that having a rigid seat hinge, an integrated seatbelt, and an active head restraint help in reducing the injury metrics. Higher BrIC values were observed at higher seatback recline angles for both frontal and rear impacts. Chest deflection was also higher at higher seatback recline angle for frontal impact but showed an inverse trend for rear impact. For occupants experiencing frontal/oblique-frontal impacts, the BrIC and chest deflection values ranged from 0.75 to 0.81 and from 33 to 45 mm respectively whereas for occupants experiencing rear/oblique-rear impacts, the BrIC and chest deflection values ranged from 0.23 to 0.93 and from 17 to 24 mm respectively. HIC_{15} values were below 300 for the various recline angles and seating configurations investigated except one instance where the head contacted the knee. The potential countermeasure (redesigned head restraint) investigated was effective in reducing BrIC by a third for the case with the highest BrIC value (0.93).

KEYWORDS: Automated driving systems (ADS), human models, head/brain injury, chest injury, carriage style seating configurations

INTRODUCTION

The development of vehicles equipped with automated driving systems (ADS) or self-driving cars is one of the most exciting and important innovations in transportation history. ADS equipped vehicles give occupants the ability to travel from one place to another with minimal or no human intervention depending on the automation level. SAE [1] specifies six automation levels which vary from 0 (no automation) to 5 (full automation). ADS focus on levels 3 through 5, in which human driver can give control to the ADS and is not expected to perform any driving related task for a period of time or it may include no human driver at all [2]. By combining advanced technology with minimal or no human intervention, ADS have the potential to reduce collision incidence. However, there will be

vehicles on the public roadways that will not be equipped with ADS. These vehicles can still crash into ADS equipped vehicles, which makes occupant protection very important.

The occupant compartment of ADS equipped vehicles has been dubbed as the "next living space." These vehicles may have non-traditional seating configurations such as front seats that can rotate to face the rear seat occupants, like a carriage style seating configuration. In addition to having forward-facing and rear-facing occupants, ADS equipped vehicles can have occupants sitting oblique to the length of the vehicle because of the flexibility to orient the front seats by various angles. Also, the occupants in these vehicles do not have to sit up right, they can sit at different recline angles of the seatback. A study [3] was conducted to explore and identify possible seating positions and desired activities in future highly automated cars. They observed that for long drives, with several occupants in the car, there was a desire to rotate the seat to a living room position (like carriage style seating) and for short drives, there was a preference to maintain forward-facing position but with seat reclined to a more relaxed position. The flexibility to rotate and recline the seats to many different positions makes occupant protection for ADS equipped vehicles particularly challenging. Current test procedures and Anthropomorphic Test Devices (ATD) might not be sufficient to evaluate injury risk for all these scenarios. Human FE models are an important tool that can allow assessment of occupant protection for new car interiors with flexible seat arrangements.

In ADS equipped vehicles, occupants may experience different types of impact depending on how they are seated. For example, if occupants are seated in a carriage style seating configuration, an impact to the front of the vehicle will cause occupants in the rear seats to experience frontal impact and occupants in front seats to experience rear impact. Currently, most of the high-speed vehicle impact tests are conducted for frontal impacts (e.g. New Car Assessment Program (NCAP) [4]), where all occupants are seated forward-facing. Low speed rear impact tests are presently conducted, however, for occupant protection in ADS equipped vehicles, high speed rear impact tests need to be conducted as well to evaluate the seats, seatbelt, and to understand the occupant kinematics.

It is the purpose of this paper to use FE human and seat models to:

- evaluate occupant kinematics in both low and high speed rear impacts to identify potential injury modes;
- analyze injury metrics trend at different seatback recline angles;
- analyze occupant kinematics and injury metrics trend in carriage style seating configuration using two occupants in different orientations of the front seat;
- evaluate a possible countermeasure for reducing BrIC.

METHODS

Injury Metrics

The following injury metrics are used in this study:

a) $$\text{BrIC} = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2} \qquad (1)$$

where, ω_x, ω_y, and ω_z are the max angular velocities computed at any time about x-, y-, and z-axes respectively. ω_{xC} *(66.25 rad/s)*, ω_{yC} *(56.4 rad/s)*, and ω_{zC} *(42.87 rad/s)* are the corresponding critical angular velocities [5]. The time limit for calculation of BrIC was the entire duration of the simulation (150 ms).

b) $$\text{HIC}_{15} = \left\{\left[\frac{1}{t2-t1}\int_{t1}^{t2} a(t)dt\right]^{2.5}(t2-t1)\right\}_{max} \qquad (2)$$

where, *t2* and *t1* are any two arbitrary times during the acceleration pulse [6].

c) Chest deflection: It was computed as the absolute maximum of the X displacement of the sternum.

Measurements at the neck have proven to be an issue in both the M50-OS and THUMS model because of the modeling techniques used. Therefore, neck injury criterion (N_{ij}) is not reported in this study.

Crash Pulse

The crash pulse used in this study (Figure 1) was chosen from an oblique test (NHTSA test # 9476), which consists of an Oblique Moving Deformable Barrier (OMDB) traveling at a target speed of 90.1 kph into a stationary mid-size four door sedan (2015 Chevrolet Malibu) positioned at 15 degrees relative to the moving barrier and impacted 35% to the left front of the vehicle. Since the crash pulses for oblique impacts are usually more severe in terms of injury

risk compared to those for frontal impacts at the same Delta-V, the crash pulse was chosen from an oblique test. Figure 1 shows the resultant of the longitudinal and lateral acceleration components only (rotational acceleration components were not considered). This resultant pulse was used for all the simulations.

