Evaluation of kinematics and injuries to restrained occupants in far-side crashes using full-scale vehicle and human body models

Mike W. J. Arun, Sagar Umale, John R. Humm, Narayan Yoganandan, Prasanaah Hadagali, and Frank A. Pintar

Department of Neurosurgery, Medical College of Wisconsin, Milwaukee, Wisconsin

ABSTRACT

Objective: The objective of the current study was to perform a parametric study with different impact objects, impact locations, and impact speeds by analyzing occupant kinematics and injury estimations using a whole-vehicle and whole-body finite element–human body model (FE-HBM). To confirm the HBM responses, the biofidelity of the model was validated using data from postmortem human surrogate (PMHS) sled tests.

Methods: The biofidelity of the model was validated using data from sled experiments and correlational analysis (CORA). Full-scale simulations were performed using a restrained Global Human Body Model Consortium (GHBMC) model seated on a 2001 Ford Taurus model using a far-side lateral impact condition. The driver seat was placed in the center position to represent a nominal initial impact condition. A 3-point seat belt with pretensioner and retractor was used to restrain the GHBMC model. A parametric study was performed using 12 simulations by varying impact locations, impacting object, and impact speed using the full-scale models. In all 12 simulations, the principal direction of force (PDOF) was selected as 90°. The impacting objects were a 10-in.-diameter rigid vertical pole and a movable deformable barrier. The impact location of the pole was at the C-pillar in the first case, at the B-pillar in the second case, and, finally, at the A-pillar in the third case. The vehicle and the GHBMC models were defined an initial velocity of 35 km/h (high speed) and 15 km/h (low speed). Excursion of the head center of gravity (CG), T6, and pelvis were measured from the simulations. In addition, injury risk estimations were performed on head, rib cage, lungs, kidneys, liver, spleen, and pelvis.

Results: The average CORA rating was 0.7. The shoulder belt slipped in B- and C-pillar impacts but somewhat engaged in the A-pillar case. In the B-pillar case, the head contacted the intruding struck-side structures, indicating higher risk of injury. Occupant kinematics depended on interaction with restraints and internal structures—especially the passenger seat. Risk analysis indicated that the head had the highest risk of sustaining an injury in the B-pillar case compared to the other 2 cases. Higher lap belt load (3.4 kN) may correspond to the Abbreviated Injury Scale (AIS) 2 pelvic injury observed in the B-pillar case. Risk of injury to other soft anatomical structures varied with impact configuration and restraint interaction.

Conclusion: The average CORA rating was 0.7. In general, the results indicated that the high-speed impacts against the pole resulted in severe injuries, higher excursions followed by low-speed pole, high-speed moving deformable barrier (MDB), and low-speed MDB impacts. The vehicle and occupant kinematics varied with different impact setups and the latter kinematics were likely influenced by restraint effectiveness. Increased restraint engagement increased the injury risk to the corresponding anatomical structure, whereas ineffective restraint engagement increased the occupant excursion, resulting in a direct impact to the struck-side interior structures.

Introduction

Occupants seated closer to the struck side in an automobile accident are defined as near-side crashes. On the contrary, far-side crashes are defined as when occupants are seated opposite to the struck side. Yoganandan et al. (2014) queried far-side cases from 2009 to 2012 using the NASS-CDS database, which resulted in 18,301 Maximum Abbreviated Injury Scale (MAIS) 2+ cases. In general, injury mechanisms in far-side impacts are thought to be significantly different from near-side impacts. Therefore, different strategies for countermeasures to protect occupants in far-side crashes may be necessary (Bostrom et al. 2008).

Previous studies based on NASS-CDS data have indicated that the head is more likely to sustain severe injuries, followed by chest and abdomen, in far-side crashes. In addition, these studies reported that the struck-side interior was the most frequent contacting structure associated with the vehicle occupant, followed by seat belt and passenger seat (Augenstein et al. 2000; Digges and Dalmotas 2004). Fildes et al. (2007) reported that the rib cage is the most frequently injured anatomical structure in the chest, followed by lungs, whereas the liver is the most frequently injured organ in the abdomen, followed by the spleen. A more recent study based on 111 Crash Injury Research and...
Engineering Network (CIREN) cases indicated that occupants sustain pelvic fractures in far-side crashes probably due to belt loading (Halloway 2016).

