Identifying Potential Injury Causations and Risk Factors in Moderate Severity Crashes through Parametric Human Modelling

Jingwen Hu, Yang-Shen Lin, Kyle J. Boyle, Matthew P. Reed, Martin Östling

Abstract The objective of this study was to use parametric human models to identify potential injury causations and risk factors in moderate severity crashes. A previously validated occupant compartment representing a midsize SUV was used for all crash simulations. THUMS v4.1 midsize male model was morphed into three male and three female (midsize, short obese, and tall obese) occupants with a nominal and three non-nominal postures based on UMTRI's statistical human geometry model. A crash pulse at 32 km/h and three impact directions were used for a total of 72 crash simulations. Obese occupants showed higher lower extremity injury risks than non-obese occupants. Female occupants showed higher lower extremity injury risks than male occupants. Leaning inboard is associated with higher chest injury risks, especially for obese occupants, while slouching is associated with higher tibia injury risks. Brain injury risks are generally much higher than other body regions. However, there is a low correlation between BrIC and DAMAGE results that requires further investigation. This study demonstrated that occupant characteristics and postures had significant effects on occupant injury risks in moderate severity frontal crashes, while impact angle affects the brain injury risk the most among all body regions.

Keywords Parametric human body model, moderate severity crashes, injury causation, risk factors, crash simulations.

I. INTRODUCTION

The current evaluation of vehicle safety designs for crashworthiness and occupant protection relies primarily on regulatory and consumer information testing programs, which emphasise high severity crash conditions. For instance, US NCAP front barrier crash tests are conducted with an initial vehicle velocity of 56 km/h and result in a delta-V of approximately 61~64 km/h, which is higher than the delta-V of 99% of all frontal crashes [1]. While occupant injury risks are generally lower in low to moderate severity crashes compared to those in high severity crashes, these crashes account for the majority of moderate to serious injuries in motor vehicle crashes due to their high frequency [2]. Field data have shown that half of the occupants with AIS3+ chest injuries were in crashes with a delta-V less than 40 km/h, while most crashes resulting in AIS2+ injuries had a delta-V less than 34 km/h [3-6]. Addressing the relatively low injury risks in low to moderate severity crashes presents substantial challenges, as current evaluation tools and assessment methods were developed primarily for high severity crashes.

While several studies have investigated injury incidence in low to moderate severity crashes, there remains limited understanding of why certain occupants sustain injuries while others do not in these crashes. Low injury tolerance could account for some of the risk in these crashes [4,7], but the risk may also be increased by some combination of non-nominal occupant size, shape, and posture, restraint conditions, and crash conditions. By non-nominal we mean conditions deviating from those used in regulatory and consumer information crash testing, which include only a small number of crash conditions and ATD body sizes and positions.

For the past decade, the Biosciences group at UMTRI has been collecting data and developing methods for setting up an automated crash simulation framework to use parametric human body models, vehicle models, and restraint models to account for the variations of occupant characteristics, occupant posture, belt fitting, vehicle geometry, restraint systems, and crash conditions [8,9]. With the recent advances of mesh morphing technologies, our previous studies have shown that a baseline finite element (FE) human body model (HBM), typically representing midsize male, can be rapidly morphed to any target geometry with a given set of subject

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parameters (e.g. age, sex, stature, weight, posture, etc.) [10-13], while maintaining similar mesh quality to the baseline model. Compared to the traditional HBMs, each of which took months to years to be developed, the parametric HBMs not only save time and cost for developing a large number of HBMs with a wide range of sizes, shapes and postures, but also enable automated simulation setup and results collection because all models share the same mesh structure and injury measure definitions.

Leveraging the parametric HBMs and automated simulation framework, the objective of this study was to use the parametric modelling and simulation framework to identify potential injury causations and risk factors in moderate severity frontal crashes.

II. METHODS

Method Overview

As shown in Fig. 1, we started with morphing a baseline midsize male HBM into six HBMs to represent occupants with a wide range of sizes and shapes. These morphed HBMs were then integrated into a validated driver compartment model through an automated HBM positioning, settling and belt-fitting process. A total of 72 frontal crash simulations under 32 km/h were conducted, with varied occupant postures and crash directions. Finally, injury outcomes were analysed to identify potential injury causations and risk factors.