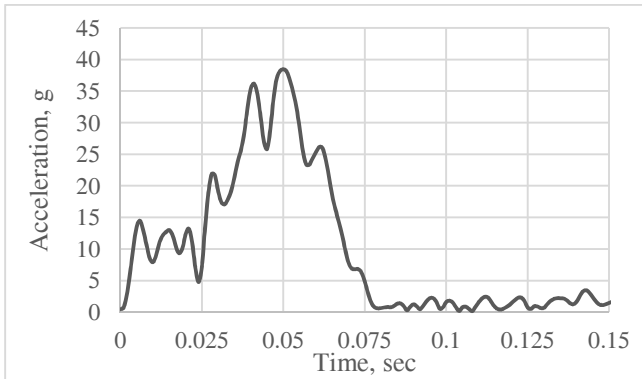


Figure 1. Crash pulse

Model Set-up

All simulations in this study were carried out in a sled type environment. Two different validated 50th percentile FE human models namely THUMS [7] and Simplified GHBMC (M50-OS) [8] along with two different generic FE deformable driver seat models of a 2010 Toyota Yaris [9] and 2014 Honda Accord [10] were used. A validated generic seatbelt system with retractor, pretensioner and load limiter was utilized. The pretensioner and load limiter were set to 1 kN and 3 kN, respectively. The floor was made rigid and no other vehicle structure was used. A sample simulation set-up is shown in Figure 2. All simulations were performed using the general-purpose finite element program, LS-DYNA Version 971 [11].

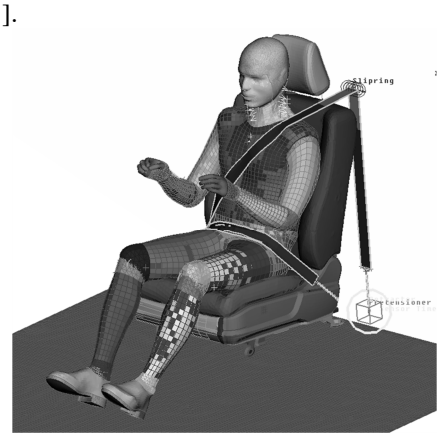


Figure 2. Sample simulation set-up

Analysis

Part 1: Preliminary investigation

This study was carried out to a) evaluate occupant kinematics and injury metrics in both low and high speed rear impacts, and b) explore ways to reduce these injury metrics values. A Toyota Yaris FE driver seat model with THUMS was used for this preliminary analysis. Since no sub system testing or validation was performed on this seat model in rear impacts, a few modifications were made. The bottom seat frame was stiffened and a seat hinge (revolute joint between the seatback and the bottom seat frame, missing in the original model) was created (Figure 3).

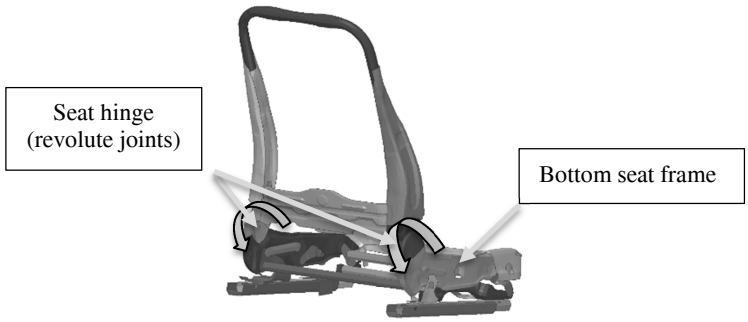
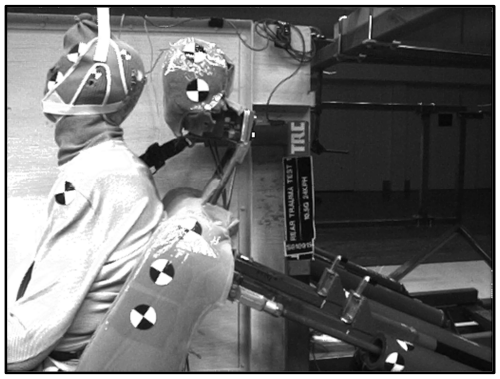
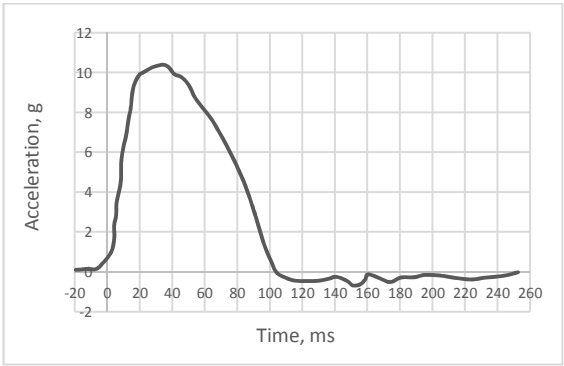


Figure 3. Toyota Yaris FE seat model (seat frame)

A baseline simulation was set up in accordance with a cadaver experiment [12], which consists of a cadaver placed in an experimental seat with non-integrated seatbelt (Figure 4a) and subjected to a rear impact using a 10g acceleration pulse (Delta-V of 15 mph, Figure 4b).



