A Human Modelling Study on Occupant Kinematics in Highly Reclined Seats during Frontal Crashes

Kyle J. Boyle, Matthew P. Reed, Lauren W. Zaseck, Jingwen Hu

Abstract The objective of this study was to use the finite element (FE) Global Human Body Model Consortium (GHBMC) simplified midsize male human body model to investigate occupant kinematics and potential safety concerns with a large range of recline angles and restraint configurations in frontal crashes. Target postures were obtained by predicting joint centre locations at three recline angles, i.e. 25°, 45°, and 60°, using a seating posture model developed based on volunteer measurements at a wide range of recline angles. A mesh morphing method was used to change the pelvis, lumbar vertebrae, and the surrounding soft tissues of the GHBMC model into the target postures. Three seat models (rigid seat, semi-rigid seat, and an OEM seat), four seatbelt lap belt angles, the presence of an anchor pre-tensioner and dynamic locking tongue (DLT), and the presence of a knee bolster and toe pan were varied in a parametric study under a generic 56 km/h frontal crash pulse. No airbag was used in any of the simulations. Occupant kinematics as well as injury measures were examined for potential safety concerns. Several interesting trends were observed in the simulations. The lap belt angle showed dominant effects on submarining. Although changing the restraints can reduce the submarining risk, there is a clear conflict between submarining and lumbar spine force in highly reclined postures, as cases without submarining were associated with higher lumbar force.

Keywords Automated driving systems, finite element, highly reclined, occupant kinematics, submarining.

I. INTRODUCTION

The current regulatory process for evaluating vehicle safety is focused on occupants that are in a standard seating posture with a seat back angle of around 25° from vertical. Current concepts for automated driving systems (ADS), however, include seats that allow the occupant to ride with a highly reclined seat back for riding comfort or resting. Highly reclined seats may pose challenges for occupant protection. The biomechanical response and injury outcomes of an occupant in a highly reclined seating posture is likely to differ substantially from those of an occupant in a standard automotive posture. Reference [1] found that mortality increased for both partially and fully reclined occupants in crashes. With a higher recline, the occupant may be more likely to submarine, a kinematic phenomenon observed in frontal impact that occurs when the lap belt slips over the occupant's pelvis, loading the soft tissues of the abdomen and adversely affecting torso kinematics [2,3]. Additionally, the orientation of the spine in reclined postures is more aligned to the loading direction of the crash, which may contribute to higher compression forces in the spine [4]. The importance of highly reclined responses is likely to rise with the prevalence of ADS, but few studies have investigated the kinematics and safety concerns of occupants with highly reclined postures in severe automotive crash scenarios.

Finite element human body models are a widely used injury assessment tool that can aid in investigating high recline angle scenarios. Reference [5] observed differences in excursions of the T1 vertebra in different recline angles using the THUMS model. In addition to demonstrating overall kinematics, [3] used human body models to demonstrate reactions of the spine and pelvis. Reference [6] observed lap belt and pelvis interactions at three different recline angles using two different d-ring locations and both the GHBMC simplified and detailed models. The current study aims to investigate these responses using a several selected recline angles and restraint configurations in frontal crashes using a volunteer study based posture positioning method.

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II. METHODS

Baseline FE Human Model

In this study, the midsize male simplified occupant FE model from the Global Human Body Model Consortium (GHBMC M50-OS V1.8.4) was used. The GHBMC M50-OS model was created from the same source geometry as the original GHBMC model, but with a comparatively coarser mesh to provide faster run times. Bones were assumed to be rigid, except for the ribcage, and mechanical joints were defined for the hip, knee, ankle, shoulder, elbow, and wrist for easy positioning and posturing of the human model. The GHBMC M50-OS model has been validated extensively against cadaver tests, including a 23 kg hub impact to the thorax with an initial velocity of 6.7 m/s, a 48 kg bar impact to the abdomen with an initial velocity of 6 m/s, a 23.4 kg plate side impact to the right arm with an initial velocity of 12 m/s, a lateral sled test condition with an initial velocity of 6.7 m/s, and a frontal sled test condition with an 11.1 m/s crash pulse. In a frontal sled condition, the CORrelation and Analysis (CORA) ratings for the body excursions of the GHBMC M50-OS ranged from 0.55 to 0.83 [7]. More details of the model validation can be found in [8].

FE Occupant Posture through Mesh Morphing

The baseline model was morphed into appropriate postures for seat back recline angles of 25°, 45°, and 60° from vertical. The posture target was determined through a posture prediction model that was developed at the University of Michigan Transportation Research Institute (UMTRI) based on volunteer testing [9]. Three-dimensional locations of landmarks on the body of each of the 24 volunteers were measured at seat back angles of 23°, 43°, and 53°. Joint positions, including the hip, L5/S1, T12/L1, C7/T1, atlanto-occipital (AO), tragion, and eye, were estimated from the surface landmarks, and regression models were developed to predict these landmark and joint locations from seat back angle and occupant body dimensions.

To adjust the posture of the original GHBMC M50-OS model to the target posture, the pelvis, vertebral bodies of the lumbar spine, and the upper torso above the lumbar spine were rotated about the associated joint centres. The pelvis was first rotated at the hip to match the predicted pelvis orientation. The lumbar joints were then rotated to match the head and thoracic spine joint locations while creating no overlap between lumbar vertebrae. The soft tissues around the pelvis and lumbar spine were then morphed around the new target locations of the pelvis and lumbar bones using a landmark-based 3D non-linear interpolation technique based on radial basis functions (RBF). The RBF mesh morphing method has been widely used in our previous studies for morphing a baseline FE human model into target geometries with a wide range of sizes and shapes [10,11]. The femurs of the models were rotated so that the thighs were parallel to the seat, and the lower legs were rotated so that the feet were resting on the floor using the joints of the GHBMC simplified model. Flesh that overlapped the seats was morphed to the seat geometry. The arms were lowered so that the hands were near the thighs.

Lower Extremity Constraint Simulations

A few initial simulations were run to examine the effects of lower extremity constraints. Three restraint conditions were considered: *no constraints*, *foot constraints*, and a *touching bolster* condition. The foot constraints condition tied the heel of the occupant to the floor of the vehicle using one-dimensional spring elements, but there was no toe pan and nothing blocking the legs above the ankle of the occupant. The touching bolster condition included a rigid toe pan and rigid knee bolster that effectively constrained the movement of the entire lower extremities. Simulations using each of the constraint conditions were completed with the human model in a 45° recline posture. The vehicle model setup included a shoulder belt retractor with a constant load limiter (3.23 kN) and a pre-tensioner (2.5 kN) fired at 12 ms. Fig. 1 shows the 56 km/h (35 mph) acceleration pulse that was used for all the simulations. Submarining was determined using the simulated animations (2 ms time interval) by checking whether there was a portion of the lap belt moving over the anterior superior iliac spine (ASIS) on either side of the pelvis to the abdomen.