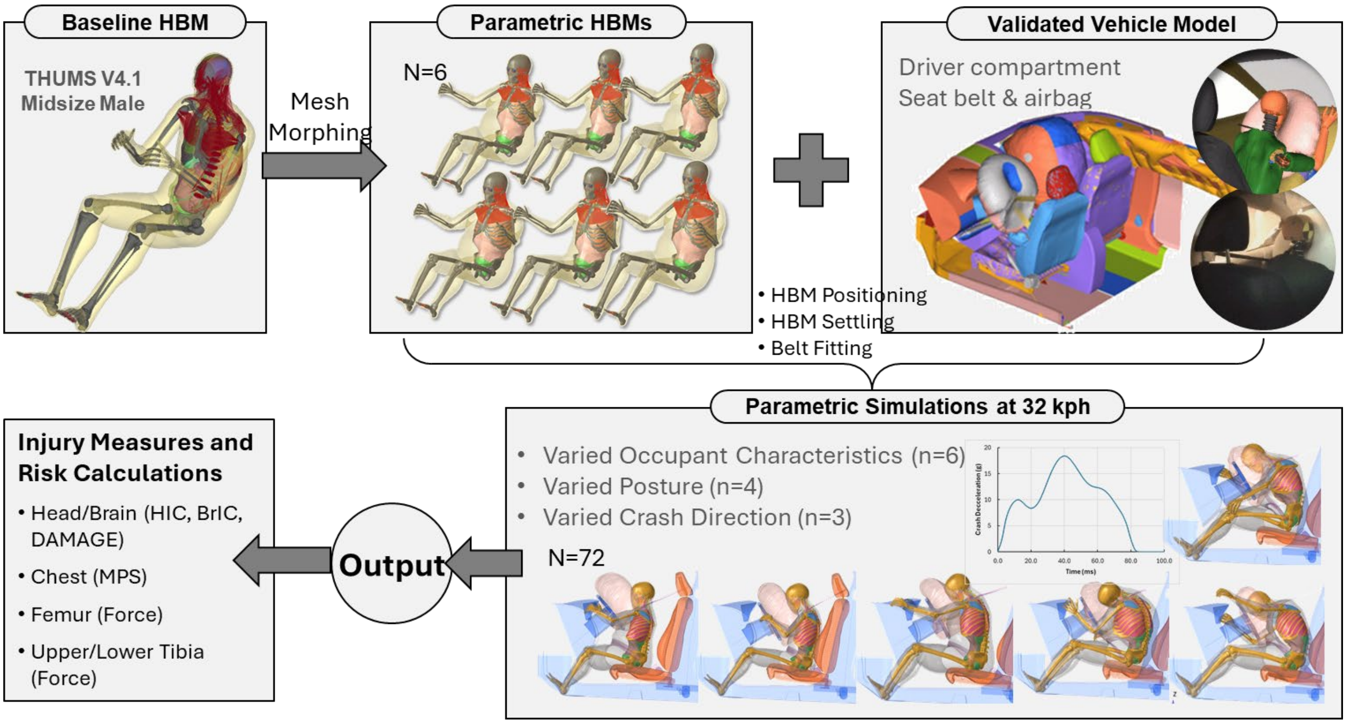

Fig. 1. Method overview for running the parametric crash simulations.

Occupant Models

THUMS version 4.1 midsize male model was used as the baseline HBM to be morphed into three male (midsize, short obese, and tall obese) and three female (midsize, short obese, and tall obese) occupants based on UMTRI's statistical human geometry models and automated mesh morphing method. The subject covariates for the six morphed HBMs are shown in Table I. Note that the age of 45-year-old was assigned for all the HBMs because we are focusing on body size and shape effects in this study rather than age. The stature values are corresponding to the 5th, 50th and 95th percentile female and male populations in the US, and the body mass index (BMI) values of 27 kg/m² and 40 kg/m² are close to the 50th and 95th BMI of the US population [14]. We prioritized higher BMI populations because field data analyses and computational studies [15,16] have shown increased injury risks of obese occupants in frontal crashes.

TABLE I
SUBJECT COVARIATES FOR THE MORPHED HBMS USED IN THIS STUDY

Sex	Stature (mm)	BMI (kg/m²)	Age (yrs)	Sitting Height / Stature SH/S
F	1498	40	45	0.52
F	1612	27	45	0.52
F	1725	40	45	0.52
M	1628	40	45	0.52
M	1770	27	45	0.52
M	1874	40	45	0.52

The details of the mesh morphing method for parametric human modelling have been well-documented in previous publications [10,17,18]. For completeness, the method is described briefly here. The morphing process started from the statistical human geometry model of the skeleton [19-23] and external body surface [24]. A demonstration version of the fully integrated full body human geometry model called HERMES [25] is publicly available online at https://humanshape.org/HERMES/MDP_web_public/HERMES_geometry_model. With any given set of subject covariates (sex, age, stature, SH/S, and BMI) for an adult, the skeleton and body shape geometry can be predicted. To morph baseline models into predicted geometries, the baseline model was linked to the statistical geometry model by mapping the baseline model's skeleton and external surface mesh onto the template mesh of the statistical geometry models. This mesh mapping process sets up the correspondence between the template meshes of the statistical geometry models and the surface meshes from the baseline models. After that, with any given age, sex, stature, SH/S, and BMI, the surface meshes of the skeleton and the external body shape in the baseline model are automatically morphed into the statistical-model-predicted geometry using a mesh morphing method based on radial basis function (RBF). These morphed surface-meshes serve as the landmarks for morphing all the other components in the whole body.

Vehicle Model

In this study, a validated driving compartment model representing Toyota RAV4 was used for all US NCAP frontal crash simulations. The RAV4 model was developed and validated as part of the UMTRI parametric vehicle model by [26]. The model was based on a generic vehicle buck model developed by [27], which includes vehicle interior, driver airbag, crushable steering column, and driver seat. This generic model was further upgraded to include the front passenger interior and associated restraint system. The updated model was morphed to the scanned RAV4 interior geometry and further validated against RAV4 crash test data under US NCAP frontal crash (35 mph) condition by tuning the seatbelt and airbag parameters. The process of mesh morphing and model validation are illustrated in Fig. 2. More detailed results of model validation can be found in a previous publication [26]. It should be noted that the knee airbag was intentionally disabled in this study to simulate worst-case scenarios, even though it is available in the vehicle model.