(a)



(b)

Figure 4. a) Rear impact cadaver experiment; b) Experimental 10g pulse

THUMS model was positioned in the seat using gravity simulation following which belt creation and fitting was performed in ANSA [13]. Between the experiment and simulation, the seat back recline angle was the same (25^0) but there was slight difference in the seat pan angle (20^0 in experiment compared to 10^0 in simulation). The seat hinge stiffness in the simulations was set close to that of the experimental seat to ensure equivalent seatback rotation during impact. Both the 10g experimental acceleration pulse (Figure 4b, Delta-V=15 mph) and the crash pulse (Figure 1, Delta-V ~35 mph) were applied to the model to study low and high speed rear impacts. Potential injury modes were identified and corresponding values of HIC_{15}, BrIC and chest deflection were recorded. In addition to the simulations performed with a compliant seat hinge (seatback rotates), similar low and high speed rear impact simulations were performed with a rigid seat hinge (no/minimal seatback rotation) to evaluate its effect on the injury metrics (HIC_{15}, BrIC and chest deflection). Also, a head restraint study was conducted wherein the distance between the head and the head restraint was varied (1 mm, 45 mm) to evaluate its effect on the injury metrics. This head restraint study was conducted with a rigid seat hinge and using the crash pulse in Figure 1(Delta-V ~35 mph).

Part 2: Reclined positions and carriage style seating configurations

Compared to the preliminary investigation (Part1) that was performed using a Yaris FE driver seat model and THUMS human model, this part of the study was conducted using a Honda Accord FE driver seat model and simplified GHBMC FE human model (M50-OS). The Honda Accord FE driver seat model was not available to us during the preliminary investigation. Upon availability, the Honda Accord FE driver seat model was compared to the Yaris FE driver seat model. It was observed that the Honda Accord FE seat was modeled in more details especially the seatback compared to the Yaris FE seat, and was thus used for this part of the study. THUMS was replaced with M50-OS because of computational efficiency. M50-OS is computationally more efficient than THUMS (runs in 2.5 hours for a 150 ms event on 40 processors compared to 24 hours for THUMS). As this part of

the study involved time intensive tasks (evaluating multiple human models in the same set up under different impacts, running multiple pre-simulations for adjusting the extremities and positioning the human model in the seat in different recline positions), the switch was made from THUMS to M50-OS.

Based on the lessons learned from the preliminary investigation (Part1), the Honda Accord FE driver seat model was updated i.e. the seat hinge was made rigid, head restraint was moved slightly forward to reduce the distance between the head restraint and the head, and integrated seatbelt was created. This updated seat model along with the M50-OS human model was used to evaluate occupant kinematics and injury metrics in a) different seatback recline positions, and b) two different carriage style seating configurations.

Reclined positions

Two different seatback recline angles, 20° (driving position) and 45° (relaxed position), were evaluated in both frontal and rear impacts. The crash pulse as shown in Figure 1 was used for all the simulations. Gravity simulations were carried out to position the human model in the seat. For seatback recline angle of 20^0, gravity was applied in Z but for seatback recline angle of 45^0, gravity was applied at 45^0 to properly settle the human model in the seat. The final settled positions are shown in Figure 5. The upper extremities of M50-OS were not modified from their default driving position in these simulations. After gravity simulations, belt creation and fitting was performed in ANSA [13].

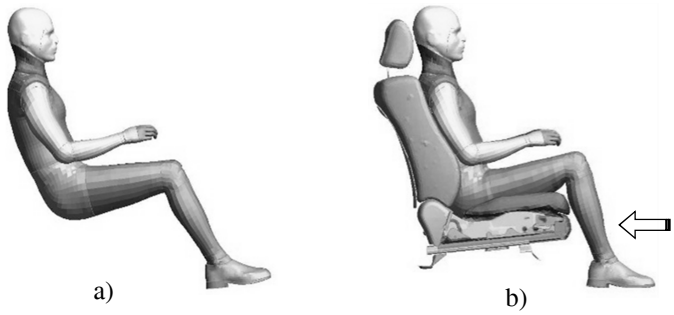
(a) (b)

Figure 5. Final settled positions for seatback recline angle of a) 20^0, and b) 45^0

A total of four simulations were run (front and rear impact for each recline angle). The injury metrics i.e. HIC_{15}, BrIC and chest deflections were recorded for each simulation to investigate their trend as function of seatback recline angle.

Carriage style seating configurations

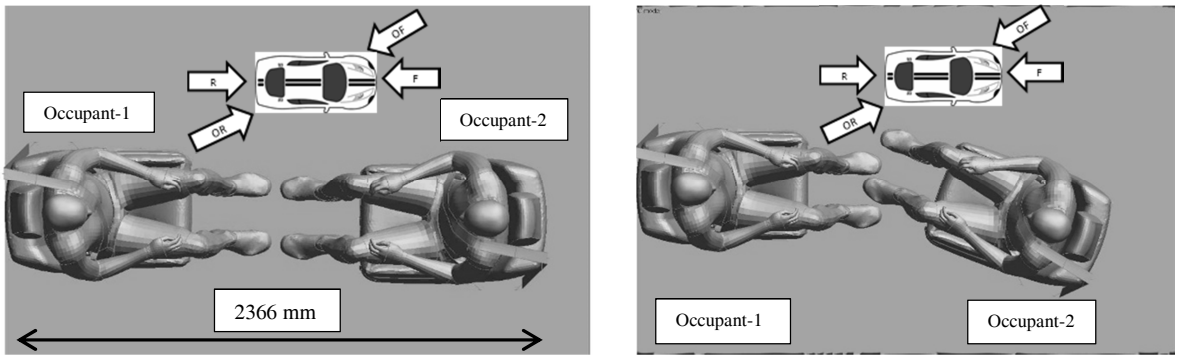
For this study, positioning of the human model in the seat was performed in two steps (Figure 6): a) the upper and lower extremities of the human model were adjusted from driving position to relaxed position, and b) this relaxed model was then settled in the seat using gravity simulation. Along with the gravity simulation, the lower extremities were further adjusted to position them as close to the seat as possible to account for the limited legroom in carriage style seating when all seats are occupied.



