SIDE AIRBAG OUT-OF-POSITION DEPLOYMENTS: THORACIC INJURY TO STATIONARY AND DYNAMIC OCCUPANTS

Jason J. Hallman, Narayan Yoganandan, Frank A. Pintar
Department of Neurosurgery
Medical College of Wisconsin,
Department of Biomedical Engineering
Marquette University,
and VA Medical Center
Milwaukee, WI, USA

ABSTRACT

Injury risk from out-of-position (OOP) side airbag deployment has been assessed using stationary occupant test protocols. However, stationary conditions may not always represent real world OOP scenarios. Therefore, the objective of the present study was to evaluate the effects of OOP torso side airbag deployment, comparing a stationary test protocol with dynamic conditions. A combination of parametric computational modeling and PMHS tests were utilized to examine the effects of impact velocity on C and VC injury metrics. Dynamic OOP scenarios induced metrics greater than the stationary condition as well as impact in absence of airbag. The sensitivity of VC and implications for visceral trauma are also discussed.

Keywords: AIRBAGS, OCCUPANTS, MULTI BODY, SIDE IMPACTS, THORAX

SIDE AIRBAGS WERE INTRODUCED into the vehicle fleet in the previous decade as a consequence of technical advancements to mitigate injury in side impact motor vehicle crashes. They currently exist in three primary configurations: head, torso, or head and torso (combination) airbags. Head side airbags (SAB) are most commonly present as side curtains deploying from the roof rail or header. Torso and combination SAB are seat- or door-mounted, i.e., deploying from within the seat back or the door panel lateral to the occupant. These technologies have already become standard or available equipment in 89% of 2009 model year vehicles in the U.S.; as many as 70% of models were available with torso SABs (Highway Loss Data Institute, 2008).

Out-of-position (OOP) can result if the occupant deviates from a normal posture. These scenarios may pose additional injury risk during airbag deployment. For example, trauma to pediatric and small female occupants in OOP scenarios from frontal airbag systems has been well-documented in the literature (Kirk et al., 2002; Durbin et al., 2003; Donaldson et al., 2008). When frontal airbags were mandated in the 1990's in the U.S. (FMVSS 208), injury risk was deemed insignificant (Melvin et al., 1993). Further, frontal crashworthiness regulations incorporated only median occupant anthropometries in ideal mid-seat postures. As a result, a collection of primarily anecdotal reports of airbag trauma followed the mandate (e.g., Kleinberger & Summers, 1997; Graham et al., 1998). These reports initiated an investigation of and subsequent revisions to the U.S. regulatory requirements to permit the use of safer "second generation" airbags (NHTSA, 1997). Public education efforts also emphasized correct child placement in airbag-equipped vehicles. Additionally, test protocols were developed to ascertain and limit the level of injury risk to OOP occupants (Berg et al., 1997; Kleinberger & Summers, 1997; Digges et al., 1998; Morris et al., 1998; Plank et al., 1998). These protocols approximated OOP scenarios with stationary anthropomorphic test devices (ATDs) in close proximity to airbag deployment. Although this methodology neglected occupant kinematics and inertial effects prior to airbag deployment, stationary approximation for frontal OOP was validated to full scale crash tests (Tylko & Dalmotas, 2001).

Because prior frontal airbag OOP studies documented an increased propensity for injury, experimental studies of side airbag risks have employed similar methods. Specifically, stationary surrogates were positioned in close proximity to SAB deployment to determine OOP injury risks. Computer simulation results indicated that small female occupants were not at risk of head, neck, and thorax injury from OOP torso SAB deployment (Khadilker & Pauls, 1998; Duma et al., 2003b). Experimental studies utilizing three- and six-year-old child ATDs identified multiple OOP scenarios in which injury metrics exceeded tolerance values (Pintar et al., 1999; Tylko & Dalmotas, 2000; Prasad et al., 2001; Gehre et al., 2003). Of note was the observation that position deviation of 2 cm reduced injury metrics by over 75% (Pintar et al., 1999). Studies utilizing post-mortem human specimens (PMHS) reported low risk to small female and median anthropometry occupants (Schroeder et al., 1998; Duma et al., 2003a; Duma et al., 2003b). These stationary OOP tests contributed to the rationale of current side airbag test procedures for motor vehicles sold in the US (IIHS, 2003).

Yet, effects of OOP side airbag deployment in a controlled dynamic event such as a sled test have not been delineated. Only one dynamic study was found in the literature, impacting PMHS at $\Delta V = 15$ m/s (Schroeder et al., 1998). Consequently, the present study was designed to evaluate the contribution of impact velocity to OOP-induced torso injury from SAB deployment using a computational model and PMHS tests. Response was compared to the currently accepted stationary test procedure.

METHODS

The fiftieth percentile facet occupant model was utilized in MADYMO (R6.3, TNO-MADYMO, Livonia, MI) simulations. The model consisted of head, neck, thorax, abdomen, and upper and lower extremities (Fig. 1). Each body region was modeled with rigid bodies of proportional mass. Masses were enclosed by facet surfaces to reproduce the complex anthropomorphic geometries. Skeletal joint motions, including extremity and vertebral column motions, were modeled as restrained rigid body joints.