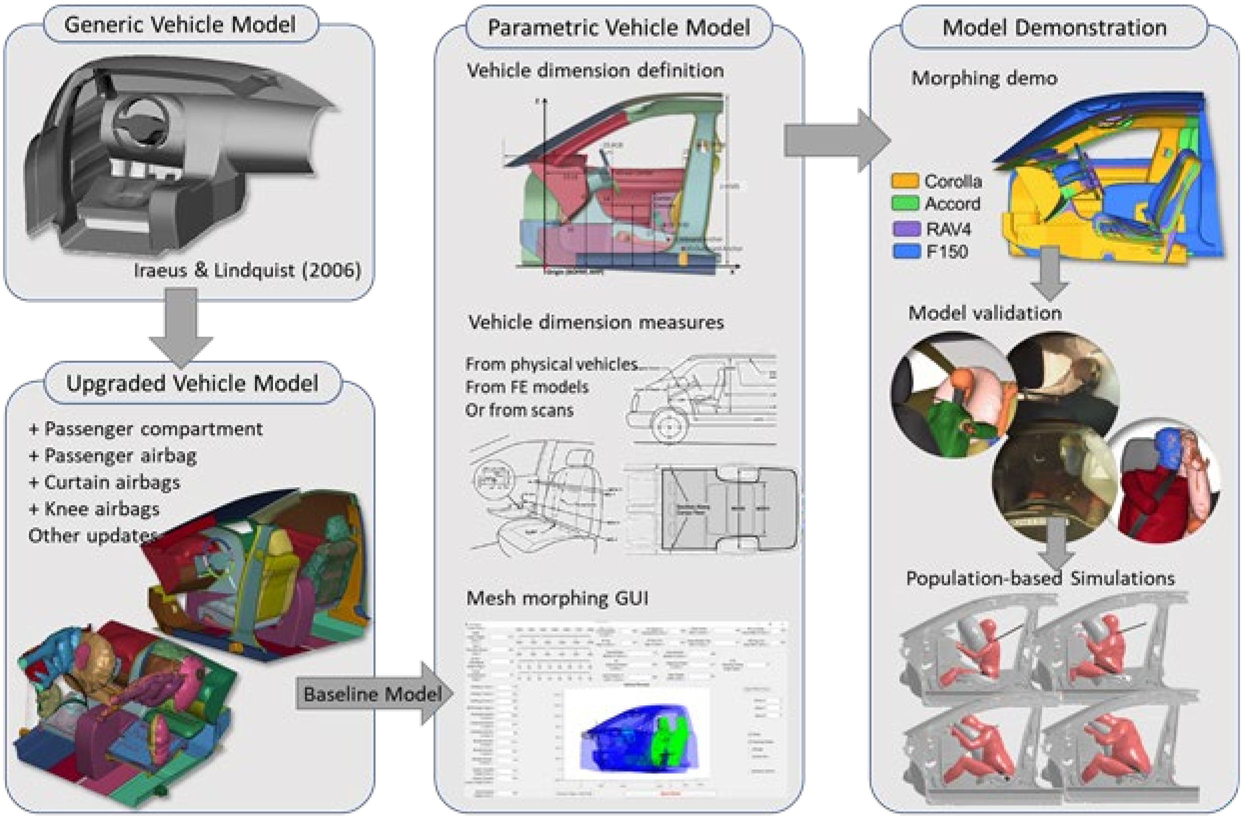

Fig. 2. Method for developing and demonstrating the parametric vehicle model [26].

Crash Conditions, Simulation Setup, and Simulation Matrix

A moderate severity crash pulse at 32 km/h (20 mph) was used for all simulations, which corresponds to ~85th percentile delta-V of tow-away crashes in the US based on NASS-CDS data. The crash pulse was based on a prediction model developed by [28], in which EDR data from the field and full vehicle crash data were used to develop the relationship between vehicle crash pulse and input parameters, including delta-V, vehicle type, impact direction, etc. Three impact directions, including full frontal and two oblique crashes (30° near-side and 30° far-side), were considered in this study. The oblique impacts were defined by rotating the frontal crash pulse by ±30°, which is consistent with the methods used in some previous studies [29].

We established a robust method to integrate the HBM, restraint, and vehicle models together for vehicle crash simulations according to our previous work [8]. Specifically, we used a statistical seating posture model based on volunteer seating posture data to predict the head and hip locations with a given set of occupant characteristics and vehicle packaging parameters. The repositioning tool calculates the joint angles associated with each joint, and each of the morphed HBMs was morphed into the target posture. A belt-fitting tool in MATLAB was used to automatically fit the seatbelt on the occupant model based on the belt-fit model from volunteer data.

A normal driving posture and three non-nominal driver postures, including slouching (20° pelvis angle tilting backward), 11° inboard leaning, and 11° outboard learning, were considered. The 11° torso lean is the maximum angle that the taller HBM can achieve without contacting the vehicle interior. We chose these non-nominal seating positions as an initial focus to explore how posture influences occupant injury risk in moderate-severity crashes. Forward-leaning posture has been relatively well studied in previous research [30,31], and was therefore excluded from the current analysis, though it may be considered in the future. Mesh morphing was used to reposition the morphed THUMS models into the target postures.

Given the six HBMs, four driver postures, and three impact directions, a full factorial design resulted in a total of 72 (6x4x3) simulations in this study.

Injury Measures and Risk Curves

Occupant injury measures to the head (HIC, BrIC, and DAMAGE), chest (maximal principal strain [MPS]), femur (force), upper tibia (force), and lower tibia (force) as well as occupant impact kinematics were collected in all simulations. The injury risk curves are shown in the Appendix. It should be noted that MPS was used for the chest injury assessment based on the THUMS 3+ injury risk curve developed by [32], because a previous study found that MPS-based chest injury risks can better reflect the injury causation than the single-point chest deflection at the mid-sternum in frontal crashes [33].

III. RESULTS

Simulation Result Overview

All 72 simulations were run smoothly without any error termination, demonstrating the robustness of the morphed HBMs and the simulation framework. As expected, the injury risk for each body region is generally low compared to the high-speed NCAP frontal crashes. Specifically, the AIS2+ femur, upper tibia, and lower tibia injury risks are less than 10% in all simulated cases, with the average injury risk at 1.5±0.7%, 1.4±0.6%, and 4.4±1.8% for the three body regions, respectively. The AIS3+ chest injury risk varied much more than the lower extremity injury risk, ranging from 0.4% to 34.7% and with an average of 4.9%. The head injury risk depends on the injury measure. HIC-based injury risk is not sensitive to the HBM and crash conditions, consistently predicting the injury risk close to zero in all conditions, while BrIC predicted AIS3+ brain injury risk at 16.0±15.8% and DAMAGE predicted AIS2+ brain injury risk at 27.7±18.1% and AIS4+ brain injury risk at 3.1±3.1%.

Sex Effects

Figure 3 shows the sex effects on injury risks for different body regions. Note that head injury risk based on HIC was neglected because they are all close to zero. Female occupants showed higher lower extremity injury risks compared to the male occupants, but there is no clear difference in brain injury risks between male and female HBMs. In terms of the chest injury risk, all the cases with chest injury risk greater than 15% are obese male occupants leaning inboard. Obese female occupants leaning inboard also showed increased chest injury risks, but not as high as their male counterparts.