Kent et al. (2013) performed sled tests using postmortem human surrogate (PMHS) to simulate occupant response in far-side crashes. The study reported injuries including rib fractures; however, the study either minimized or neglected a realistic boundary condition including intrusion, deformable seat pan and seatback, center console, etc. Pintar et al. (2007) performed 18 sled tests on 6 PMHS seated on a generic far-side buck at low (11 km/h) and high (30 km/h) speeds. The study compared the PMHS responses with those of mechanical surrogates; a limitation of the study, however, was reduced boundary conditions that might influence the kinematics and kinetics of the specimens.

Therefore, the information currently available regarding occupant kinematics in far-side crashes is based on accident databases and experiments using simplified boundary conditions. The literature is lacking a representational occupant kinematics using realistic boundary conditions. In addition, the kinematics of vehicles and occupants under varying impact far-side setups, such as impact location and principal direction of force (PDOF), is also equally less understood. Although conducting far-side experiments using a whole vehicle and PMHS is an option, such experiments are expensive in cost and human resources. Additionally, even with the availability of resources, the design of experiments might result in a very large testing matrix due to the lack of prior biomechanics data on far-side impact conditions. In addition, response variations due to the diversity in PMHS anthropometry may impede delineating certain injury mechanisms. An alternative is to use validated whole-vehicle and whole-body finite element (FE) human body models (HBMs) for these types of studies. Simulations with HBM may be used to supplement experiments by performing parametric studies with different impact objects, impact locations, impact speeds, etc., that will reduce the experimental matrix to a manageable size by eliminating boundary conditions that are less important with greater confidence. In addition, HBM represents a specific target population avoiding uncertainties due to anthropometry variations.

Therefore, the objective of the current study was to perform a parametric study with different impact objects, impact locations, and impact speeds by analyzing occupant kinematics and injury estimations using a whole-vehicle, whole-body FE-HBM. To confirm the HBM responses, the biofidelity of the model was validated using data from PMHS sled tests.

**Methods**

The Global Human Body Model Consortium (GHBMC) FE-HBM was used in the current study. The biofidelity of the model was evaluated using matched-pair simulations using far-side sled test data. The goodness of fit with the sled test data was quantified using correlation and analysis (CORA) methodology. Following the head, torso, and pelvis biofidelity evaluation of the HBM, the model was seated inside a Taurus 2001 whole-vehicle model. A parametric study was performed with different impact locations, impact objects, and speeds. Finally, the occupant kinematics and injury estimates were compared between the cases to identify the worst-case scenario among the simulated cases.

**FE-HBM validation**

The kinematics and kinetics of the human body model were reasonably validated in the frontal and near-side impacts on a gross scale (Hayes et al. 2014; Li, Kindig, Kerrigan, et al. 2010; Li, Kindig, Subit, and Kent 2010; Park et al. 2013, 2014; Vavalle et al. 2015); in addition, the pelvis and the soft organs of the model were validated using force vs. penetration data from PMHS tests (Beillas and Berthet 2012; Yue et al. 2011). However, the model lacks biofidelity evaluation in far-side impacts. Thus, in the present study the head, torso, and pelvis of the GHBMC model were evaluated for biofidelity under far-side impact using kinematics and kinetics test data. This evaluation was performed by correlating the anatomical model response data with similar data from matched-pair sled experiments. To quantify the correlation, CORA was performed using the HBM model and experimental data.

**PMHS far-side sled test data**

The experiments used in this study were previously conducted by Pintar et al. (2007). Only a brief description of their test setup is presented here; a detailed description can be found in Pintar et al. (2007). Three PMHS (2 male, 1 female; age, 82 ± 2 years; height, 174 ± 0.2 cm; weight, 71 ± 8 kg) were tested on a far-side impact buck with an assumption of seated driver struck on the passenger side at a 3-o'clock PDOF (90°). The buck system included a seat pan, seatback, 3-point belt system, horizontal center console, and vertical lateral load plates designed to engage specific anatomical regions (pelvis and legs). The plates were independently mounted on triaxial load cells that measured loads specific to the respective anatomical region. The center console was also mounted on triaxial load cells that were supported by a steel frame. The 3-point belt system included a standard low-elongation lap and shoulder belts that are anchored at standard locations.