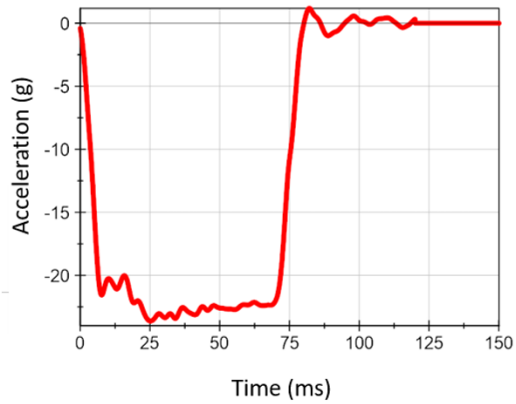

Fig. 1. Acceleration pulse used, based on 56 km/h pulse run at UMTRI sled lab.

Seat and Recline Angle Simulations

In the second parametric study, we varied the seat model and the recline angle (25°, 45°, and 60°) to create a 3x3 matrix of nine simulations. The three seats include a rigid seat, semi-rigid seat, and an original equipment manufacturer (OEM) vehicle seat. The completely rigid seat geometry was developed using geometry that represents typical seat frames. The length and angles of the seat pan and anti-submarining features of six OEM seat models were measured to construct the rigid seat geometry. Three of the seat models were already available to UMTRI and three were downloaded from the NHTSA website [12]. The semi-rigid seat was based on the seat designed to mimic the behavior of real seats by [13]. This semi-rigid seat has a seat pan plate that rotates and is controlled by two sets of springs that results in a two-slope angular stiffness. A second anti-submarining plate is also present and is controlled by two springs for a single slope angular stiffness. The stiffness of the springs were chosen to represent the behaviour of a previously physically tested OEM seat. The semi-rigid seat stiffness is modelled as torsion springs with the front seat moment/rotation relationships reported in [13]. The OEM vehicle seat was taken from the Toyota Yaris model downloaded from the NHTSA website [12]. Fig. 2 shows cross sections of the seats used as well as the seats used as a source for the models. A seat H-Point location was estimated for each of the seats in order to place occupant models according to the posture predication model. A floor surface was placed 270 mm below the H-Point location of the seat. The lap belt anchorage locations are the same as those used for the volunteer study. The outboard lap belt anchorage creates a lap belt angle of 65° from horizontal in the XZ plane while the buckle location creates an angle of 70°. The shoulder belt anchorage for the 25° recline angle simulations was located at the same location as the volunteer study in the most upright position and was rotated about the seat H-point location for the 45° and 60° recline angle simulations. The vehicle model setup also included a shoulder belt retractor with constant load limiter (3.23 kN) and pre-tensioner (2.5 kN). No lower extremity constraints were applied. The same 56 km/h acceleration pulse was used for all simulations. Kinematics, status of submarining, and some injury values were output and examined. Chest deflection was measured from T8 to mid-sternum of the occupant.

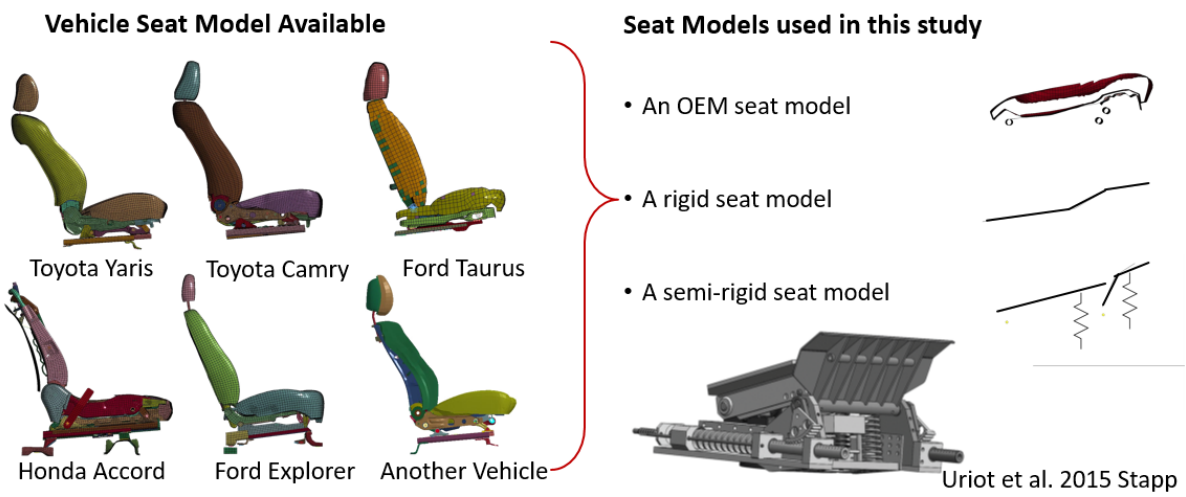

Fig. 2. Seats used to create each of the models (left) and cross sections of the seat models used (right).

Lap Belt Angle and Advanced Restraints Simulations

In the third parametric study, we varied the outboard lap belt angle and the presence of an anchor pre-tensioner (AP) and a dynamic locking tongue (DLT). Each of these simulations used the semi-rigid seat model at a 45° recline angle posture. The shoulder belt anchorage was adjusted slightly to a location 20 mm above the shoulder of the occupant, 175 mm rearward of the occupant shoulder, and 225 mm outboard of the seat centreline. This location definition change was made so that a better shoulder belt fit would likely result for occupants other than the midsize male in future studies. This minor change was made after the second parametric series and was determined to be unlikely to affect the first and second parametric series results so they were not re-run with the change. The location of the buckle was kept constant and the outboard anchor was placed in different fore-aft locations to create four different lap belt angles with respect to forward horizontal (45°, 51°, 58°, and 65°). A total of 16 (4 lap angles x 2 AP levels x 2 DLT levels) simulations were conducted. Fig 3 shows the parameters that were varied in the study. The same 56 km/h acceleration pulse was used for all simulations. Kinematics, visual signs of submarining, and selected injury values were output and examined.