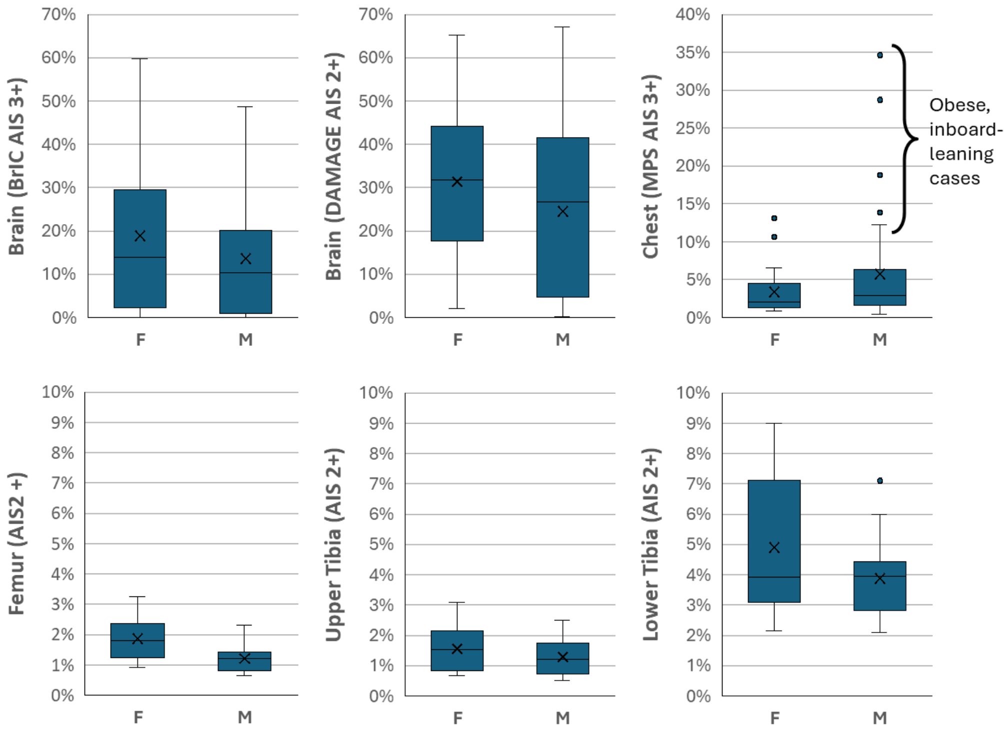

Fig. 3. Sex effects on occupant injury risks in all simulated conditions.
Note: the box represents the middle 50% spanning from the first to the third quartile. The line and x within the box mark the median and mean, respectively. Whiskers are the lines extending from the box to the minimum and maximum values, excluding outliers. Points that fall outside the whiskers are marked as outliers, indicating values that are significantly different from the rest of the data.

BMI Effects

Figure 4 shows the BMI effects on injury risks for different body regions. Higher BMI occupants tend to have substantially higher femur and tibia injury risks compared to normal-BMI occupants, but higher BMI occupants also tend to have slightly lower brain injury risks albeit not significant. As mentioned earlier, obese occupants who leaned inboard had higher chest injury risks compared to all other simulated cases.

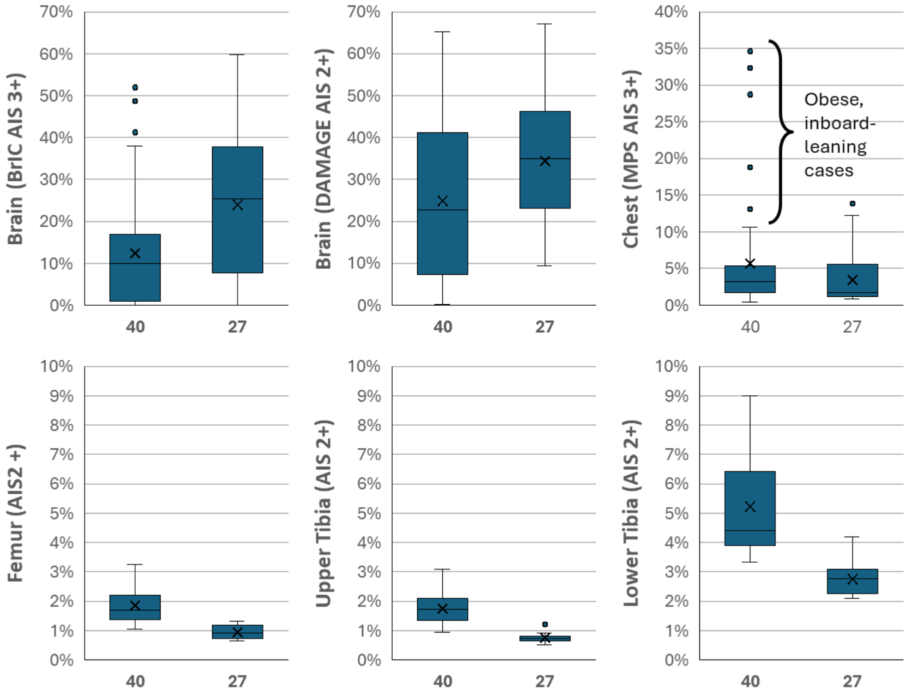

Fig. 4. BMI effects on occupant injury risks in all simulated conditions.

Posture Effects

Figure 5 shows the occupant posture effects on injury risks for different body regions. Compared to the sex and BMI effects, the posture effects on occupant injury risks are not as strong. The only significant posture effects are when occupant leaning inboard, which increases the chest injury risks, and the slouch effect on lower tibia. It should also be noted that the slouch posture increases the femur and tibia injury risks for every HBM compared to other postures.