The leg and pelvis load plates were instrumented with 3 and 2 load cells, respectively. The center console was instrumented with another 2 load cells. The PMHS were also instrumented with triaxial accelerometer arrays at the T1, T12, and sacrum. A custom-designed Pyramid Nine Accelerometer Package attached to the surface of the head was used to derive the accelerations at the head center of gravity (CG; Yoganandan et al. 2007). All coordinate systems followed SAE-J211 standard sign convention. Due to the variations in PMHS anthropometry, all sensor data were scaled to the 50 percentile male anthropometry using an equal stress and equal velocity method (Eppinger et al. 1999).

**FE-HBM validation simulation setup**

A detailed finite element model of the experimental setup was created using various element types and appropriate materials (Figure 1). The rigid seat pan and backrest were constructed using shell elements and assigned steel material property for contact approximations. A rigid pole representing the B-pillar was constructed using shell elements and used to anchor the shoulder belt. The dimensions of the load walls were directly taken from the physical experimental setup. Each load wall was
constructed using two parts: a rigid plate and paper honeycomb. The plates on the load walls were constructed using shell elements and the paper honeycombs were constructed using solid elements (208 kPa paper honeycomb property). A 3-point seat belt system was used to restrain the GHBMC model. The seat belt material property corresponded to 5% elongation at 11 kN. The D-ring anchor point was in line with the outside of the shoulder in the anterior–posterior direction. Similar to the experimental setup, the D-ring was 90 mm above the superior surface of the shoulder and 120 mm behind the midpoint of the shoulder. A friction coefficient of 0.3 was used to simulate the contact between the PMHS and the seat pan (Park et al. 2013). A detailed description of the simulation setup is given in Appendix C2 (see online supplement).

In addition to CORA ratings, mean ± SD was calculated for the regional nodal accelerations of the HBM and contact forces between the plates and HBM. In addition, AIS of the rib fractures were compared between the simulation and experiments. The simulations used an element deletion technique to simulate rib fractures.

Correlation analysis
A detailed description on CORA can be found in Appendix C1. (see online supplement).

Full-scale simulations

FE-HBM and whole vehicle models simulation setup
Full-scale simulations were performed using a restrained GHBMC model seated on a sedan using far-side lateral impact condition. A 2001 Ford Taurus was selected from the National Crash Analysis Center consortium for the current study. The kinematics (CG acceleration, floor accelerations, change in CG velocity) and kinetics (impact force vs. crush distance) of the vehicle model were validated in both frontal and side impact modes (Marzougui et al. 2012). However, the default original equipment manufacturer (OEM) seat position of this model was full-forward. In all of the simulations, to simulate a nominal seating position, the fore–aft position of the driver and passenger seats was adjusted to their mid-positions. The GHBMC model was translated and rotated and placed just above the OEM seat. The HBM was then settled on the OEM seat using acceleration due to gravity. The same simulation was also used to gravity settle the whole-vehicle model on the rigid floor that was constructed using shell elements. A 3-point seat belt system was used to restrain the GHBMC model. The system also included a pretensioner (100 mm pull-in at 10 ms) and retractor with a 4-kN load limiter. The pretensioner was triggered using the crash pulse to encumber the GHBMC model, and the load limiter maintained a constant load of 4 kN on the shoulder belt to simulate a realistic loading condition. A parametric study was performed by varying impact locations, impacting object, and impact speed using the full-scale models. In all 12 simulations, the PDOF was selected as 90°. The impacting objects were a 10-in.-diameter rigid vertical pole and a movable deformable barrier (MDB; Figure 2). The MDB was validated using force vs. displacement and velocity data from tests (Balsod and Krebs 2011). The impact locations were at the C-pillar in the first case, at the B-pillar in the second case, and at the A-pillar in the third case. The vehicle and the GHBMC models were defined as an initial velocity of 35 km/h (high speed) and 15 km/h (low speed). The combination of these parameters resulted in 12 unique cases. The lap and shoulder belt forces were measured from a 1D belt element and retractor, respectively. Because of the gross rotation and translation of the vehicle, the excursion measurements of the HBM at head CG, T6, and pelvis with respect to the global coordinate system may not be appropriate. Therefore, a local coordinate system was created on the vehicle model, which translated and rotated with the model.