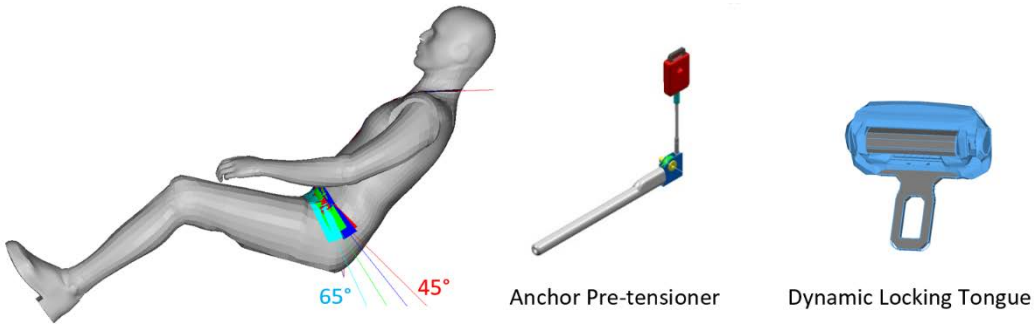

Fig. 3. Belt angles and advanced restraints used in this study.

III. RESULTS

Occupant Posture

The three morphed human models are shown in Fig. 4 along with the comparison to the original GHBMC simplified model. Joint angles from the statistical model and resulting morphed models are shown in Table 1 with standard deviations listed from the volunteer study. After morphing, the mesh quality slightly decreased for each of the models. The smallest Jacobian values for shell and solid elements were 0.3 and 0.26 respectively, compared to 0.4 and 0.3 of the GHBMC M50-OS model. The difference in pelvis angle between the predictive model and the morphed model never exceeded 1°. Differences in joint angles between the model prediction and positioned model did not exceed 9° and averaged 2.5°. Differences in joint positions did not exceed 50 mm, and averaged 15 mm.

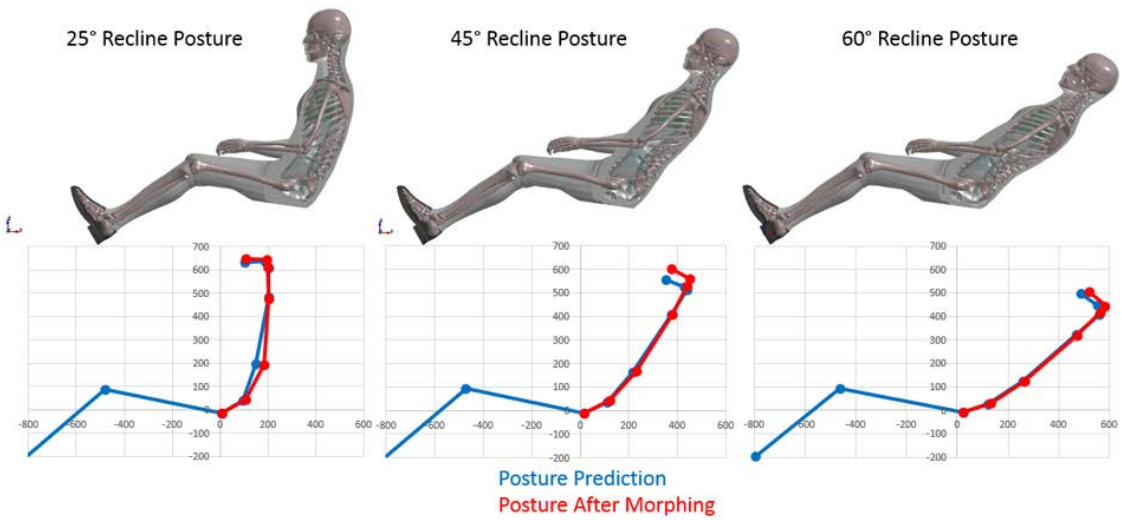

Fig. 4. Posture comparisons between morphed models and seating posture model predictions

TABLE I

JOINT ANGLES OBTAINED FROM POSTURE PREDICATION MODEL AND MORPHED MODELS AFTER POSITIONING (UNIT: DEGREE)

Recline Angle	Body Region	Body Region Definition	Statistical Model Angle	Volunteer Study SD	Positioned Model Angle
25 (23 SD listed)	Pelvis	Hip to L5/S1	57.9	12.7	58.0
	Lumbar	L5/S1 to T12/L1	18.7	9.5	27.1
	Thorax	T12/L1 to C7/L1	10.1	6.0	3.7
	Neck	C7/T1 to AO	1.0	8.0	-0.2
	Head	Eye to Tragion	-2.8	10.1	1.2
	Hip-Eye	Hip to Eye	8.4	5.8	8.4
45 (43 SD listed)	Pelvis	Hip to L5/S1	64.5	15.1	64.0
	Lumbar	L5/S1 to T12/L1	39.1	11.8	40.5
	Thorax	T12/L1 to C7/L1	32.9	5.9	31.7
	Neck	C7/T1 to AO	32.1	10.3	27.8
	Head	Eye to Tragion	24.2	10.1	28.3
	Hip-Eye	Hip to Eye	30.9	4.1	30.3
60 (53 SD listed)	Pelvis	Hip to L5/S1	69.5	15.3	70.2
	Lumbar	L5/S1 to T12/L1	54.4	15.0	55.1
	Thorax	T12/L1 to C7/L1	46.7	6.1	46.7
	Neck	C7/T1 to AO	47.1	8.9	42.8
	Head	Eye to Tragion	37.8	8.9	44.4
	Hip-Eye	Hip to Eye	43.8	3.4	43.9

Note: Standard deviations listed are at 23°, 43°, and 53° instead of 25°, 45°, and 60°.

Lower Extremity Constraints Results

Differences in occupant kinematics as well as excursions at the head, left shoulder, and pelvis from the three lower extremity constraint conditions can be found in Fig. 5 and Fig. 6. During the touching bolster case, pelvis excursion was limited, and the upper body pitched forward more than the other lower extremity restraint cases. As a result, although no submarining occurred, the lumbar spine of the occupant deformed into extension before transitioning back to flexion under the increased loading during the event. Except for the lower legs kicking up, the simulation with no constraints behaved very similarly to the simulation with foot constraints. Submarining occurred in the simulations with no constraints and with foot constraints each at 50 ms, and maximum head and shoulder excursions were within 6 mm of each other.