The thorax, the region of interest to meet the objectives of the present study, consisted of four discrete deformable structures, or contours. The nodes of these elliptical contours collectively defined the circumference of the thorax facet surface (Fig. 2). Progressing caudally, thoracic contours were located at lateral levels of the fourth, sixth, eighth, and tenth rib (R4, R6, R8, and R10), where each intersects with the anterior axillary line. Deformation in response to lateral forces occurred through medial superposition of nodes. Resistance to deformation (thoracic stiffness) was provided by numerically defined nonlinear elastic and viscoelastic restraint elements. These elements resisted relative translation of the lateral nodes with respect to the vertebral column and with respect to the adjacent superior and inferior thoracic contours. Further details are presented in the Discussion.

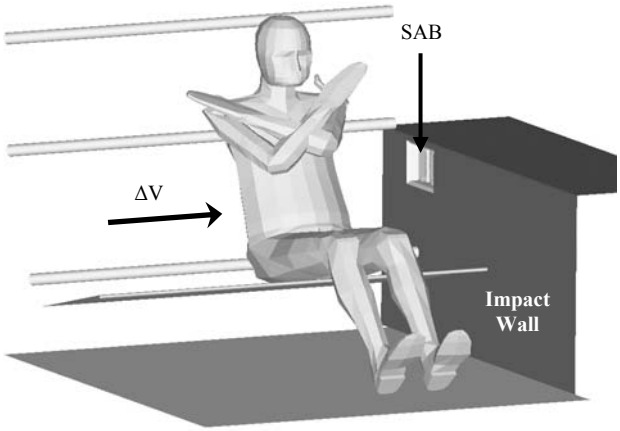

Fig. 1 – Model and sled environment

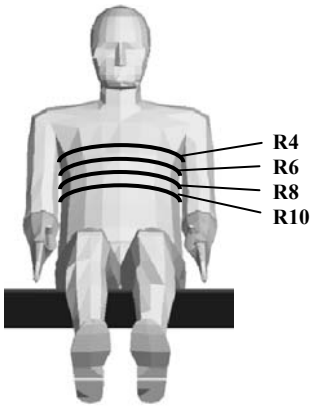

Fig. 2 – Occupant model with thorax contours at four levels

The dynamic environment simulated a Heidelberg-type sled device. The occupant model was seated on a zero-friction bench seat fixed to the platform of the sled. The occupant was positioned such that the Frankfort plane was horizontal, legs stretched parallel to the mid-saggital plane, and posture was forward upright with normal spinal curvature and alignment. For simulating nearside impact, a load wall was positioned such that the occupant left side contacted the rigid surface boundary at a predefined impact velocity. The lateral thorax was fully exposed to impact, and a door-mounted torso SAB was simulated by housing the bag in a module within the load wall (Fig. 1).

To approximate a typical SAB in current automobiles, a finite element frontal airbag was tethered by line elements to an inflated depth of 18 cm and inflated volume of 13 L. The airbag fabric consisted of two circular halves composed of 13,000 triangular membrane elements with 0.5 mm thickness. The airbag mesh strain field was referenced to the unfolded mesh throughout the deployment sequence using the Initial Metric Method (Tanavde et al., 1995). Inflation behavior was defined by a mass inflow function with isothermal expansion. The airbag required 14 ms to inflate. In a 60 L tank test, the prescribed inflator achieved 95 kPa maximum pressure and 5,000 kPa/s maximum onset rate. Thermodynamic calculations utilized a lumped-parameter approach, assuming uniform pressure and temperature throughout the airbag volume. Inflator nozzle gas flow was approximated by applying additional momentum to nodes within the nozzle outflow stream; this additional momentum was defined using an Idelchik nozzle diffusion approximation. Although recent simulations have incorporated gas dynamics to better replicate airbag-environment interactions in the early deployment phase, i.e., OOP (Marklund & Nilsson, 2002; Pyttel et al., 2007; Ruff et al., 2007), lumped-parameter models with nozzle effects have also been successfully validated to OOP experiments (Roychoudhury et al., 2000; Petit et al., 2003; Park & Hong, 2005). Surface-to-surface contacts were defined between the airbag fabric, the rigid impact wall, and the occupant facet surface skin.

Biomechanics were quantified by determining chest deformations at the four thoracic contours (Fig. 2). From the contours full chest compression (C) and the viscous criterion (VC) were calculated as shown in Equations 1 and 2. In these equations, D_0 is initial chest breadth and D(t) is chest breadth at time t. Model D(t) was calculated between most lateral left and right contour nodes.

$$C = \frac{D_0 - D(t)}{D_0} \quad (\%) \tag{1}$$

$$VC = \frac{dD(t)}{dt} \times \frac{D(t) - D_0}{D_0} \quad (\text{m/s}) \tag{2}$$

Fig. 3 – OOP thorax injury protocol for door-mounted SAB (IIHS, 2003)

According to the adult OOP test protocol specified in Section 3.3.4.5 of the *Recommended Procedures for Evaluating Occupant Injury Risk from Deploying Side Airbags* (IIHS, 2003), the occupant surrogate was positioned against the SAB module with the lateral thorax exposed to maximum inflation force (Fig. 3). Injury metrics resulting from the model positioned in this stationary scenario were compared to metrics resulting from dynamic occupant OOP conditions. In dynamic conditions, the SAB was triggered when the occupant was two centimeters from the module; preliminary simulations indicated this location to represent "worst case" deployment during impact (Hallman et al., 2008). To simulate dynamic sled impact, impact velocity (ΔV) was applied directly to the occupant toward the impact wall. Increments of 1.0 m/s from $\Delta V = 4 – 9$ m/s were chosen, and impacts without airbag were also simulated.