Injury estimation in the FE-HBM
To colligate the injuries obtained from simulations to that reported in the literature, injury predictions were performed for head, rib cage, lungs, kidneys, liver, spleen, bladder, and pelvis. Head injury was determined by verifying its contact
kinematics with the interior structures. A strain-based normalized relative injury severity estimation was used for the soft organs. A predefined strain threshold value specific to an anatomical structure was used to check whether an element in the structure indicated a likelihood of an injury. For each anatomic structure, the elemental strains were collected every 2.5 ms and compared to the corresponding strain threshold. If an element exceeded the threshold, the volume of the element was added cumulatively, and this process was repeated for all elements in an organ. Finally, the percentage of volume of the elements that exceeded the threshold was calculated using the total volume of the organ. Although higher strain values at a small region are sufficient to initiate failure in organs, it is reasonable to assume that higher "injured" volume may indicate higher likelihood of an injury. For each organ, the injured volume was normalized using the highest value among the 12 impact cases, namely, A-, B-, and C-pillar impacts with pole and MDB under 2 impact speeds. Although this method is not equivalent to the traditional injury risk estimation using an injury risk curve, it shows the relative injury severity in the soft organs between the different simulated cases. The rib cage and pelvis were evaluated for fractures using an element elimination technique that used a strain threshold value of 1.8%. Appropriate AIS scores were assigned based on the number of observed fractures in the rib and pelvis. Nominal strain threshold values of 15, 30, 20, 30, and 65% were used for lung, liver, spleen, kidney, and bladder, respectively (Kemper et al. 2012; Martins et al. 2011; Snedeker et al. 2005; Untaroiu et al. 2015; Yamada and Evans 1970; Appendix B2, see online supplement). A custom MATLAB script was used to extract all of the values from the LS-DYNA output database and to perform the normalized relative injury severity estimation in the soft organs. In addition, to estimate the gross injury risk of the soft organs, the viscous criterion was calculated using virtual chest bands. Preprogrammed virtual chest bands (upper, middle, and lower) were used to derive the peak chest deflection of the HBM and the nodal data were used to calculate the maximum viscous criterion (Lau and Viano 1986). Nodal displacement data were used to calculate the peak chest deflection (Yoganandan et al. 2011). Because of the postimpact excursion (both translation and rotation) of the HBM, in order to overlay the deformed chest band with the initial state, 2 points from the rigid vertebral body were used as a reference of alignment. For each time step, these reference points were employed to perform affine transformations on the deformed chest contour to move it to the reference chest contour. These operations were performed by using a custom MATLAB script.

Results

FE-HBM validation

Figure A1 (see online supplement) shows the initial torso deformation due to the belt tightening simulation. A visual comparison of kinematics between the simulation and a PMHS experiment is shown in Figure 3 and, overall, the kinematics showed good agreement as indicated by the average combined CORA rating of 0.7. In the PMHS experiment, at approximately 100 ms when the deceleration reached its peak, the hands and the head translated to right lateral side, which was simulated by the GHBM model. The shoulder belt showed good retention during the peak deceleration but slipped somewhat when the HBM rebounded. Figure 4 shows a comparison of the simulation responses with the ±1 SD experimental corridors for the 3 resultant accelerations and contact forces. The combined CORA ratings were highest at the console force, followed by pelvis and leg forces, whereas resultant head acceleration was highest, followed by T12 and T1 accelerations (Table 1).