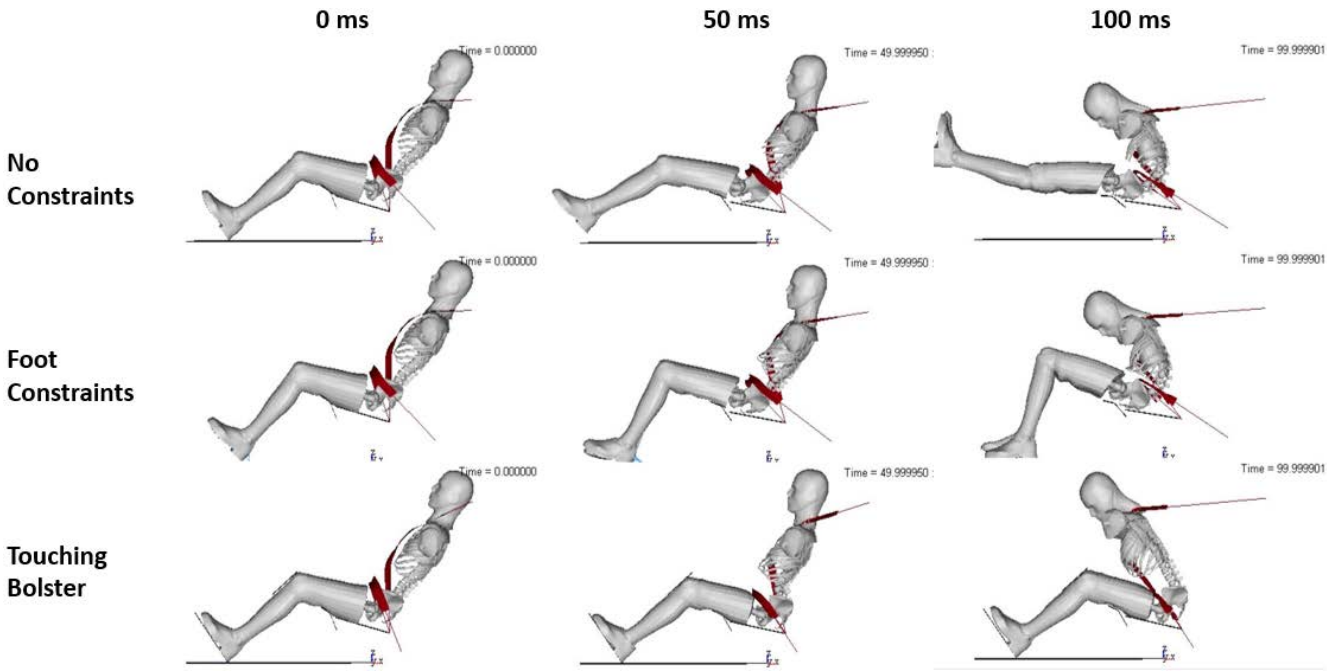

Fig. 5. Kinematics during simulations examining lower extremity constraints.

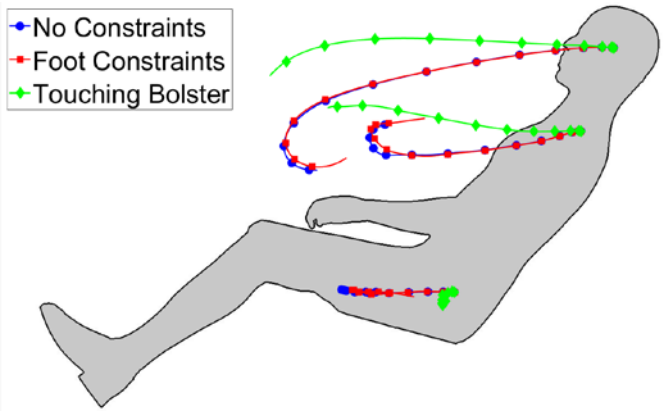

Fig. 6. Excursion traces of head, left shoulder, and pelvis. Markers are shown after every 10 ms.

Seat and Recline Angle Study Results

Differences in occupant kinematics as well as excursions at the head, left shoulder, and pelvis from the nine simulations are shown in Fig. 7 and Fig. 8. Recline angle had large effect on occupant kinematics including whether the occupant submarined. The seat type had a smaller effect on upper body kinematics. The vehicle OEM seat resulted in the most forward and downward excursions of the pelvis, followed by the semi-rigid seat, and then the rigid seat. This motion resulted in submarining at a 45° recline angle for the vehicle seat while submarining was not observed for the semi-rigid and rigid seat at the same recline posture. Fig. 9 shows trends in selected kinematics and injury measures. In all three simulations at a 25° recline angle, contact between the head and lower extremities can be observed. Chest deflection decreased with recline angle for each seat while compression force in the lumbar spine increased. Upper body kinematics and occupant pitch angle changed greatly with recline angle.