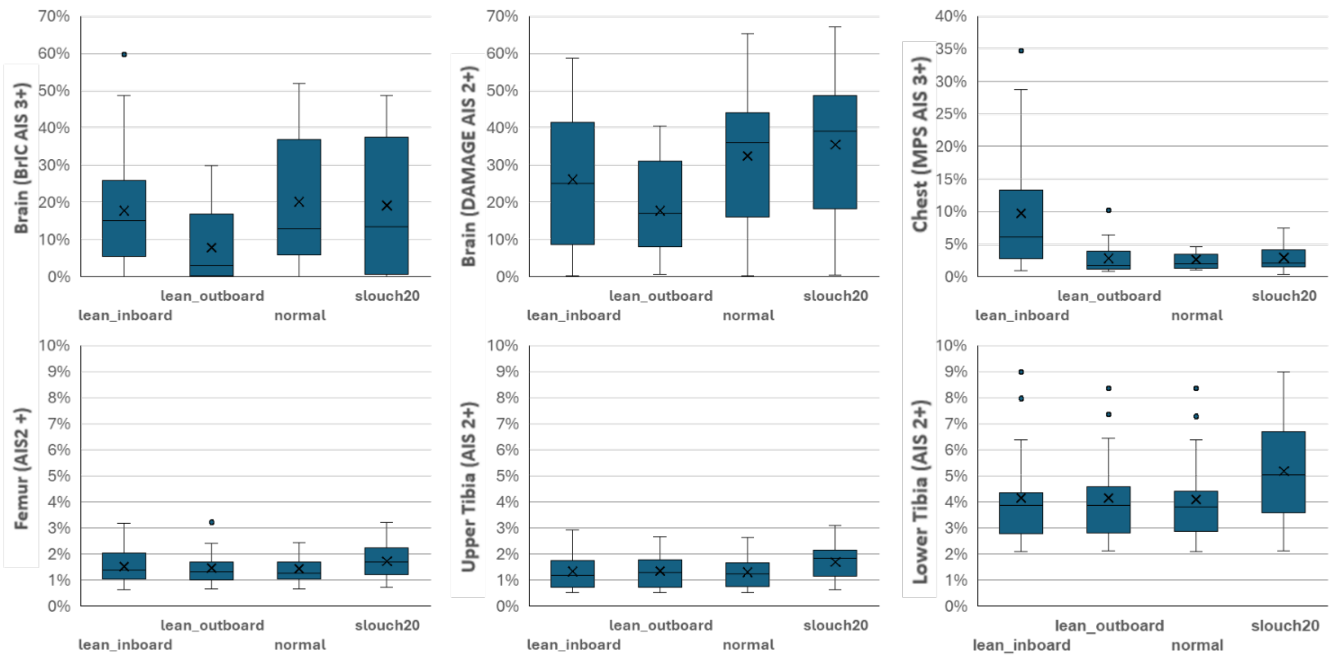

Fig. 5. Occupant posture effects on occupant injury risks in all simulated conditions.

Impact Direction Effects

Figure 6 shows the impact direction effects on injury risks for different body regions. In general, there is no clear effect from impact direction on any of the body regions. The only exception is the brain injury risk predicted by the BrIC. In oblique impacts, the head of the HBM tends to rotate laterally, which increases the BrIC value. However, DAMAGE did not show the same trend.

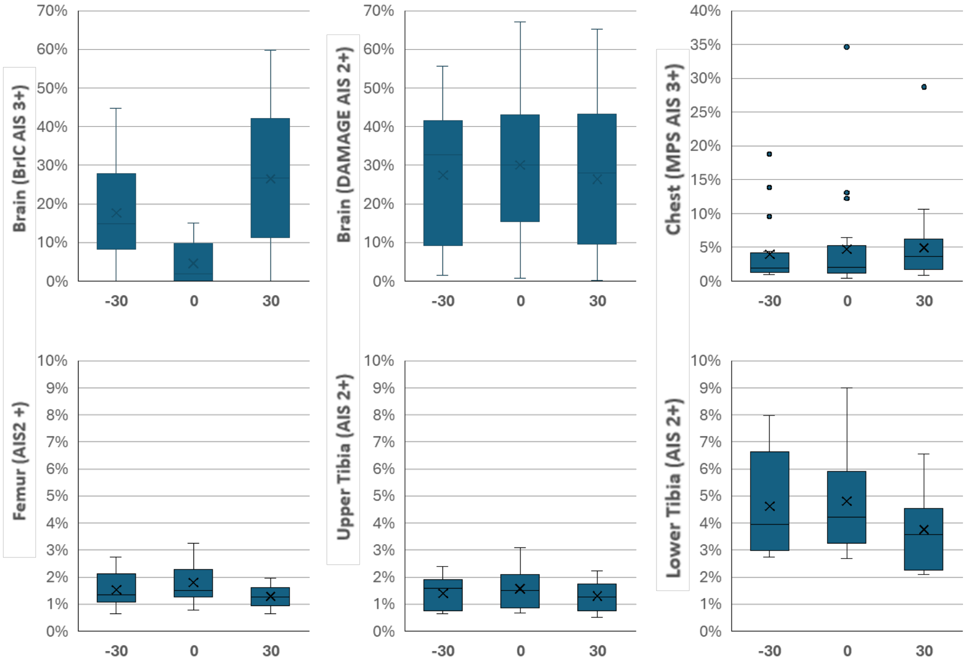

Fig. 6. Impact direction effects on occupant injury risks in all simulated conditions.

IV. DISCUSSION

General Trends

In this study we performed 72 frontal crash simulations at 32 km/h, varying occupant size, shape, posture, and impact direction. Several trends emerged that may contribute to increased injury risks in moderate severity crashes. Female occupants exhibited higher lower extremity (including femur and tibia) injury risks compared to male occupants, and obese occupants faced higher lower extremity injury risks than their non-obese counterparts. These trends were consistently observed in the simulations and align with findings from field data analyses [34]. Additionally, we found that an inboard-leaning posture is associated with elevated chest injury risks, while slouching increases tibia injury risks. Among all body regions, the brain exhibited the highest overall injury risk, with oblique impacts leading to higher brain injury risks based on BrIC, but not on DAMAGE.