Full-scale simulations

As indicated in the Methods section, 12 impact scenarios were simulated by changing the impact location, impacting object, and impact speed. In general, the results indicated that the high-speed impacts against the pole resulted in severe injuries, higher excursions followed by low-speed pole, high-speed MDB, and low-speed MDB impacts. However, for brevity the high-speed pole scenarios (A-, B-, and C- pillars) will be given priority in further discussions because this case showed the highest kinematics and injury metrics. The other 9 cases will be summarized accordingly. The summary of the kinematics and injury metric for all 12 cases are shown in Table 2.

FE-HBM kinematics

In all 12 cases, immediately after the vehicle contacted the rigid pole, the vehicle structures near the contact point deformed locally (Figure 4). However, as more vehicle structures engaged the intruding pole, the local deformation reduced, resulting in gross rotation of the vehicle. Although local deformation followed by gross rotation was a common sequence of events
in all cases, the rotation of the vehicle with respect to the z-axis varied depending on the location of the contact point with respect to the CG of the vehicle. In addition, the crush in the pole simulations was higher compared to MDB simulations. In all cases, the HBM moved to the right relative to the vehicle coordinate system at approximately 15 ms into the event.

While the vehicle decelerated after contact with the rigid pole, the HBM continued to move to the struck side with its initial velocity. In all cases, the lap belt engaged and retained the occupant on the seat throughout the event. The shoulder belt was somewhat retained in the A-pillar impact, whereas the belt slipped out in the other 2 cases. The peak lap and shoulder belt forces were 1.5 and 2.8, 3.4 and 2.3, 2.5 and 0.5 kN for the A-, B-, and C-pillar cases, respectively. In the B- and C-pillar cases, the occupant slipped out under the shoulder belt and the upper torso traversed toward the struck side. However, throughout the event, the shoulder belt was in contact with the lower right quadrant of the rib cage. In all cases, as the impacting vehicle decelerated, the driver seatback flexed rearward and rotated about the vertical axis of the vehicle. The increased gap between the torso and seatback on the right edge of the driver seat may

Table 2. Summary of the calculated kinematics, kinetics, and injury measures for the simulated cases.

| Case         | Pole (HS) | Pole (LS) | MDB (HS) | MDB (LS) |
|--------------|-----------|-----------|----------|----------|
| Impact pillar| A B C     | A B C     | A B C    | A B C    |
| Head         | 556.6 693.5 707.3 | 287.1 354.5 409.6 | 382.4 517.6 498.2 | 151.3 249.9 239.9 |
| T6           | 236.5 377.7 343.8 | 103.9 138.4 168.8 | 147.4 221.8 216.0 | 57.5 85.8 82.3 |
| Pelvis       | 91.1 173.6 113.0 | 99.6 80.2 101.6 | 90.1 165.8 155.5 | 68.4 95.5 60.3 |

Peak excursion in the lateral direction (mm)

| Normalized injury prediction (%) |
|----------------------------------|
| Lung (L)                        | 68 100 82 | 43 16 17 | 37 51 38 | 14 23 19 |
| Lung (R)                        | 100 72 34 | 5 7 17 | 33 23 2 | 2 2 4 |
| Liver                           | 53 74 100 | 1 7 1 | 15 30 13 | 3 0 0 |
| Spleen                          | 61 94 100 | 0 0 6 | 3 34 0 | 0 0 0 |

Viscous criterion (m/s)

| Fracture (AIS) |
|----------------|
| Ribs           | 4 2 4 | 1 0 1 | 1 1 2 | 0 0 0 |
| Pelvis         | 0 2 0 | 0 0 0 | 0 0 0 | 0 0 0 |
have contributed to the ineffectiveness of the shoulder belt. In the B-pillar case, the passenger seat was crushed by the intruding pole and pushed toward the driver seat. However, in the other 2 cases the passenger seat was not significantly deformed. The low-speed pole and MDB impact cases showed higher engagement compared to the high-speed pole case with lower seatback deformations.