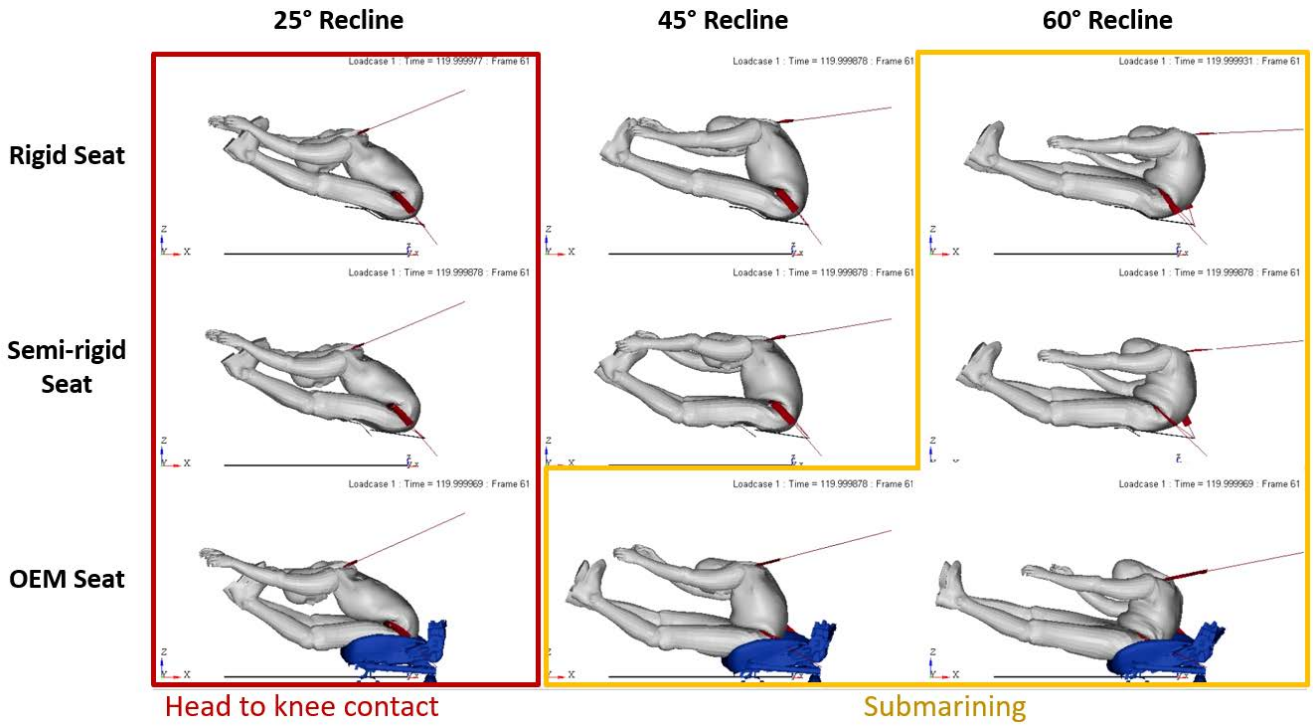

Fig. 7. Occupant position during simulation at 120 ms.

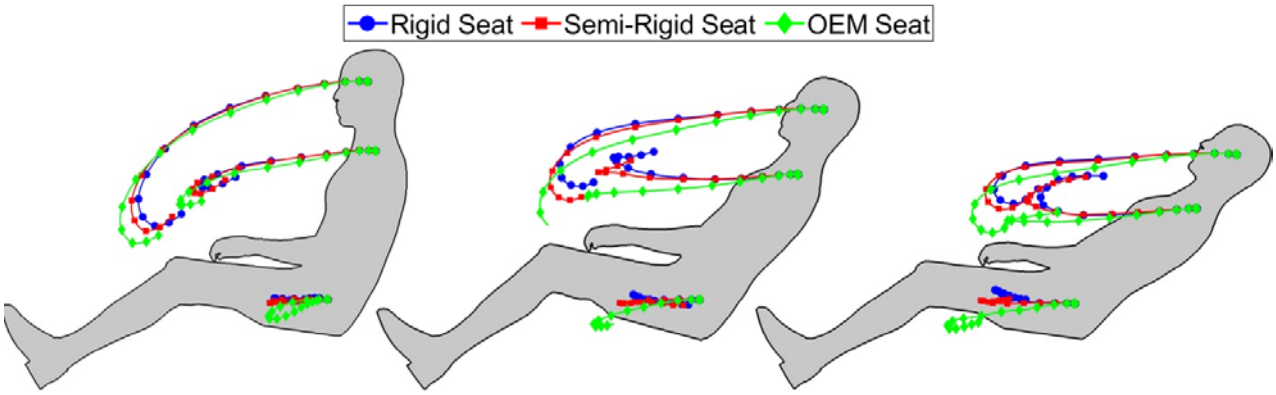

Fig. 8. Excursion traces of head, left shoulder, and pelvis. Markers are shown after every 10 ms.

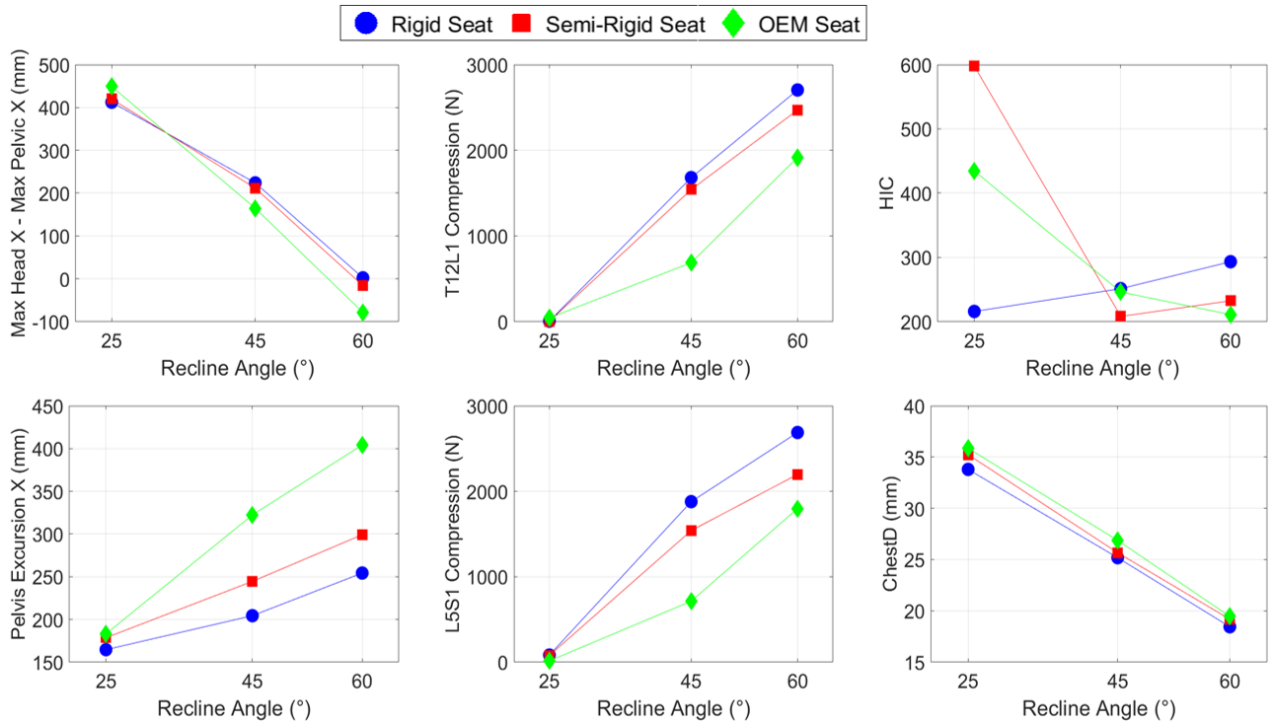

Fig. 9. Trends in maximum values of kinematic measures.