a) b)

Figure 6. a) Relaxed model, and b) Settled model with lower extremities positioned close to the seat

Carriage style seating was evaluated using two M50-OS human models. Two different configurations of the carriage style seating were investigated (Figure 7). In configuration 1, the occupants were positioned at 0° (forward-facing, occupant-1) and 180° (rear-facing, occupant-2) relative to the horizontal axis, whereas in configuration 2, the occupants were positioned at 0° (forward-facing, occupant-1) and 200° (rear-facing-oblique, occupant-2) relative to the horizontal axis. In both configurations, the seatback recline angle was 20°.



a) b)

Figure 7. Carriage style seating: a) configuration 1, and b) configuration 2

(*Impacts: **F**: Frontal, **OF**: Oblique-frontal, **R**: Rear, **OR**: Oblique-rear)

The two seats were positioned to keep the initial intersection between the feet of the occupants to a minimum. Such seat positioning resulted in the maximum distance of 2366 mm between the seatbacks (Figure 7a). Keeping no/minimal initial feet intersection between the occupants is reasonable since ADS equipped vehicles with carriage style seating are expected to have more legroom compared to the current mid-size sedans and intersection may not be required. For example, the wheelbase (which gives an idea about the size of the occupant compartment) for Mercedes-Benz F015 equipped with ADS [14] with carriage style seating is around 3610 mm compared to 2829 mm for 2018 Honda Accord [15]. This is almost 28% longer. The vehicle structure and its design might pose additional constraints on seating and will be considered in future studies. Four different impacts were simulated for each configuration (Figure 7) for a total of 8 simulations. Impacts simulated were a) frontal; b) oblique-frontal 20°; c) rear, and d) oblique-rear 20°. For all these simulations, the same crash pulse was used (Figure 1).

Part 3: Countermeasure Evaluation

The combination of seating orientation and impact direction that gave the highest BrIC value was identified from Part 2. The cause of high BrIC value was examined and a potential countermeasure was investigated. The high BrIC value was due to a high z-component of angular velocity. To counter this, the head restraint was redesigned. Morphing was performed using ANSA [13] to redesign the head restraint from a nearly flat to a curved shape as shown in Figure 8. This combination of seating orientation and impact direction was simulated again with the countermeasure in place, and the BrIC value was recorded.

Figure 8. Redesigned head restraint

RESULTS

Part 1: Preliminary investigation

The seatback rotation in the rear impact cadaver experiment that was performed using a 10g pulse (Delta-V of 15 mph) was between 30 - 35°. The corresponding simulation showed similar seatback rotation of ~ 30° as well (Figure 9).

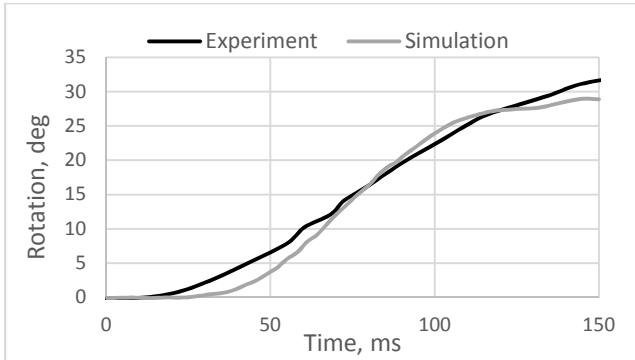


Figure 9. Seatback rotation comparison

The injury metrics observed in the 15 mph and 35 mph Delta-V simulations with a compliant and a rigid seat hinge are shown in Table 1. It can be observed that for both Delta-V's, injury metric values decreased considerably when seat hinge was made rigid. Chest deflection and BrIC were reduced by almost half whereas HIC_{15} was reduced by 15% for lower Delta-V (15 mph). At higher Delta-V (35 mph), the effect of a rigid seat hinge was more pronounced. Each injury metric was reduced at least in half.

Table 1. Injury metrics with compliant and rigid seat hinge

	Delta-V=15 mph			Delta-V=35 mph		
Seat hinge type	BrIC	HIC_{15}	Chest deflection (mm)	BrIC	HIC_{15}	Chest deflection (mm)
Compliant	0.58	92	28	1.85	3466	85
Rigid	0.28	78	15	0.93	490	36

Using a rigid seat hinge, the effect of changing the distance between the head restraint and the head was analyzed. The results are shown in Table 2. BrIC did not change much, however, chest deflection and HIC_{15} decreased by 14% and over 25% respectively.

Table 2. Injury metrics for various distances between head and head restraint

Distance between head-head restraint (mm)	BrIC	HIC_{15}	Chest deflection (mm)
45	0.93	490	36
1	0.90	357	31

Part 2

Reclined positions

The injury metrics for the two seatback recline angles of 20^0 and 45^0 in frontal and rear impacts are shown in Table 3. Chest deflection increased with increase in the seatback recline angle for frontal impacts but showed the opposite trend for rear impacts. However, the BrIC value increased with increase in the seatback recline angle for both rear and frontal impacts. HIC_{15} values were below 250 for all the cases except for the frontal impact in driving position (20^0), which had a very high HIC_{15} value (1106) due to the head contacting the knee.