PMHS were screened for HIV and Hepatitis A, B, and C, and anthropomorphic data were recorded. Pretest radiographs were taken to identify various body and musculoskeletal components. This included the head, neck, chest, thorax, upper and lower extremities, and pelvis. The specimens were dressed in leotards and positioned on a lateral impact buck fixed to the platform of a deceleration sled identical to the simulations. Specimens were seated facing forwards and upright with the Frankfort plane horizontal, legs stretched parallel to the mid-sagittal plane, and normal curvature and alignment of the thoracolumbar spine maintained without pre-torso rotation. PMHS tests were conducted with and without SAB deployment. Twelve PMHS were impacted at $\Delta V = 6.4 - 7.0$ m/s; eight in a rigid wall condition and four with in-position torso SAB. In-position was defined as thorax contact at full airbag inflation. One OOP deployment was conducted at $\Delta V = 6.7$ m/s, in which the thorax contacted the SAB within the first four milliseconds following activation. Thoracic deflections were obtained from two 59-channel chest bands at the fourth and eighth rib levels. For OOP, a third chestband was placed at the tenth rib level. Chest wall velocity, used to calculate VC, was taken to be the rate of chest deflection. After each test, an autopsy was conducted to assess trauma to hard and soft tissues. Trauma scores were assessed to organ systems according to the Abbreviated Injury Scale (AIS) and the entire specimen was categorized according to the maximum AIS (MAIS). PMHS metrics from rigid wall tests, reported earlier (Pintar et al., 1997), were compared to the non-airbag simulation at $\Delta V = 7.0$ m/s. Metrics from fully-inflated SAB tests were compared to simulations at $\Delta V = 7.0$ m/s with full SAB inflation. PMHS metrics from OOP were compared to OOP simulations at both $\Delta V = 6$ and 7 m/s.

RESULTS

PMHS TESTS: Shown in Table 1 are the peak injury metrics resulting from thirteen PMHS sled impacts at $\Delta V = 6 – 7$ m/s. Rigid wall boundary condition was considered baseline response. Compared to rigid wall, full side airbag inflation reduced peak C and VC metrics. For OOP occupant, metrics were elevated compared to fully-inflated airbag contact: VC metric in OOP exceeded range

obtained from in-position occupants, and C metric was within C =3% of range upper limit. Metrics were also near upper limits obtained from rigid wall boundary condition. Simulations with ΔV = 6 – 7 m/s were selected to compare with PMHS results, also shown in Table 1. Rigid wall contact at ΔV = 7 m/s induced peak metrics within range obtained from PMHS tests. For fully-inflated airbag contact, occupant position was varied ± 2 cm to accommodate variations in experimental setup. The injury metric range obtained from this variation intersected the range obtained from PMHS experiments. OOP occupant was simulated at both ΔV = 6 and 7 m/s, yielding a metric range which encompassed OOP PMHS results at ΔV = 6.7 m/s. These OOP metrics are depicted in time domain in Figure 4, which also demonstrated the performance of the lumped-parameter airbag inflation model. The resulting simulation response ranges in these three boundary conditions fell within or intersected with PMHS response data.

Table 1. Peak injury metrics from PMHS and simulations in dynamic scenarios

Test Method	Trait	Rigid Wall		Optimum SAB		OOP	
		C range (%)	VC range (m/s)	C range (%)	VC range (m/s)	C range (%)	VC range (m/s)
PMHS	# Tests	8		4		1	
	Results	26 – 41	0.6 – 2.0	23 – 38	0.2 – 1.5	35	1.6
Model	Results	27	0.8	21 – 24	0.4 – 0.6	35 – 39	1.5 – 2.0

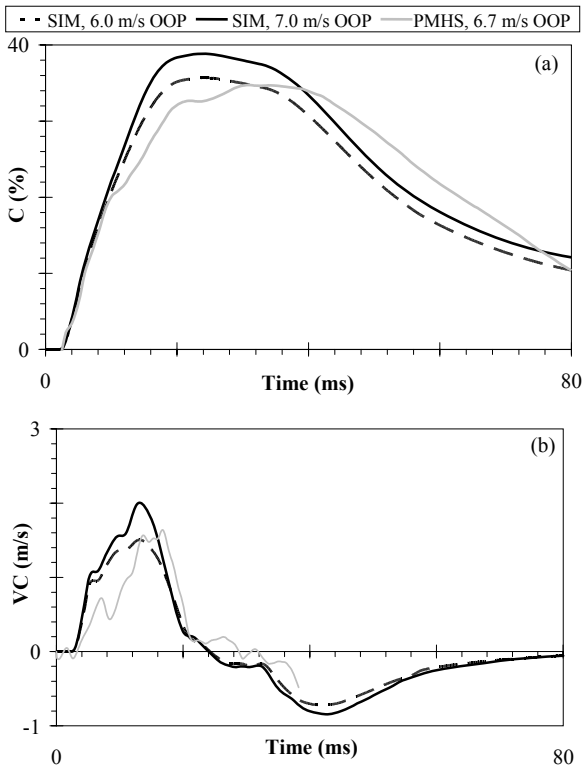

Fig. 4 – Biomechanical response of simulation and PMHS occupants in OOP at ΔV = 6 – 7 m/s as measured by (a) compression and (b) viscous criterion