Belt Angle and Advanced Restraints Study Results

Differences in the interactions between the pelvis and lap belt as well as excursions at the head, left shoulder, and pelvis for the 16 simulations in the third parametric study, all conducted at 45˚ recline, are shown in Fig. 10 and Fig. 11. Submarining occurred in all of the 45° and 51° belt angle simulations but in none of the 65° simulations. The 58° simulations were split with two simulations resulting in submarining (no AP/DLT and DLT only) and two simulations resulting in no submarining (AP only and both AP/DLT). Fig. 12 shows selected kinematic and injury trends. Pelvis excursion and rotation decreased with increasing belt angles. Both the presence of an anchor pre-tensioner and a dynamic locking tongue reduced the amount of pelvis excursion and rotation. Higher belt angles result in higher spine compression forces and higher chest deflection. The changes in belt configurations does not show any major trends with head injury criterion (HIC).

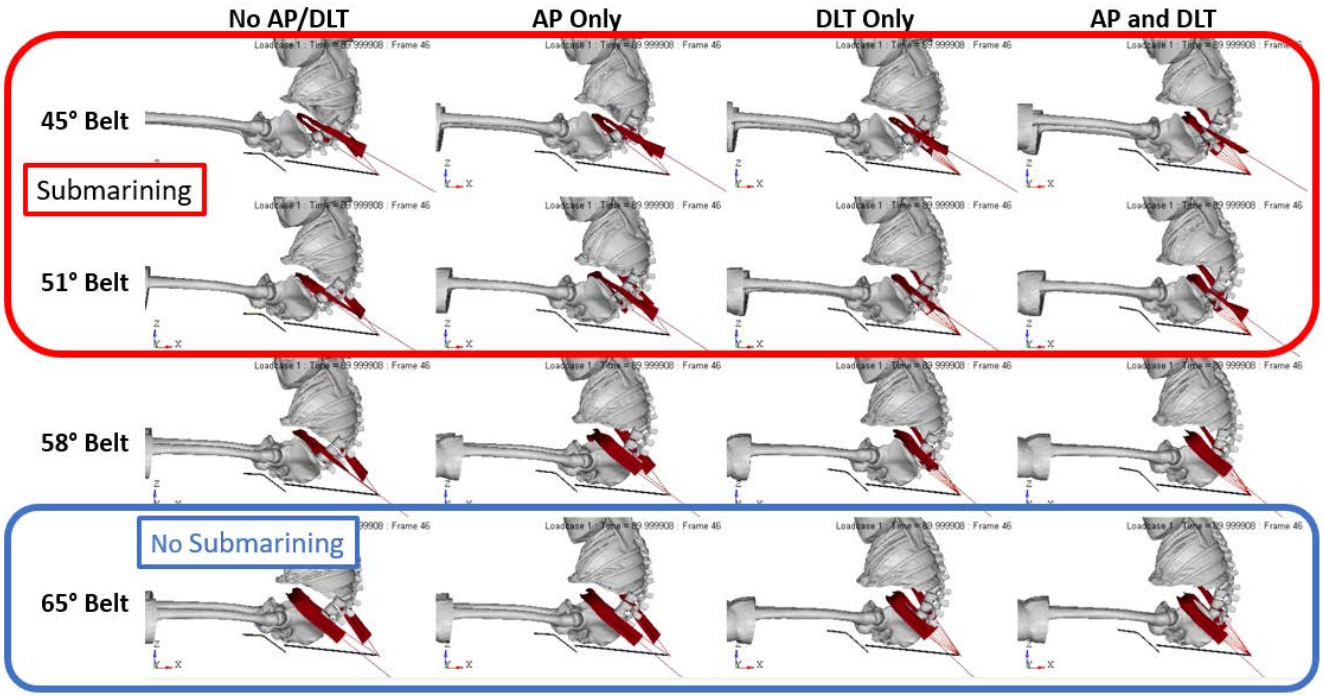

Fig. 10. Occupant position during simulation at 90 ms.

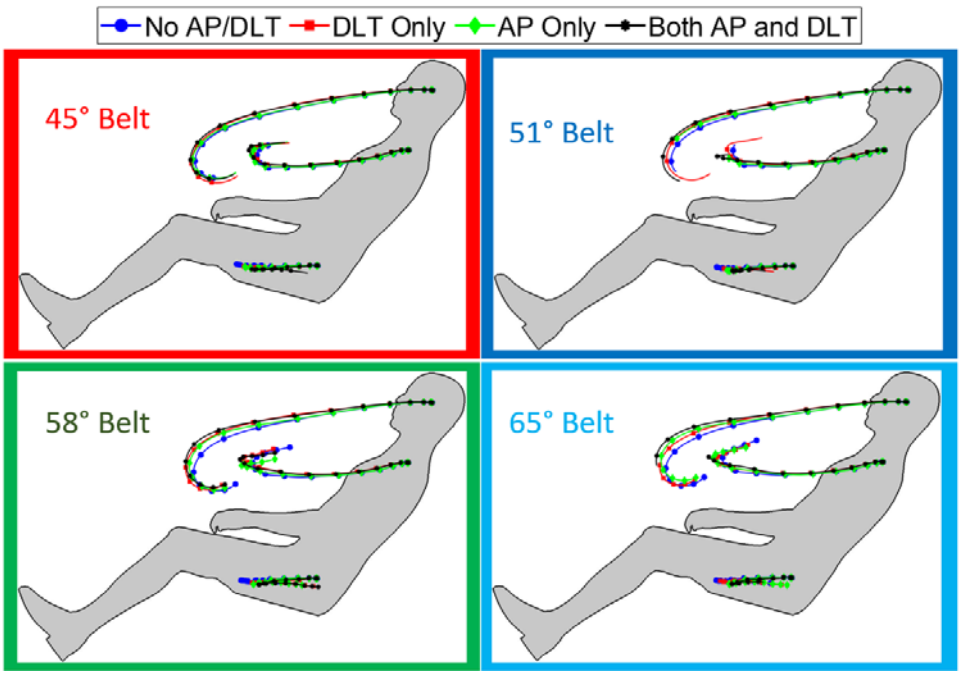

Fig. 11. Excursion traces of head, left shoulder, and pelvis. Markers are shown after every 10 ms.