Table 3. Injury metrics for various recline angles

	Frontal Impact			Rear Impact		
Recline angle	BrIC	HIC_{15}	Chest deflection (mm)	BrIC	HIC_{15}	Chest deflection (mm)
20^0	0.81	1106	38	0.22	145	21
45^0	0.88	236	44	0.42	166	18

Carriage style seating configurations

The injury metrics for each occupant in the two different seating configurations are presented in Figure 10. The arrows in Figure 10 show the direction of impact with respect to the vehicle (Figure 7). It can be observed that for each impact simulated in this study, the occupants in the carriage style seating experience completely different crash events. For example, an impact to the front of the vehicle causes the rear seat occupants (occupant 1) to experience frontal impact and front seat occupants (occupant 2) to experience rear impact. For clarity, the results are discussed in terms of the type of impact experienced by the occupant. The HIC_{15} value was below 300 for all the cases analyzed. In configuration 1, BrIC and chest deflection were the highest for both the occupants when experiencing oblique-frontal impact. In configuration 2, BrIC was highest for occupant 1 when experiencing frontal impact and for occupant 2 when experiencing oblique-rear impact, whereas chest deflection was highest for occupant 1 when experiencing frontal/oblique-frontal impacts and for occupant 2 when experiencing frontal impact.

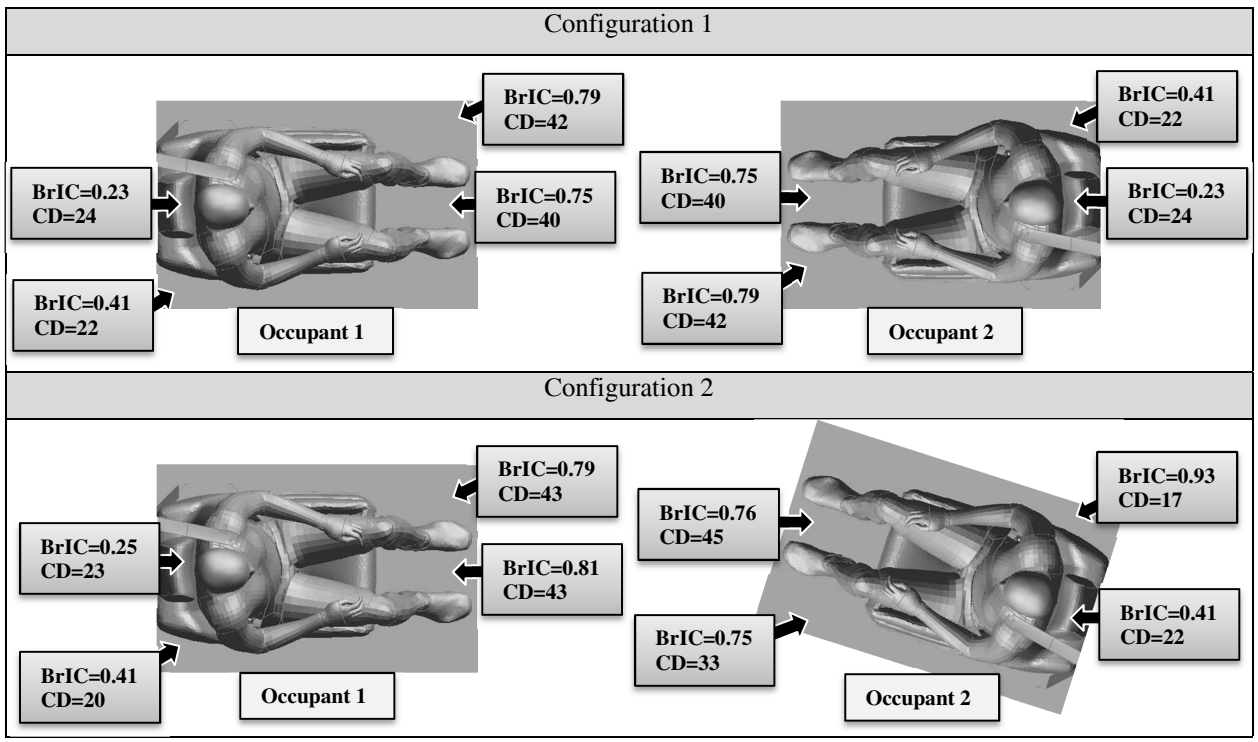
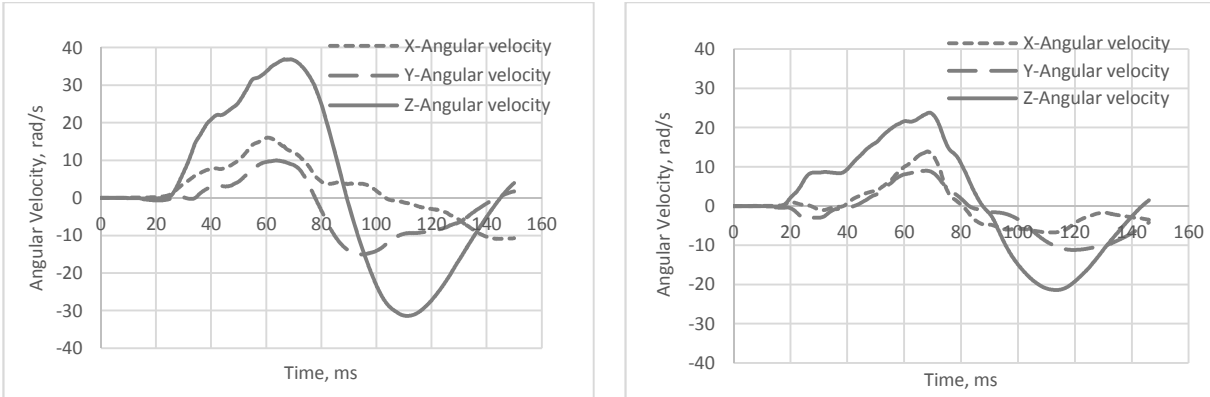