STATIONARY OOP SIMULATION: The OOP performance of the airbag model was evaluated with stationary occupant simulation. Biomechanical response resulting from stationary OOP occupant is shown in Figure 5. Contour at R4 (Fig. 2) indicated the lowest metrics, followed by R6 and R10. The greatest metrics were demonstrated at R8; this contour corresponded to the locus of airbag inflation. At these contours, compression was found to attain between C = 14% (R4) and C = 21% (R8). The viscous metric attained between VC = 0.3 m/s and VC = 0.7 m/s at these respective contours.

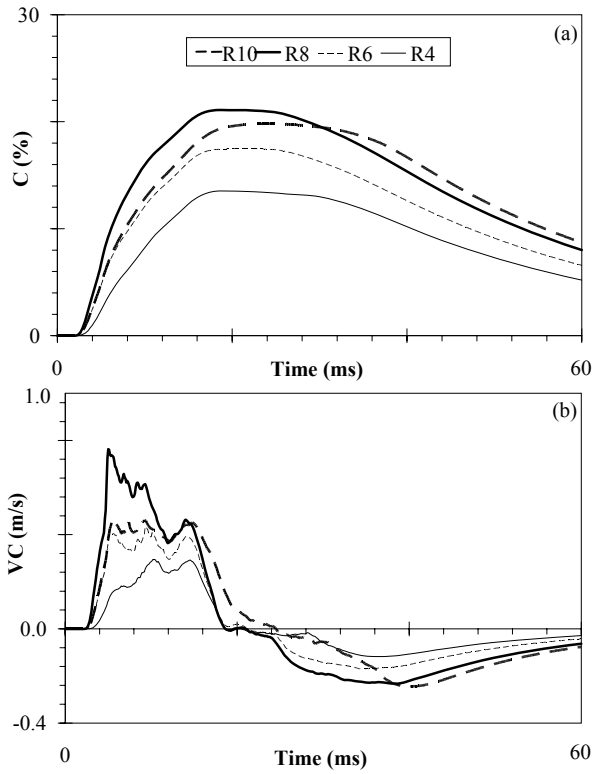

Fig. 5 – Biomechanical response in time domain from simulation of stationary OOP occupant as measured by (a) C and (b) VC at the four thoracic levels

DYNAMIC OOP SIMULATION: To examine the effect of impact velocity on OOP occupant, the computational model was subjected to ΔV = 4 – 7 m/s with late (OOP) airbag deployment. Shown in Table 2 are peak metrics obtained from OOP simulations at all impact velocities and thoracic contours. The greatest C response occurred at R8 for all velocities. Peak VC response occurred at R8 for all velocities except ΔV = 9 m/s, for which peak response occurred at R10. Peak metrics from ΔV = 6 and 7 m/s are also presented in Table 1 for comparison to PMHS results. Time traces of biomechanical response are shown in Figure 6 for stationary and dynamic OOP occupant at all ΔV; only the thoracic contour demonstrating peak metrics for each ΔV is depicted. Occupant position was equivalent at SAB activation (t = 0). During the initial three milliseconds following SAB contact, injury metrics from stationary and dynamic OOP differed by less than 5%. With the progression of impact, biomechanical response to dynamic OOP deviated from response to stationary OOP. Peak injury metrics increased with increased impact velocity. Time at which peaks occurred was not considerably affected by ΔV, decreasing marginally with increasing velocity. Peak compression occurred between 24.7 ms (ΔV = 9 m/s) and 26.9 ms (ΔV = 4 m/s) following airbag activation; peak VC metric occurred between 15.0 ms (ΔV = 9 m/s) and 16.1 ms (ΔV = 4 m/s).

Table 2. Peak metrics from OOP simulations at all contours and impact velocities

Level	Metric	Stationary	4 m/s	5 m/s	6 m/s	7 m/s	8 m/s	9 m/s
R4	C (%)	13	20	21	22	23	24	25
	VC (m/s)	0.3	0.5	0.7	0.8	0.9	0.9	1.0
R6	C (%)	18	25	27	29	32	34	36
	VC (m/s)	0.4	0.8	1.0	1.1	1.4	1.5	1.7
R8	C (%)	21	30	33	35*	39*	42	44
	VC (m/s)	0.7	1.0	1.3	1.5*	2.0*	2.2	2.4
R10	C (%)	20	30	32	35	38	41	43
	VC (m/s)	0.5	1.0	1.2	1.4	1.9	2.2	2.5

* - denotes metrics presented in Table 1.