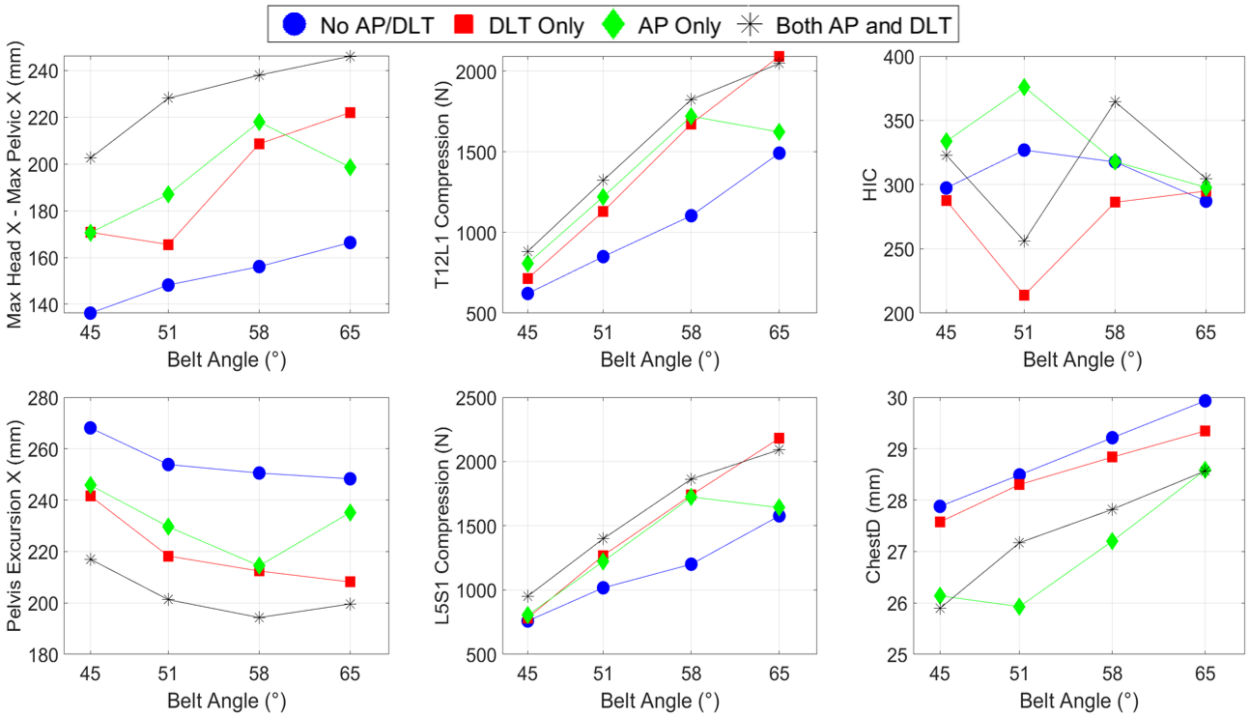

Fig. 12. Trends in maximum values of kinematic measures.

IV. Discussion

Lower Extremity Constraints

These simulations show a conflict between restraining the pelvis and increasing lumbar force in highly reclined postures. Because the initial angle of the spine is more in line with the impact direction, a larger portion of the restraining force on the upper body is transferred through the lumbar spine. During the simulation with the bolster contacting the knees of the occupant, the pelvis could not move forward and the spine deformed from flexion into extension before returning to flexion later in the event. The simulations with only foot constraints and with no constraints show that the presence of an obstacle blocking the feet does not have a large effect on the upper body kinematics or occupant submarining behaviour. However, the foot restraints were set up by constraining the heels of the occupant so that limited force was transferred to the upper legs. A different type of foot constraint, for example a toe pan, or a constraint that also restricts a larger portion of the legs may produce different results.

Seat and Recline Angle

This set of simulations are the first focusing on human model simulations for highly reclined occupants under a wide range of seat and restraint conditions. The second parametric study reported here shows that recline angle has a large effect on occupant kinematics in a frontal crash. Based on measurements of volunteers, the initial angle of the pelvis in these simulations was rotated rearward with the seat back recline angle. As a result, the ASIS was positioned lower and more rearward making it easier for the lap belt to slip over the ASIS, contributing to the observed submarining. The submarining observed at higher recline angles is consistent with [6]. Due to the large pelvis excursion, the torso does not pitch forward over the pelvis even with a load limit on the shoulder belt. These kinematics are similar to those observed in the reclined simulation studies done previously by [5]. Previous studies also showed that, due to highly likelihood of submarining at higher recline postures, a large increase in injury risk can be expected for the internal organs near the abdomen [14]. Our results showed much higher compression forces at the lumbar joints for higher reclined conditions. On the contrary, for the 25° reclined cases, the relatively low belt load limit in the absence of an airbag causes the occupant to pitch forward such that the head contacts the thigh area, causing a spike in the head acceleration.

The seat used in the simulations did not have a large effect on upper body excursions of the occupant. However, the OEM seat did result in a larger pelvis excursion due to plastic deformation of bracket elements attached to the side of seat frame. This deformation of the seat resulted in an increased downward movement of the pelvis and submarining. It is not clear whether this phenomenon is realistic as the seat model was not

validated under these conditions. The Toyota Yaris full vehicle model that the OEM seat was taken from has been validated against frontal vehicle crash test results using acceleration results at various points in the vehicle[15]. However, there is no validation of the seat model against any crash tests. For the seat models in this study, stiffer seats reduced the amount of excursion of the pelvis resulting in higher lumbar spine compression forces, particularly in higher reclined postures as the initial angle of the reclined spine is more in line with the impact direction and a larger portion of the restraining force on the upper body is transferred through the lumbar spine.

Belt Angle and Advanced Restraints

The third parametric study did not show a large difference in upper body excursions as the posture and seat remained the same for all simulations. However, the interaction between the pelvis and lap belt was greatly influenced by the restraint system. A more vertically oriented lap belt is less likely to slip over the top of the ASIS and result in submarining. However, with the pelvis better captured by the lap belt, higher lumbar compression forces resulted. The presence of an anchor PT and DLT each reduced the amount of pelvis excursion, and in turn reduced the submarining risk. The trend of higher lumbar compression loads with lower pelvis excursion was again observed. This again is because the initial angle of the reclined spine is more in line with the impact direction and a larger portion of the restraining force on the upper body is transferred through the lumbar spine. A larger lap belt angle typically resulted in less pelvis excursion, and in turn led to higher lumbar force. The use of an AP was shown to be effective against submarining for the 58° belt angle case, but further study is needed to determine the effectiveness of PT and DLT on submarining in highly reclined postures.

Limitations

Only one high severity pulse was used due to the already high number of simulations being run and the pulse chosen represents a change in velocity that is used for many current crash regulations and assessments. Kinematics and injury risk may be different at lower severity pulses which may be more prevalent in ADS.