Figure 10. Injury metrics for carriage style seating

(***CD**=Chest deflection (mm), ***Arrows** shown indicate the direction of impact w.r.t the vehicle, Figure 7)

Part 3: Countermeasure Evaluation

Of all the seating orientations analyzed in the carriage style seating configurations, occupant 2 positioned at 200° relative to the horizontal in configuration 2 demonstrated the highest BrIC value when experiencing oblique-rear impact (Figure 10). The BrIC values and angular velocities with and without the countermeasure (redesigned head restraint) are shown in Figure 11. It can be observed that the BrIC value reduced by a third with the redesigned head restraint. This reduction was obtained due to a significant decrease in the z-component of angular velocity.



(a) BrIC=0.93 (b) BrIC =0.63

Figure 11. Angular velocities and BrIC with a) standard head restraint and b) redesigned head restraint

DISCUSSION

Rear impacts were evaluated using a Yaris FE driver seat model and the THUMS human model. Initially, only Yaris seat model was available to us and was thus used for this part of the study. Potential injury modes were identified from the low speed (Delta-V =15 mph) and high speed (Delta-V ~35 mph) rear impact simulations. It was observed that when the seat hinge was compliant, the rotation of the seatback allowed occupant ramping along the seatback. This ramping caused the head to whip back and hit/wrap around the head restraint (Figure A1 of Appendix A), which could cause serious head and neck injuries. Similar kinematics were observed in the low speed rear impact cadaver experiment as well where the compliant seat hinge allowed the cadaver to slide up the seatback causing the head to wrap around the head restraint. In this study, the vehicle structure was not considered but excessive ramping may cause the head to contact the car roof, which can also lead to serious head and neck injuries. In addition, the lower legs contacted the seat pan, which may cause tibia and fibula injuries. To mitigate the potential injury modes, the seat hinge was made rigid. It was observed that with a rigid seat hinge, the seat frame and cushion are forced to absorb more energy and as a result the occupant does not accumulate velocity and the ride down begins immediately (Figure A2 of Appendix A). All the injury metrics evaluated in his study were reduced when a rigid seat hinge was used (Table 1). It was also observed that a non-integrated seatbelt (coming from D-ring) does not provide much protection for the occupants in rear impacts. However, a seatbelt integrated with the seat can help mitigate potential injury risk by reducing the occupant ramping along the seatback by restraining the upper body with the shoulder belt. This is beneficial if the seat hinge is compliant. With a rigid seat hinge, the occupant ramping is not as much when the occupant is in the driving position (seatback recline angle of 20^0 to 25^0), however at higher seatback recline angles (45^0, 60^0), there may be significant ramping and integrated seatbelt can help reduce this ramping considerably. It was also observed that having an active head restraint (that can help minimize the distance between the head and the head restraint) in addition to a rigid seat hinge helps in reducing HIC (Table 2).

In this study, lower chest deflection was observed in rear impacts with a rigid seat hinge compared to a complaint seat hinge (Table 1). Yaris seat model was used for this study. Since this seat model was developed for frontal impacts, the seatback was not fully modeled. The seatback consists of just an outer frame on which the cushion is mounted with no supporting structure in the middle of the seatback (Figure 3). During simulations, it was observed that with a complaint seat hinge the upper body loaded the top of the seatback (due to occupant ramping) which is stiff due to the frame underneath whereas with a rigid seat hinge the upper body loaded the upper-middle to middle area of the seatback with no supporting structure behind it causing the seatback cushion to deform more. This may be the reason for lower chest deflections seen in this study with a rigid seat hinge. For rear impacts, stiffness of the seatback supporting structure is an important parameter that was not investigated in this study. This will be done as part of future work with newer FE seat models to better understand the relationship between seat hinge rigidity and chest deflection.

Simulations with higher seatback recline angle showed higher BrIC values for both frontal and rear impacts. It was observed that for frontal impacts, the x- and y-components of angular velocity were higher at higher seatback recline angles leading to higher values of BrIC, whereas for rear impacts, y- and z-components of angular velocity were

higher at higher seatback recline angles. The trend observed is based on limited number of simulations (two for each impact) and needs to be evaluated at more seatback recline angles.

For occupant 1, positioned at 0° relative to the horizontal in both configurations 1 and 2 (Figure 10), there was a difference in the type of impact that had the highest BrIC value. Oblique-frontal impact had the highest BrIC value for occupant 1 in configuration 1 (0.79) compared to frontal impact for the same occupant in configuration 2 (0.81). This difference was due to the feet interaction between the occupants during impact. In configuration 1, since the occupants were positioned face to face with their feet aligned along the horizontal axis, there was feet interaction between them during impact where as in configuration 2 because of different seating orientations of the occupants, there was no feet interaction. The lack of interaction caused the feet of occupant 1 to contact the seat in front of it, thereby changing the kinematics. This change in kinematics caused the head to contact the knee during impact leading to higher BrIC value. The kinematics can be different for different set-ups.