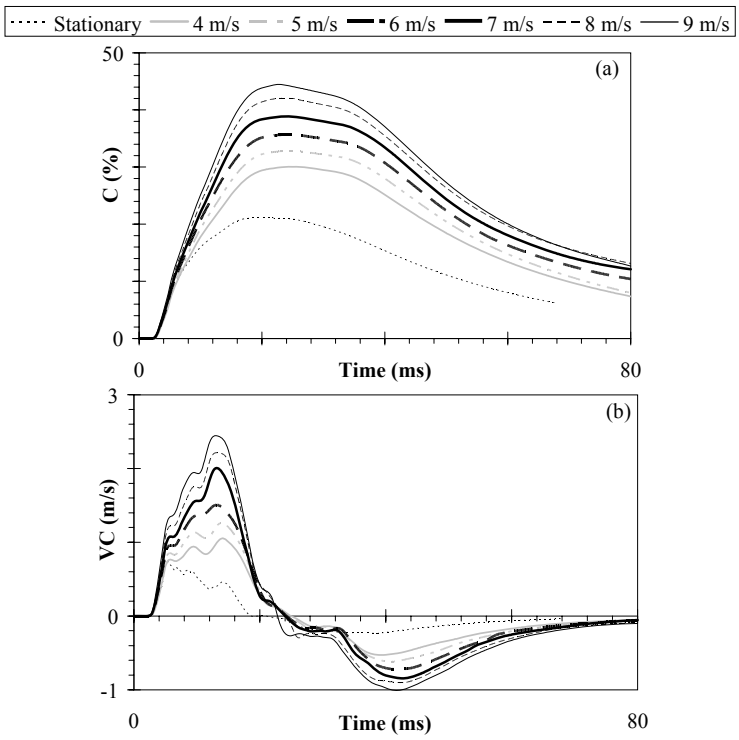

Fig. 6 – Peak thoracic biomechanical response from simulations in stationary scenario and all ΔV as measured by the (a) compression metric and (b) viscous metric

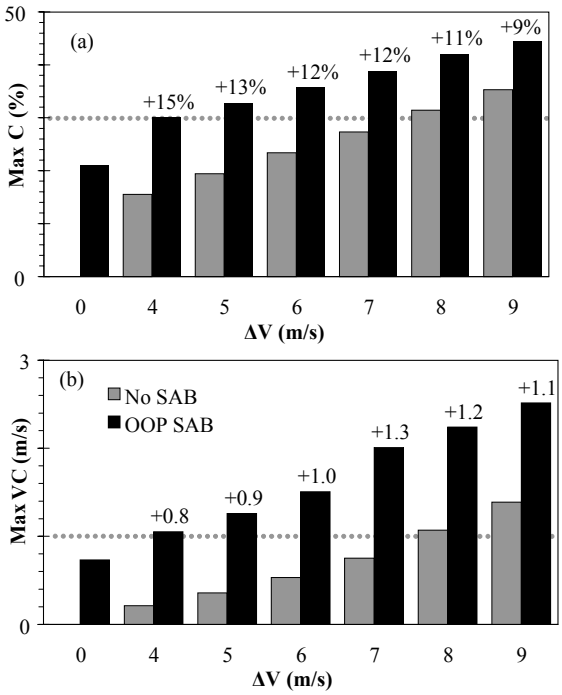

Fig. 7 – Thoracic biomechanical response to ΔV in simulation with and without SAB measured by (a) C and (b) VC

Shown in Figure 7 are peak thoracic biomechanical metrics from the computational model in all dynamic scenarios, i.e., both rigid wall and OOP. Without airbag, metrics ranged C = 16 – 35% and VC = 0.2 – 1.4 m/s. In OOP, metrics ranged C = 21 – 44% and VC = 0.7 – 2.5 m/s. Also indicated in Figure 7 are increases in biomechanical response induced by OOP at each ΔV. Dynamic

OOP induced biomechanical response greater than response without SAB at all ΔV considered. Metric increase from OOP was found to be C = 9 – 15% and VC = 0.8 – 1.3 m/s over non-SAB impact at identical ΔV. The greatest increases in C occurred at ΔV = 5 m/s or less; the greatest increases in VC occurred at ΔV = 7 m/s and greater.

DISCUSSION

The purpose of this study was to examine the effect of realistic occupant dynamics on OOP-induced injury from torso side airbags. The MADYMO environment was used in this study to facilitate a parametric analysis. The model has been utilized in other studies, and its geometric details have been described elsewhere (Happee et al., 2000; de Lange et al., 2005). This model was used in conjunction with a new side airbag model. Because a side airbag was not available with the current version of MADYMO, a frontal airbag was modified as described within the Methods.

The modified airbag in this study accommodated a reasonable approximation to torso side airbag characteristics. The maximum tank test pressure was reasonable for a door-mounted SAB (Pintar et al., 1999). Airbag aggressivity was conservative as measured by the maximum tank pressure onset rate; onset rates 300% greater than specified in this study have been reported with door-mounted SABs (Pintar et al., 1999). Lumped-parameter analysis was employed, assuming uniform pressure and temperature throughout the airbag control volume; this may inhibit the interpretation of quantitative conclusions with regard to injury metric magnitudes. Recent advances in coupled fluid-structure algorithms have demonstrated that the contribution of gas dynamics to the early stages of airbag inflation can affect deployment kinematics (Marklund & Nilsson, 2002; Pyttel et al., 2007; Ruff et al., 2007). Yet, previous studies have successfully validated lumped-parameter analysis of computational OOP occupant models (Roychoudhury et al., 2000; Petit et al., 2003; Park & Hong, 2005). Furthermore, comparisons between computational and PMHS results obtained in this study (Fig. 4) suggested that the effect of these analysis assumptions may be limited.