Figure 5 shows the displacements of the head CG, T6 vertebral body, and pelvis CG with respect to the vehicle coordinate system. The displacement was greatest in the negative y-axis direction in all cases and was observed to be the highest at the head. The displacements progressively decreased from head to pelvis. For all body regions, the predominant displacements were observed to occur in the negative y-axis direction with the exception of the A-pillar impact. In the A-pillar impact, the occupant moved in both the x- and y-directions and directly struck the passenger side seat. In the C-pillar impact, at approximately 100 ms into the event, the right upper arm (humerus) of the HBM marginally struck the inside edge of the backrest of the passenger seat but continued to move in the negative y-direction and eventually rebounded. In the B-pillar impact, the HBM missed the crushed passenger side seat, resulting in a direct impact of the head with the intruding interior components on the passenger side. The low-speed pole and MDB impact cases showed lower y-direction excursion compared to the high-speed pole case with no direct contact to the intruding structures.

Injury estimation in the FE-HBM

The injury estimation analysis did not predict injury to the kidneys and the bladder. Figure A2 (see online supplement) shows the chestband contours for the 12 cases. The peak chest deflections were observed as 56, 60, and 66 mm for the A-, B-, and C-pillar high-speed cases, respectively. In all cases, significant deformation in the left anterolateral side of the rib cage was consistently observed. The higher deformation in the A-pillar case was due to the direct impact of the torso against the passenger seat. In addition, the A-pillar case showed the highest shoulder belt retention among the 3 cases. Figure A3 (see online supplement) shows the regional plastic strain distribution in the rib cage for one of the cases (A-pillar). Ribs in the A- and C-pillar cases sustained AIS 2 injuries and the B-pillar case sustained AIS 2 injury. The timing analysis indicated that the nominal time of the elements yielded in the right and left halves of the rib cage were 97 and 122 ms, respectively.

The injury risk estimation in the right lung was observed to be the highest in the A-pillar case, followed by B- and C-pillars; this trend was consistent with the rib cage peak deformations. The left lung estimation showed the highest probability in the B-pillar case, followed by C- and A-pillar cases. In general, the highest lung strains were observed in the vicinity of the highest rib displacements and strains. The liver and spleen showed the highest risk in the C-pillar impact, followed by the B- and A-pillar impacts. Injury risk based on the gross viscous criterion also indicated a similar trend in the soft organs (Table 2). Pelvis injury estimation indicated AIS 2 injury in the B-pillar case, and no injuries were predicted in the C- and A-pillar impacts. These results strongly corresponded to the lap belt forces, which was maximum in the B-pillar case. The head contacted struck-side interior structures only in the B-pillar impact, indicating a high risk of sustaining an injury as indicated by the higher neck forces and moments (Table 2). The low-speed pole and MDB impact cases showed lower injury risks compared to the high-speed pole case, with a highest AIS score of 2 in the ribs in the high-speed MDB case.

Discussion

FE-HBM validation

The objective of this study was to analyze the influence of the far-side impact condition on the impact response of the occupant using a whole-body HBM and a whole-vehicle finite element model. To confirm that the occupant responses were relevant, the biofidelity of the HBM was evaluated, compared, and quantified using data from sled experiments. The average combined CORA rating was obtained as 0.7 with a poor rating to the leg contact force with the loading plate. The obtained average CORA rating may be sufficient for a majority of the applications using the GHBMC model in far-side pure lateral impact mode. Cumulative stiffness definitions at the hip and at the knee may have influenced the leg contact forces. Preliminary simulations without seat belt pretightening showed poor seat belt retention; however, the retention improved with the tightening. Kent et al. (2013) showed a similar phenomenon using pretensioners in PMHS experiments.

Full-scale simulations

In general, the high-speed pole impact cases showed higher excursions and injury severity compared to low-speed pole and MDB cases. As indicated in the Results, the crush in the MDB cases was significantly lower than that in the pole cases. One
possible explanation could be that in the pole cases the contact areas between the pole and the vehicle were much smaller compared to the MDB cases. The higher contact area in the MDB case promoted the engagement of more structures in the vehicle, resulting in lower crushing deformation.