Lower Extremity Injury Risks

Based on previous post-mortem human subject (PMHS) tests and HBM studies, the higher risk of lower extremity injury associated with obese occupants is mainly due to the adipose tissues, especially around the abdominal area, which increase the mass and induce poor belt fit, leading to greater forward body excursions. Women tend to have higher percentage of body fat compared to men, sit closer to the knee bolster, and have lower injury tolerance, all of which may have contributed to their higher risks for lower extremity injury. Figure 7 shows kinematic comparison between midsize female, short obese female, and short obese male in full frontal crash condition. The short obese female had the largest body excursion among the three HBMs, followed by the short obese male, and the midsize female had the smallest body excursion. This is likely due to a combined effect from body mass, belt fit, and seating location, as explained earlier. Furthermore, the chest of the short obese female model pushed the airbag up, leading to a submarining-type of kinematics (i.e. higher knee excursion than head excursion), which may also have contributed to the higher lower extremity injury risk.

We also found that a slouching posture increases the risk of lower extremity injuries. In this study, slouching was defined as a 20^{0} backward tilt of the pelvis while maintaining the HBM head position. This posture change shifts the torso away from the driver airbag, causing the lower extremities to absorb more crash energy than a normal driving posture and resulting in a higher lower extremity injury risk.

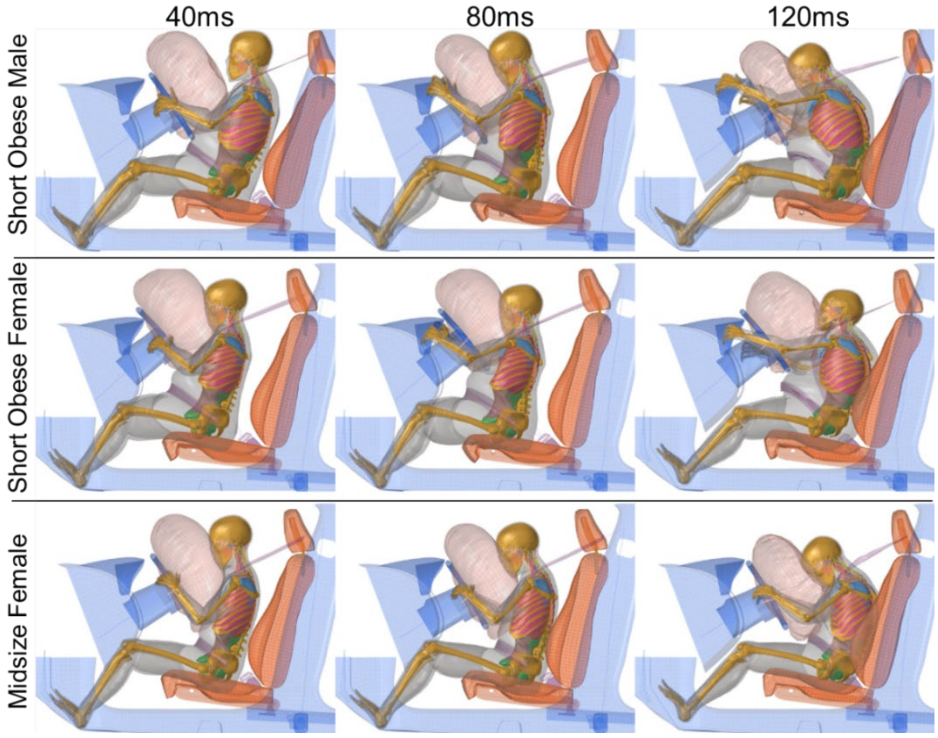

Fig. 7. Kinematics comparison between occupants with respect to sex and obesity in full frontal crash condition.

Chest Injury Risks

We found that an inboard-leaning posture can elevate chest injury risk, particularly for obese male occupants. Figure 8 illustrates the ribcage MPS distribution and occupant kinematics for the tall obese male HBM with two postures under full frontal crash conditions. At around 60 ms, the seatbelt fully engaged with the upper right side of the chest, resulting in elevated MPS values in the right 3rd to 5th ribs. By approximately 80 ms the airbag also made full contact with the chest. The combined forces from the belt and airbag caused compression on the right side of the ribcage and tension on the left, leading to higher MPS in the right upper chest and left lower chest. The variation in rib MPS values between the inboard-leaning and normal driving postures is primarily due to differences in belt positioning relative to the chest. Overall, the combined effects from obesity, inboard leaning, and belt locations contributed to the increased risk of chest injury.

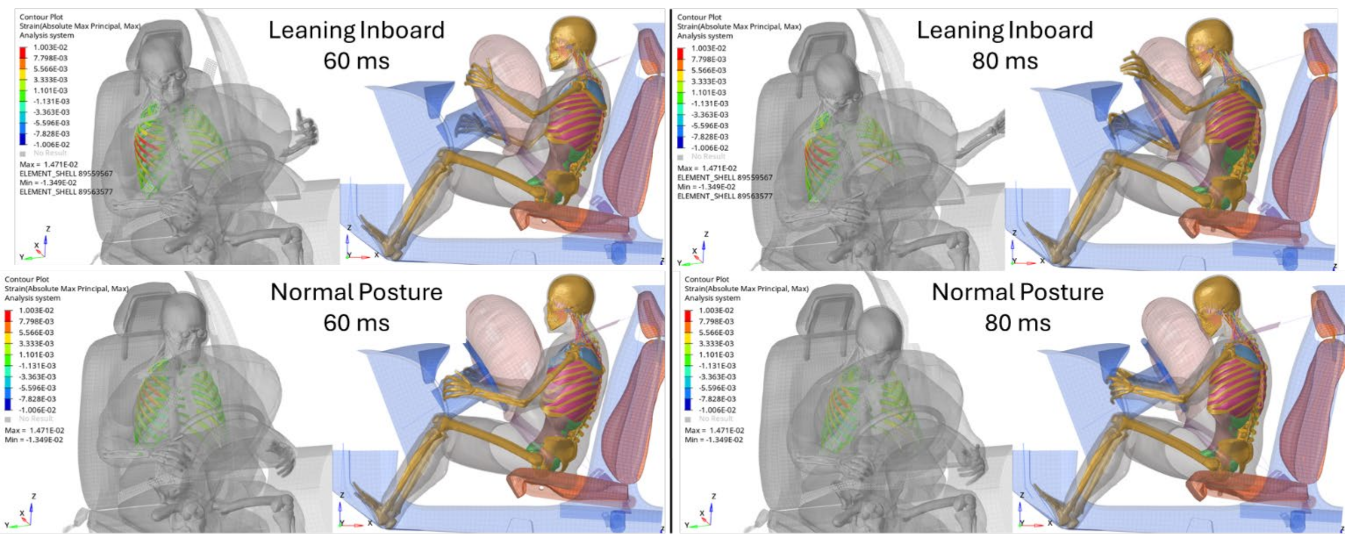

Fig. 8. MPS distribution and occupant kinematics for the tall obese male HBM with two postures in the full frontal crash condition.