The simulation also included a sled apparatus with Teflon-coated seat. This apparatus was geometrically similar to that which was utilized previously in lateral impact sled tests (Pintar et al., 1997). To obtain maximum thoracic biomechanical sensitivity to accomplish study objectives, the lateral thorax was fully exposed by flexing and medially rotating the left shoulder; a similar method has been utilized in previous cadaveric sled experiments (Pintar et al., 1997; Kuppa et al., 2003). The simulated door-mounted torso SAB was also positioned within the impact wall to interact with the maximum thoracic surface area when deployed, ensuring maximum metric sensitivity.

The impact velocity of 4 – 9 m/s chosen in the present study is representative of real world crash data. For example, analysis of the National Automotive Sampling System (NASS) database revealed this ΔV range to approximate the middle 50% of side impacts in the U.S. (Zaouk et al., 2001). For comparison purposes this same ΔV range was used with and without side airbags to evaluate C and VC metrics. These biomechanical metrics were shown to correlate with thoracic injury risk in PMHS experiments (Viano, 1989; Kuppa et al., 2003). Biomechanical tolerances have been developed to quantify injury risk (Lau & Viano, 1986; Viano et al., 1989b; Cavanaugh et al., 1990; Pintar et al., 1997; EuroNCAP, 2004). Injury tolerance has been considered to be C = 30% or VC = 1.0 m/s; these magnitudes reportedly correspond to 25% risk of AIS 4+ thoracic injury (Viano, 1989; Pintar et al., 1997; EuroNCAP, 2004). These metrics have also been used to extensively validate the MADYMO model to pendulum and sled impacts of PMHS (Happee et al., 2000; de Lange et al., 2005).

Because the airbag was altered to achieve the objectives of the present study, it was deemed necessary to ensure that the revised model produced realistic results. Consequently, PMHS test data were used (Kuppa et al., 2003) and in the process, C and VC metrics with both rigid wall and side airbag boundary conditions were compared to the simulations. From this perspective, modeling and simulation processes presented in this study serve as an additional validation of the fiftieth percentile occupant model. As shown in Table 1 and Figure 4, the present modeling results agreed well with

PMHS findings, ensuring confidence in the parametric analysis and evaluation of the potential mitigating effects of torso side airbags using C and VC criteria. Although the airbag deployment characteristics were not validated in isolation, observations regarding the parametric effect of ΔV in comparison to stationary occupant are still relevant and have not been delineated in the literature.

As expected, C and VC increased with increasing ΔV in all cases: without airbag and in OOP (Figs. 6 and 7). In Figure 7, injury tolerance is also indicated by dotted line. Under the stationary condition tested, metrics did not exceed injury tolerances. During impact without side airbag, both metrics exceeded tolerance when $\Delta V \geq 8$ m/s. With airbag deployment in OOP, metrics exceeded biomechanical tolerance at $\Delta V \geq 5$ m/s; these were also greater than metrics from stationary OOP at all ΔV. Therefore between ΔV = 5 and 8 m/s, injury tolerances were exceeded only with OOP airbag deployment. This velocity range, although quantitatively specific to this airbag model only, suggests that similar circumstances may exist for other forms of torso-interacting airbags in which their deployment can exacerbate injury. Because the airbag did not induce metrics in excess of tolerances during stationary testing, the accepted stationary evaluation procedure may be insufficient to characterize OOP injury from torso-interacting side airbags. Dynamic scenarios may be more relevant to the propensity for SAB to induce injury.

An evaluation of the normalized biomechanical response indicated that VC may be a more sensitive parameter to OOP-induced injury. Shown in Figure 8 is the difference in OOP injury metrics in OOP with respect to rigid boundary condition, normalized to injury threshold. Rate of increase in the C metric decreases monotonically as a function of ΔV. Rate of increase in the VC metric is not monotonic; the change in VC metric increases up to ΔV = 7 m/s. This result implies that VC may be a more sensitive parameter to OOP injury risk. This conclusion is further reinforced by PMHS tests at ΔV = 6.7 m/s, in which VC response in OOP was elevated to a greater extent than C response when compared to fully-inflated airbag tests. Because simulations identified a local VC maximum at ΔV = 7 m/s, this may represent the worst-case scenario for OOP side airbag deployments. However, these findings should be interpreted with caution as the performance of side airbags in the field is dependent on many variables, including the airbag design, impact velocity, tank test results, and vehicle design. Performance evaluation is further complicated by the lump-parameter airbag simulation method discussed previously. With this in mind, it is relevant to note that prior studies have reported VC to be correlated to visceral injury (Viano et al., 1989a) and C to be correlated to rib fracture (Pintar et al., 1997). Therefore, because VC response in this study appeared more sensitive to OOP deployment than C response, visceral (soft tissue) trauma risk may be more sensitive to OOP torso SAB deployment than skeletal (hard tissue) trauma under the conditions examined.