**FE-HBM kinematics**

In the simulations, the impact location of the pole was found to vary both the kinematics of the vehicle and the HBM. The change in the rotational directions of the vehicle between the A-pillar and the C- and B-pillars was found to be similar to that reported by Haight and Diggles (2010). The location of the impact point with respect to the CG of the vehicle influenced the gross rotational direction of the vehicle. However, impact along the line of the CG of the vehicle is likely to result predominantly in gross translation of the vehicle in the direction of the impact vector. The HBM excursion was predominantly in the negative y-axis direction in the B- and C-pillar cases, whereas the A-pillar excursion had components in both x- and y-directions. There is a strong possibility that the shoulder belt retention influenced the differences in the HBM kinematics. In addition, the occupant kinematics is likely to be influenced by the interaction with the struck-side interior components, especially the passenger seat. As an example, in the A-pillar impact, the HBM directly struck the passenger seat, whereas in the C-pillar impact, the trajectory of the head was changed after the HBM contacted the inside edge of the passenger seat. Although not explored in the current study, the height of the console in the vehicle may dictate the kinematics of the occupant. Pelvis excursion is most likely to be restricted by higher consoles, in addition to lap belts; however, the restriction may alter torso kinematics.

**Injury estimation in the FE-HBM**

The location of the impact point was found to affect the injury risk estimations at the different anatomical structures. The HBM predicted likelihood of an injury in the head, rib cage, lungs, liver, spleen, and pelvis in the 3 high-speed pole cases with varying probabilities, with the exception of the head, where the injury was predicted only in the B-pillar impact. Comparison of these findings with previous studies suggests that the HBM predictions were in good agreement with injuries observed in the field. Fildes et al. (2007) reported that the rib cage is the most frequently injured anatomic structure in the chest, followed by the lungs, whereas the liver is the most frequently injured organ in the abdomen, followed by the spleen. The resulting simulations using the GHBMC model also predicted injuries to the ribs, lungs, liver, and spleen. These injuries to the ribs and soft organs were based on simulated head trajectories; it can be argued that the head excursions alone may not result in a head injury, but in combination with B-pillar regional intrusions may increase the probability of head injury. However, the width of the vehicle compartment may also influence this risk.

HBM chest deflections strongly corresponded to the lap/shoulder belt loading and lateral inertial loading by the torso. In addition, simulated local plastic strain distribution predicted injury risk to the lower left quadrant of the rib cage, consistent with the experimental observations reported by Kent et al. (2013). A likely explanation is that the rib cage along with the spine underwent lateral bending due to torso inertia, while the lap belt constrained the pelvis. This phenomenon loaded the lower right quadrant and the loads were transferred by the costal cartilage to the left lower quadrant, resulting in injurious strains (Figure A4, see online supplement). The injury risk predicted at the pelvis strongly corresponded to the lap belt engagement. In the A- and C-pillar cases, the HBM decelerated by contacting the passenger seat, resulting in reduced lap belt forces but loading the shoulder complex. The lap belt in the B-pillar impact measured the highest forces due to the lack of HBM contact with the passenger seat.

Although strain thresholds were selected based on experimental results available in the literature, a key limitation of this study is that the reported injury risk estimations may not accurately represent the probability of sustaining an injury in an absolute perspective. Rather, the purpose of these estimations was to compare the probability of sustaining injuries and severity of injuries between the 3 reported cases, namely, A-, B-, and C-pillar impact exposures. Notwithstanding this limitation, the risk estimation used in this study predicted injuries that are reported in literature based on accident databases. In summary, the GHBMC model showed an acceptable biofidelity under far-side impact exposure. In general, the results indicated that the high-speed impacts against the pole resulted in severe injuries and higher excursions followed by low-speed pole, high-speed MDB, and low-speed MDB impacts. The vehicle and occupant kinematics varied with different impact setups and the latter kinematics were strongly influenced by restraint effectiveness.

Increased restraint engagement increased the injury risk to the corresponding anatomic structure, whereas ineffective restraint engagement increased occupant excursion, resulting in a direct impact to the struck-side interior structures. One of the major applications using the outcome of the present study is facilitating design countermeasures for the far-side vehicle impacts. In particular, information on the occupant kinematics behavior with different impact locations would assist the safety system design process to optimize the effectiveness of the restraints in crashes with different impact locations.

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