Brain Injury Risks

In this study we used three injury measures to evaluate the head/brain injury risks, HIC, BrIC, and DAMAGE. Head injury risks by HIC are all close to zero, but BrIC and DAMAGE predicted much higher brain injury risks. Furthermore, our simulation results show that BrIC is highly sensitive to impact direction, while DAMAGE is not. Figure 9 shows the kinematics of midsize female model leaning inboard in far-side oblique impact condition. Due to the initial leaning posture and the oblique impact angle, the head of the occupant rolled off the airbag, causing significant lateral rotation and resulting in a brain injury risk of 49% by BrIC and 40% by DAMAGE. Interestingly, we found that the correlation between the brain injury risks predicted by BrIC and DAMAGE is low (R^2=0.116), as shown in Fig. 10, which may need further investigation.

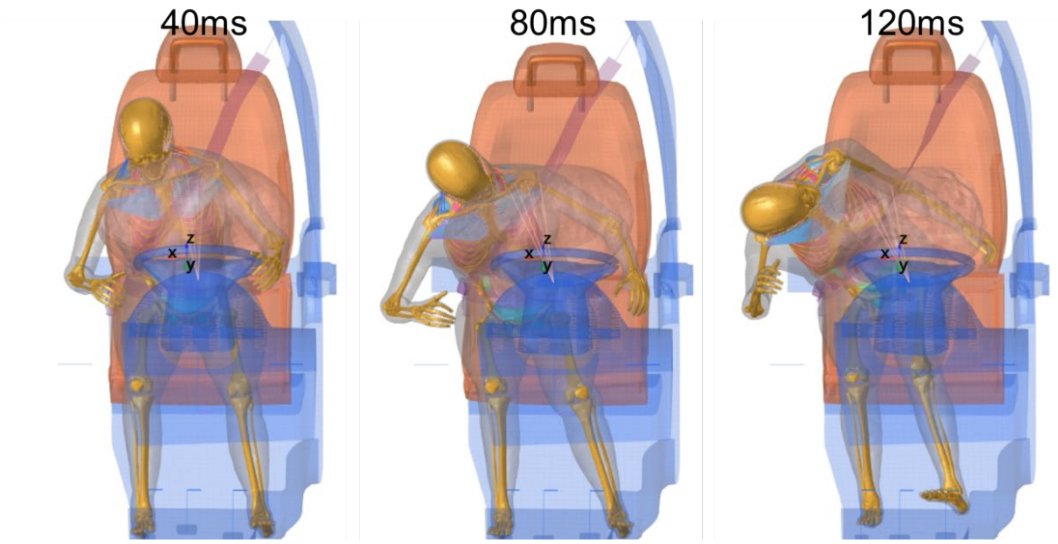

Fig. 9. Kinematics of midsize female model leaning inboard in far-side oblique impact condition.

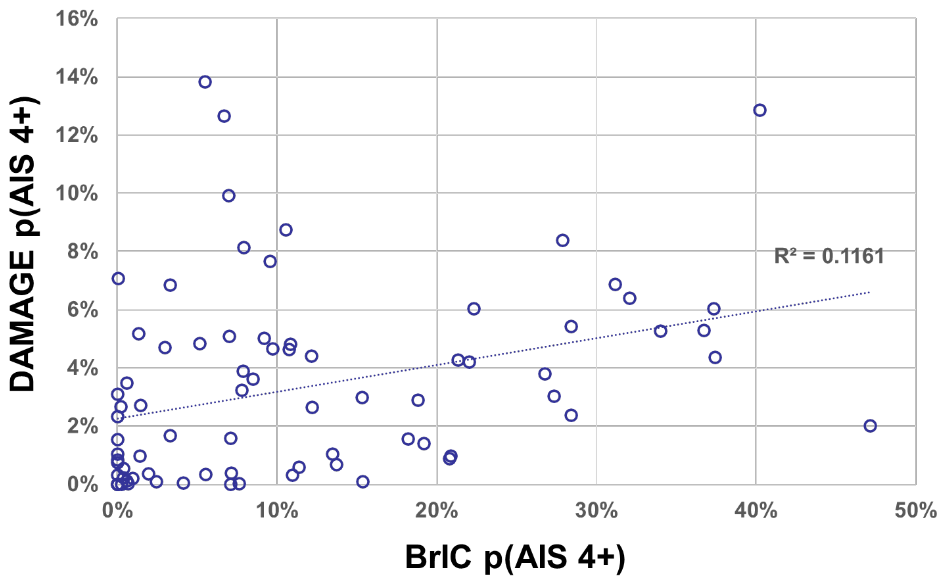

Fig. 10. Correlations of brain injury risks predicted by BrIC and DAMAGE.

Limitations and Future Work

This study has several limitations. First, all simulations were performed using a single vehicle driver compartment model and a single crash pulse. As a result, the findings may not be generalisable to other vehicle types, seating positions, or crash scenarios. Second, strain-based injury assessment was applied solely for chest injury prediction and not extended to other body regions. Given that THUMS can estimate MPS for the brain and lower extremities, future research could broaden strain-based injury evaluations to additional body regions as relevant injury risk curves become available. Furthermore, we employed multiple injury metrics for head injury assessment, but relied on a single metric for evaluating chest and lower extremity injuries. The scaling method for femur and tibia injury risk curves may be refined in the future. Third, the sample size of HBMs was limited to six, suggesting that future work should consider a larger sample to more fully capture population variability. Fourth, this study examined only a limited number of variables affecting occupant injury risk, leaving out many other potentially relevant factors. Future research could incorporate additional considerations such as occupant age and physical fragility, alternative out-of-position postures like forward leaning, and scenarios where safety systems do not function optimally, such as delayed deployment of airbags [35] or seatbelt pretensioners. Despite these limitations, the varied HBMs used in this study clearly demonstrated that occupant size, shape, and posture have a substantial influence on injury risk in moderate severity frontal crashes. Future population-based simulations with a larger sample size, considering crashes at varied severities, vehicle designs, and occupant variability, could further improve our understanding of crash severity effects on occupant protection.