PMHS data are not yet available under these types of conditions to validate the models. As seen in this study, recline angle can greatly change the occupant kinematics. The lumbar spine of the human model may have greater importance for accuracy than in typical simulations of occupants in upright postures. The compression of the lumbar is currently modelled by joints without any contact between the vertebrae. Lumbar forces in this study did not exceed 3 kN, which is lower than reported values for ultimate force in previous studies [16]. However, for some of the simulations the compression force was high enough that there was overlapping between the vertebrae with very little flexion or extension, which is unrealistic for an actual occupant. Looking into the GHBMC simplified model more closely, the compression definition of the beam elements for the lumbar is much less stiff than the tension definition. It is likely that either the defined polarity is reversed or the compression definition needs to be stiffer. A more detailed lumbar spine model that is validated in these conditions may be necessary.

A single belt load limit was used for all simulations. Different load limits may be advantageous for different recline scenarios. The 25° degree recline postures resulted in head to leg contact indicating a higher load limit might be a better option. Conversely, the 60° recline postures resulted in a limited amount pitching forward of the torso before and after submarining occurred, indicating a lower load limit might be a better option.

Because actual seating environments in ADS are still unknown, and lower extremity constraints could affect potential submarining, no knee bolster or other lower extremity constraint was used for the second and third parametric series of simulations in this study. A range of relevant lower extremity constraint scenarios could be considered in future studies.

Finally, body shape can have large effects on occupant impact responses in frontal crashes [17-19]. In this study only the midsize male occupant was considered. Obesity has been shown to have a large effect on belt fit and lap belt interaction. Considering that submarining risk increases with highly reclined postures, further study is needed with occupants of different shapes and sizes.

V. CONCLUSIONS

This study investigated occupant kinematics and potential safety concerns for the midsize male with a large range of recline angles and restraint configurations in frontal crashes. The results indicated several interesting injury concerns that are not as common in frontal occupants with normal seating postures and an airbag. Based on these simulations, the differences in posture that result from changing the recline angle have strong effects

on the kinematics and injury outcomes of the occupant including the likelihood of submarining. The configuration of the restraint system also affects whether the occupant submarines as well as other injury related outcomes. Future studies are needed to validate the models for these scenarios and to investigate the effects of these conditions on occupants other than the midsize male.

VI. Acknowledgement

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VII. References

[1] Dissanaike, S., Kaufman, R., Mack, C.D., Mock, C., and Bulger, E. The effect of reclined seats on mortality in motor vehicle collisions. *Journal of Trauma-Injury Infection and Critical Care*, 2008. 64(3): p.614–619

[2] Thorbole, C.K. Dangers of Seatback Recline in a Moving Vehicle: How Seatback Recline Increases the Injury Severity and Shifts Injury Pattern. *Proceedings of the Asme International Mechanical Engineering Congress and Exposition, 2015, Vol 3*, 2016

[3] Ji, P.J., Huang, Y., and Zhou, Q. Mechanisms of using knee bolster to control kinematical motion of occupant in reclined posture for lowering injury risk. *International Journal of Crashworthiness*, 2017. 22(4): p.415–424

[4] Amenson, T.T., Begeman, P.C., Melvin, J.W., and Grimm, M.J. Development of the MADYMO Race Car Driver Model for Frontal Impact Simulation and Thoracolumbar Spine Injury Prediction in Indianapolis-type Racing Car Drivers. *SAE Int. J. Passeng. Cars - Electron. Electr. Syst.*, 2008. 1(1): p.630–651

[5] Kitagawa, Y., Hayashi, S., Yamada, K., and Gotoh, M. Occupant Kinematics in Simulated Autonomous Driving Vehicle Collisions: Influence of Seating Position, Direction and Angle. *Stapp Car Crash J*, 2017. 61: p.101–155

[6] Lin, H., Gepner, B., Wu, T., Forman, J., and Panzer, M. Effect of Seatback Recline on Occupant Model Response in Frontal Crashes in *IRCOBI conference*. 2018: Athens, Greece.

[7] Shaw, G., Parent, D., et al. Impact response of restrained PMHS in frontal sled tests: skeletal deformation patterns under seat belt loading. *Stapp Car Crash J*, 2009. 53: p.1–48

[8] Schwartz, D., Guleyupoglu, B., Koya, B., Stitzel, J.D., and Gayzik, F.S. Development of a Computationally Efficient Full Human Body Finite Element Model. *Traffic Injury Prevention*, 2015. 16: p.S49–S56

[9] Reed, M.P. and Ebert, S.M. Effects of Recline on Passenger Posture and Belt Fit. 2018, University of Michigan Transportation Research Institute.

[10] Hwang, E., Hu, J., et al. Development, Evaluation, and Sensitivity Analysis of Parametric Finite Element Whole-Body Human Models in Side Impacts. *Stapp Car Crash J*, 2016. 60: p.473–508

[11] Zhang, K., Cao, L., et al. An automated method to morph finite element whole-body human models with a wide range of stature and body shape for both men and women. *Journal of biomechanics*, 2017. 60: p.253–260

[12] "Crash Simulation Vehicle Models" Internet https://www.nhtsa.gov/crash-simulation-vehicle-models. 10/19/2018].

[13] Uriot, J., Potier, P., et al. Reference PMHS Sled Tests to Assess Submarining. *Stapp Car Crash J*, 2015. 59: p.203–23

[14] Poplin, G.S., McMurry, T.L., et al. Nature and etiology of hollow-organ abdominal injuries in frontal crashes. *Accid Anal Prev*, 2015. 78: p.51–57

[15] Marzougui, D., Samaha, R.R., Cui, C., Kan, C.-D.S., and Opiela, K.S. Extended Validation of the Finite Element Model for the 2010 Toyota Yaris Passenger Sedan 2012, The National Crash Analysis Center.

[16] Stemper, B.D., Yoganandan, N., et al. Rate-dependent fracture characteristics of lumbar vertebral bodies. *Journal of the Mechanical Behavior of Biomedical Materials*, 2015. 41: p.271–279

[17] Reed, M.P., Ebert-Hamilton, S.M., and Rupp, J.D. Effects of Obesity on Seat Belt Fit. *Traffic Injury Prevention*, 2012. 13(4): p.364–372

[18] Jones, M.L.H., Ebert, S.M., Hu, J., and Reed, M.P. Effects of High Levels of Obesity on Lap and Shoulder Belt Paths, in *IRCOBI conference*. 2017: Antwerp, Belgium.

[19] Hu, J., Zhang, K., et al. Stature and Body Shape Effects on Driver Injury Risks in Frontal Crashes: A Parametric Human Modelling Study, in *IRCOBI conference*. 2017: Antwerp, Belgium.