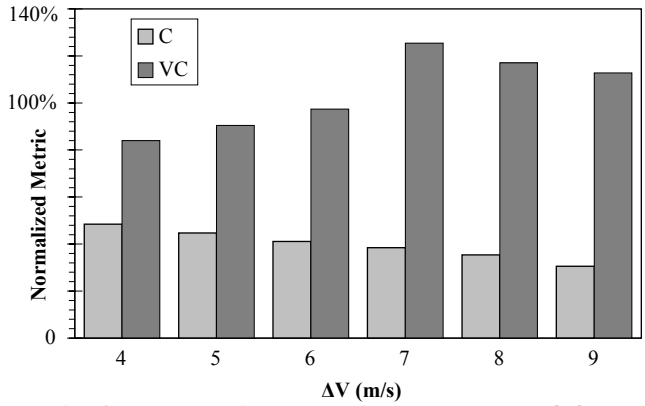

Fig. 8 – Normalized Metric Response to OOP

Analyses of these results suggest that organs in close proximity to the SAB during deployment such as spleen and kidney may be more susceptible to trauma in OOP side airbag scenarios. Recognizing that the present study is primarily a hybrid experimental-simulation approach,

field-related injury data can reinforce these laboratory-based observations. In other words, it would be necessary to examine the efficacy of airbag performance in the real world to support such conclusions. Epidemiological analyses using the U.S. National Automotive Sampling System (NASS) database have not been conclusive (Yoganandan et al., 2005). Other studies have utilized the General Estimates System (GES) and Fatality Analysis Reporting System (FARS) databases in the US and other national databases. Analyses of GES and FARS data (1997-2002) have suggested statistically insignificant effects from torso airbags for adult occupants and detrimental effects for elderly occupants (McGwin et al., 2003; Braver & Kyrychenko, 2004). Analysis of U.K. National Accident Data has also suggested possible detrimental effects from torso side airbag deployment (Morris et al., 2005). More recent GES and FARS analysis (1999-2004) suggested mortality risk reductions for occupants in SAB-equipped vehicles (McCartt & Kyrychenko, 2007). Cases studies, which first identified frontal airbag OOP risks in the previous decade, have not yet identified consistent OOP injury patterns resulting from SAB deployment (Baur et al., 2000; Dalmotas et al., 2001; Kirk & Morris, 2003; Weber et al., 2004).

These literature-based findings in conjunction with the current results underscore the need to more closely examine the potential for OOP injury in side airbag deployment cases. Although this study was limited by the availability of a rigorously validated coupled fluid-structure side airbag model, analysis revealed a broader trend in the propensity of torso SAB to induce injury in a dynamic OOP scenario. Laboratory analysis of this scenario has received only cursory treatment in the literature (Schroeder et al., 1998). Therefore, further work will include more PMHS tests in dynamic OOP scenarios, as well as real-world examinations such as matched-pair analysis with side airbag in similar model year vehicles. This type of effort has begun to emerge from the Crash Injury Research and Engineering Network (CIREN), in which spleen injuries have been identified as a possible consequence of OOP side airbag deployment.

CONCLUSIONS

A hybrid experimental-computational approach was undertaken to comparatively evaluate thoracic injury metrics between stationary and dynamic OOP scenarios. While the SAB utilized in this analysis did not exceed biomechanical tolerance in stationary OOP evaluation, tolerance was exceeded in dynamic OOP scenarios when $\Delta V = 5$ m/s or greater. Because unprotected side impacts did not exceed biomechanical tolerance unless $\Delta V = 8$ m/s or greater, response exceeded tolerance between $\Delta V = 5$ and 8 m/s only with SAB. Results further suggested that VC response, and therefore visceral trauma, may be more sensitive to side airbag injury. These dynamic OOP conditions cannot be identified through stationary testing and should be addressed in future work by continued field performance analyses and laboratory testing.

ACKNOWLEDGMENTS

This research was supported in part by NIH Grant (R01AG024443), the National Science Foundation Graduate Research Fellowship Program, and Veterans Affairs Medical Research.

REFERENCES

Baur, P., Lange, W., Messner, G., Rauscher, S., & Pieske, O. Comparison of real world side impact/rollover collisions with and without thorax airbag/head protection system: A first field experience study. *Annual Proceedings of the Association for the Advancement of Automotive Medicine,* 44, 2000. 187-201.

Berg, F. A., Schmitt, B., Epple, J., & Mattern, R. Dummy-loadings caused by an airbag in simulated out-of-position situations. *1997 International IRCOBI Conference,* Hannover, Germany. 1997. 419-431.

Braver, E. R., & Kyrychenko, S. Y. Efficacy of side air bags in reducing driver deaths in driver-side collisions. *American Journal of Epidemiology,* 159(6), 2004. 556-564.

Cavanaugh, J. M., Walilko, T. J., Malhotra, A., Zhu, Y., & King, A. I. Biomechanical response and injury tolerance of the thorax in twelve sled side impacts. *34th Stapp Car Crash Conference,* Orlando, FL. 1990. 23-38.