V. CONCLUSION

To the best of our knowledge, this is the first study to apply a diverse set of HBMs to investigate driver injury mechanisms and risk factors in moderate severity frontal crashes. Six HBMs, representing a broad spectrum of body sizes and shapes, were used and combined with a validated SUV driver compartment model, a 32 km/h crash pulse, and three impact angles, resulting in a total of 72 crash simulations. The results showed that obese occupants experienced significantly higher risk of lower extremity injury compared to non-obese occupants. Female occupants also exhibited higher risk of lower extremity injury than male occupants. Inboard-leaning postures were found to be associated with elevated chest injury risks, particularly for obese occupants, while slouched postures were associated with increased tibia injury risks. Brain injury risks were generally higher than those for other body regions. Oblique impacts led to higher brain injury risks based on BrIC, but not when assessed using DAMAGE, and a low correlation between BrIC and DAMAGE was observed, warranting further investigation. The highest chest injury risk across all simulated scenarios occurred in obese male occupants with an inboard-leaning posture.

Overall, this study highlights that occupant characteristics and postures substantially influence occupant injury risks in moderate severity frontal crashes, with impact angle playing a particularly important role in brain injury risk.

VI. ACKNOWLEDGEMENTS

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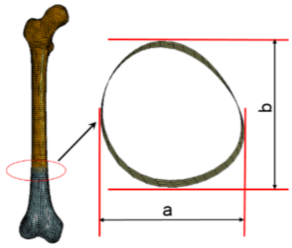
VIII. APPENDIX

INJURY MEASURES AND RISK CURVES USED IN THIS STUDY

<table>
<tr><th>Body Region
(Injury Measure)</th><th>Injury Risk Curve</th><th>Scaling</th></tr>
<tr><td>Femur
and Tibia
(F in kN)</td><td>

$$P_{KTH}(AIS2+) = \frac{1}{1+e^{5.795-0.5196F}}$$
$$P_{TibiaU}(AIS2+) = \frac{1}{1+e^{5.7415-0.8189F}}$$
$$P_{TibiaL}(AIS2+) = \frac{1}{1+e^{3.7544-0.4683F}}$$

</td><td>



$$F = F_{subject}\times\frac{a_{midsize}\times b_{midsize}}{a_{subject}\times b_{subject}}$$

</td></tr>
<tr><td>Head
(HIC15)</td><td colspan="2">

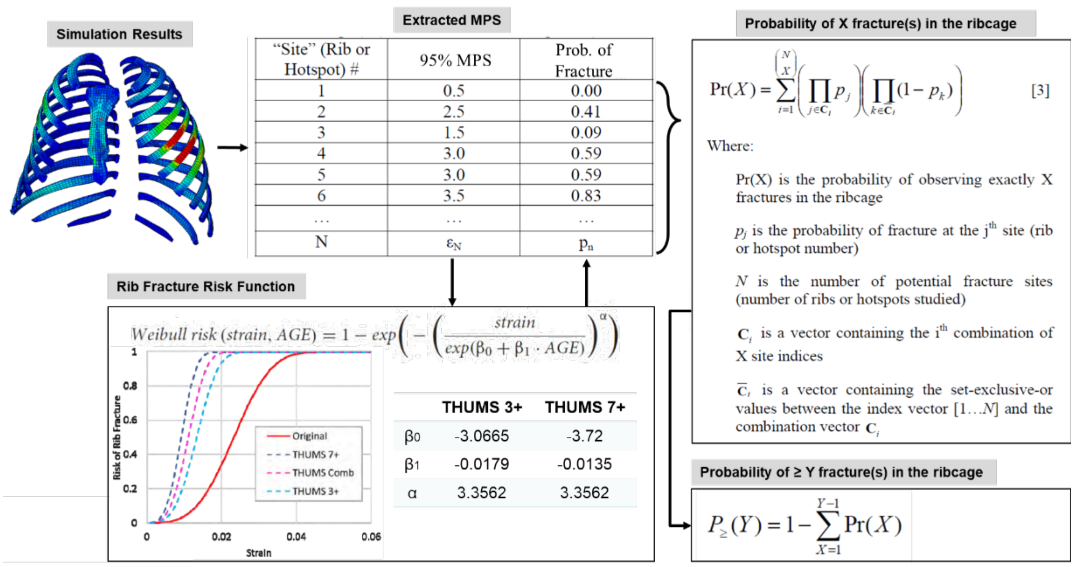
$$\begin{cases} P_{HIC}(AIS3+) = \emptyset\left(\dfrac{ln(HIC15)-7.45231}{0.73998}\right) \\ \text{where } \emptyset = \text{cumulative normal distribution} \end{cases}$$

</td></tr>
<tr><td>Brain
(BrIC and
DAMAGE)</td><td colspan="2">

$$P_{BrIC}(AIS\ 3+) = 1 - e^{-\left(\frac{BrIC-0.523}{0.531}\right)^{1.8}}$$
$$P_{BrIC}(AIS\ 4+) = 1 - e^{-\left(\frac{BrIC-0.523}{0.647}\right)^{1.8}}$$

$$P_{DAMAGE}(\text{AIS }2+) = 1 - e^{-e^{3.875\times(\log(0.957\times DAMAGE+0.017)-\log(0.459))}}$$
$$P_{DAMAGE}(\text{AIS }4+) = 1 - e^{-e^{6.051\times(\log(0.957\times DAMAGE+0.017)-\log(0.646))}}$$

</td></tr>
<tr><td>Chest
(MPS
distribution)</td><td colspan="2">



</td></tr>
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