Occupants, when experiencing frontal/oblique-frontal impacts (Figure 10), demonstrated BrIC values in the range from 0.75 to 0.81, and chest deflections in the range from 33 to 45 mm. The lowest chest deflection (33 mm) was for occupant 2 in configuration 2 because of high degree of obliqueness (40° between the seating direction and impact direction). Occupants, when experiencing rear/oblique-rear impacts (Figure 10), demonstrated BrIC values in the range from 0.23 to 0.93 and chest deflections in the range from 17 to 24 mm. Again, occupant 2 in configuration 2 showed the lowest chest deflection (17 mm) because of high degree of obliqueness (40° between the seating direction and impact direction). Chest deflection in frontal/oblique-frontal impacts is due to belt loading whereas in rear/oblique-rear impacts it is due to inertial loading. High degree of obliqueness between the seating direction and impact direction reduces the belt loading on the sternum in frontal impacts and the inertial loading on the seatback in rear impacts leading to lower chest deflection.

The BrIC and chest deflection values for occupants, when experiencing rear/oblique-rear impacts (Figure 10), are quite low. This is due to the rigid seat hinge used in the simulations. A complaint seat hinge would lead to much higher values for rear impacts (Table 1). The actual values here represent those measured by the simplified GHBMC model and are not directly translated to any ATD, i.e. if the chest deflection is about 45 mm, it doesn't mean that if a model of an ATD was used instead of the GHBMC, this value would've been the same.

It can be observed from Figure 10, that there are certain combinations of impact direction and seating orientation which are better than others in terms of BrIC. For occupant positioned at 0^0 or 180^0 relative to the horizontal and experiencing rear/oblique-rear impacts, the BrIC value is very low whereas for occupant positioned at 200^0 relative to the horizontal, the BrIC value is low when experiencing rear impact but very high when experiencing oblique-rear impact. The high degree of obliqueness between seating direction and impact direction causes substantial increase in the z-component of angular velocity, thus increasing the BrIC value. Similar results were observed in Kitagawa study [16], where higher degree of obliqueness gave higher BrIC values in rear impacts. Therefore, while high degree of obliqueness between the seating direction and impact direction reduces chest deflection, it increases BrIC.

Occupants seated in different orientation of the front seat (occupant 2 in configuration 1 and 2, Figure 10), when experiencing frontal/oblique-frontal impacts, did not show any clear trend for the chest deflections nor did they demonstrate any significant changes in the BrIC values. Thus, no seating orientation (180^0, 200^0) stood out as the safest in terms of injury metrics. However, when experiencing rear/oblique-rear impacts (Figure 10), occupant in seating orientation of 180^0 (occupant 2 in configuration 1) demonstrated substantially lower BrIC values compared to occupant in seating orientation of 200^0 (occupant 2 in configuration 2). Although the lowest chest deflection was observed for the occupant in seating orientation of 200^0, the considerable increase in BrIC value outweighs any advantages in terms of chest deflection for this seating orientation. In addition, the chest deflection values are relatively low for rear/oblique-rear impacts (17 to 24 mm) compared to the threshold of 63 mm [6] and may be irrelevant as a selection parameter when choosing safe seating orientation.

The highest value of BrIC obtained in this study was 0.93 for occupant 2 in configuration 2 experiencing oblique-rear impact (Figure 10). The potential countermeasure, modified head restraint, reduced the z- component of angular velocity by more than a third thereby reducing the BrIC value by almost a third as well to 0.63. Although the BrIC value reduced substantially with the countermeasure, occupants in seating orientations of 0^0 and 180^0 demonstrated lower BrIC value (0.41, Figure 10) when experiencing the same impact.

Limitations

- The results presented are limited to the occupant parameters (size, stature, gender) and restraint parameters (seatbelt) used in this study.
- Simulations were run in a sled type environment without any vehicle structure, which may pose additional constraints.
- Only head and chest injury metrics were considered.
- Crash pulse shape was not varied based on PDOF which may affect the results.
- The resultant crash pulse used in this study only considers the longitudinal and lateral components. The results may be different if rotational acceleration components are applied to the model.
- Results presented are based on limited number of simulations and impact types.
- Finally, any outcome from the models is only as good as the models themselves and experimental validation of the results is required, which may be carried out in the future.

Future work may also involve evaluating kinematics and injury metrics for the 50th percentile male by changing the crash pulse, load limiter, occupant location relative to each other (move them closer); simulating carriage style seating configurations for 50th percentile male at different seatback recline angles, and evaluating 5th percentile female in similar scenarios.

CONCLUSIONS

The following observations were made from this study:

- Injury metrics for occupants in rear impacts sitting at various recline angles can be reduced by using:
  - Rigid seat hinge
  - Head restraint positioned close to the head
  - Integrated seatbelts
- Higher seatback recline angles lead to higher BrIC values for both frontal and rear impacts.
- For carriage style seating, occupant interaction during impact can affect the kinematics and hence the injury metrics.
- High degree of obliqueness between the seating direction and impact direction reduces chest deflection but increases BrIC.
- Seating orientations of 0^0 and 180^0 relative to horizontal may be safer for occupants in rear impacts.
- Redesigned head restraint (possible countermeasure) may be useful in reducing BrIC.

REFERENCES

[1] Taxonomy and Definitions for Terms Related to On-Road Motor Vehicle Automated Driving Systems. SAE Information Report (J3016)

[2] https://www.nhtsa.gov/sites/nhtsa.dot.gov/files/documents/13069a-ads2.0_090617_v9a_tag.pdf

[3] Jorlov, S., Bohman, K., Larsson, A. (2017). Seating Positions and Activities in Highly Automated Cars - A Qualitative Study of Future Automated Driving Scenarios. IRCOBI.

[4] The New Car Assessment Program Suggested Approaches for Future Program Enhancements (2007). DOT HS 810 698.

[5] Takhounts, E.G., Craig, M.J., Moorhouse, K., McFadden, J., Hasija, V. (2013) Development of brain injury criteria (BrIC). Stapp Car Crash Journal 57:243-66.

[6] Eppinger, R., Sun, E. et al. (1999) Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems – II, NHTSA, USA.

[7] Iwamoto, M., Nakahira, Y., Kimpara, H. (2015). Development and Validation of the Total HUman Model for Safety (THUMS) Toward Further Understanding of Occupant Injury Mechanisms in Precrash and During Crash. Traffic Injury Prevention, Volume16.

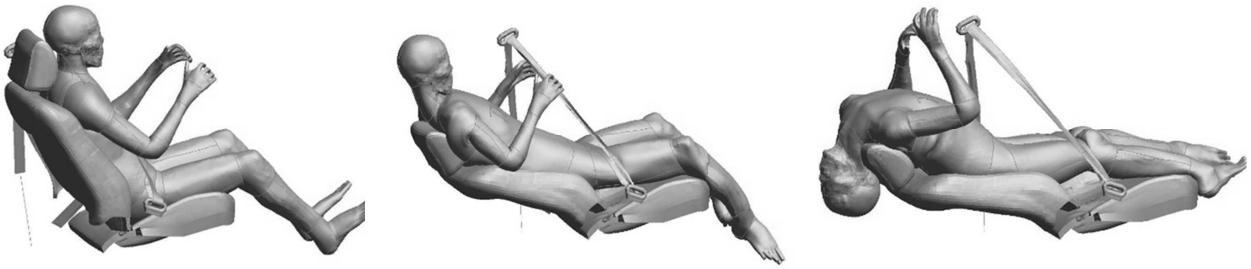
[8] Schwartz, D., Guleyupoglu, B., Koya, B., Stitzel, J.D., Gayzik, F.S. (2015) Development of a computationally efficient full human body finite element model. Traffic Injury Prevention.16 Suppl 1: S49-56.

[9] Reichert, R., Park, C-K., & Morgan, R. M. (2014). Development of integrated vehicle-occupant model for crashworthiness safety analysis (Report No. DOT HS 812 087). Washington, DC: National Highway Traffic Safety Administration.

[10] Singh, H., Ganesan, V., Davies, J., Paramasuwom, M., & Gradischnig. (2018). Vehicle interior and restraints modeling development of full vehicle finite element model including vehicle interior and occupant restraints

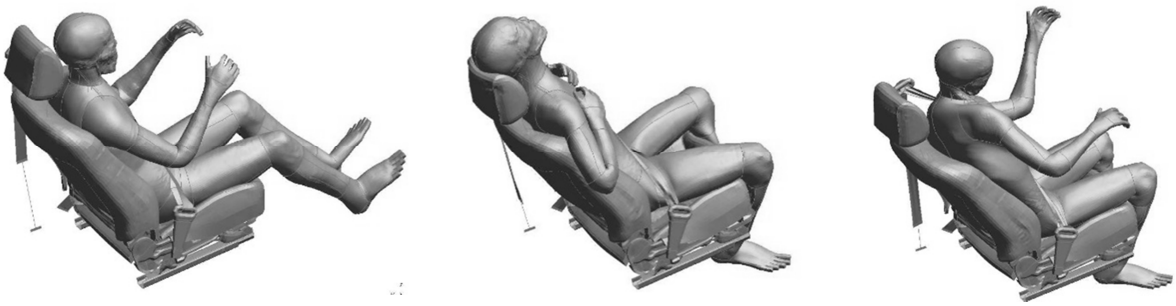
systems for occupant safety analysis using THOR dummies (Report No. DOT HS 812 545). Washington, DC: National Highway Traffic Safety Administration.
[11] LS-DYNA v.971. Livermore Software Technology Corporation.
[12] Kang, Y.S., Bolte, J.H., Moorhouse, K., Donnelly, B. (2012). Biomechanical Responses of PMHS in Moderate-Speed Rear Impacts and Development of Response Targets for Evaluating the Internal and External Biofidelity of ATDs. Stapp Car Crash Journal.
[13] ANSA, version 17.1.2, BETA CAE Systems.
[14] https://media.daimler.com/marsMediaSite/en/instance/ko/Overview-Mercedes-Benz-F-015-Luxury-in-Motion.xhtml?oid=9904624
[15] http://owners.honda.com/vehicles/information/2018/Accord-Sedan/specs#mid^CV2F5JJNW
[16] Kitagawa, Y., Hayashi, S., Yamada, K., Gotoh, M. (2017). Occupant Kinematics in Simulated Autonomous Driving Vehicle Collisions: Influence of Seating Position, Direction and Angle. Stapp Car Crash Journal.

APPENDIX A: OCCUPANT KINEMATICS



t=0 ms t=78ms t=135ms

Figure A1. Occupant kinematics in rear impact with a compliant seat hinge (Delta-V ~35 mph)



t=0 ms t=78ms t=135ms

Figure A2. Occupant kinematics in rear impact with a rigid seat hinge (Delta-V ~35 mph)