Dalmotas, D., German, A., & Tylko, S. The crash and field performance of side-mounted airbag systems. *17th International Technical Conference on the Enhanced Safety of Vehicles,* Amsterdam. 2001.

de Lange, R., van Rooij, L., Mooi, H., & Wismans, J. Objective biofidelity rating of a numerical human occupant model in frontal to lateral impact. *Stapp Car Crash Journal,* 49, 2005. 457-479.

Digges, K., Noureddine, A., Eskandarian, A., & Bedewi, N. E. Effect of occupant position and air bag inflation parameters on driver injury measures. *SAE 980637*. 1998.

Donaldson, W. F., Hanks, S. E., Nassr, A., Vogt, M. T., & Lee, J.Y. Cervical spine injuries associated with the incorrect use of airbags in motor vehicle collisions. *Spine,* 33(6), 2008. 631-634.

Duma, S. M., Boggess, B. M., Crandall, J. R., Hurwitz, S. R., Seki, K., & Aoki, T. Upper extremity interaction with a deploying side airbag: A characterization of elbow joint loading. *Accident; Analysis and Prevention,* 35(3), 2003a. 417-425.

Duma, S. M., Crandall, J. R., Rudd, R. W., & Kent, R. W. Small female head and neck interaction with a deploying side airbag. *Accident; Analysis and Prevention,* 35(5), 2003b. 811-816.

Durbin, D. R., Kallan, M. J., Elliott, M., Cornejo, R. A., Arbogast, K. B., & Winston, F. K. Risk of injury to restrained children from passenger airbags. *Traffic Injury Prevention,* 4(1), 2003. 58-63.

European New Car Assessment Program. *Assessment protocol and biomechanical limits.* 2004.

Gehre, C., Schaefer, R., & Schindler, V. Interaction of child restraint systems and side airbags. *2003 International IRCOBI Conference,* Lisbon, Portugal. 2003.

Graham, J. D., Goldie, S. J., Segui-Gomez, M., Thompson, K. M., Nelson, T., Glass, R., Simpson, A., & Woerner, L. G. Reducing risks to children in vehicles with passenger airbags. *Pediatrics,* 102(1), 1998. e3.

Hallman, J. J., Yoganandan, N., & Pintar, F. A. Torso side airbag out-of-position evaluation using stationary and dynamic occupants. *Biomedical Sciences Instrumentation,* 44, 2008. 123-128.

Happee, R., Ridella, S., Nayef, A., Morsink, P., de Lange, R., Bours, R., & van Hoof, J. Mathematical human body models representing a mid size male and a small female for frontal, lateral, and rear impact loading. *2000 International IRCOBI Conference,* Montpellier, France. 2000. 67-89.

Highway Loss Data Institute. *Vehicles equipped with side airbags.* 2008. Retrieved 15/12 from <http://www.iihs.org/ratings/side_airbags/side_airbags.aspx >.

Insurance Institute for Highway Safety. *Recommended procedures for evaluating occupant injury risk from deploying side airbag*s. 2003.

Khadilker, A.V., & Pauls, L.S. Assessment of injury protection performance of side airbags for out-of-position and other than 50th percentile adult male occupants. *The 16th International Technical Conference on the Enhanced Safety of Vehicles,* Windsor, Ontario, Canada. 1998. 1858-1867.

Kirk, A., Frampton, R., & Thomas, P. An evaluation of airbag benefits/disbenefits in European vehicles-a combined statistical and case study approach. *2002 International IRCOBI Conference,* Munich, Germany. 2002.

Kirk, A., & Morris, A. Side airbag deployments in the UK - initial case reviews. *18th International Technical Conference on the Enhanced Safety of Vehicles,* Nagoya, Japan. 2003.

Kleinberger, M., & Summers, L. Mechanisms of injuries for adults and children resulting from airbag interaction. *Annual Proceedings of the Association for the Advancement of Automotive Medicine,* 41, 1997. 405-420.

Kuppa, S., Eppinger, R. H., McKoy, F., Nguyen, T., Pintar, F. A., & Yoganandan, N. Development of side impact thoracic injury criteria and their application to the modified ES-2 dummy with rib extensions (ES-2re). *Stapp Car Crash Journal,* 47, 2003. 189-210.

Lau, I. V., & Viano, D. C. The viscous criterion - bases and applications of an injury severity index for soft tissues. *30th Stapp Car Crash Conference,* San Diego, CA. 1986. 123-142.

Marklund, P. O., & Nilsson, L. Simulation of airbag inflation processes using a coupled fluid structure approach. *Computational Mechanics,* 29(5), 2002. 289-297.

McCartt, A. T., & Kyrychenko, S. Y. Efficacy of side airbags in reducing driver deaths in driver-side car and SUV collisions. *Traffic Injury Prevention,* 8(2), 2007. 162-170.

McGwin, G.,Jr, Metzger, J., Porterfield, J. R., Moran, S. G., & Rue, L. W. Association between side air bags and risk of injury in motor vehicle collisions with near-side impact. *The Journal of Trauma,* 55(3), 2003. 430-434.

Melvin, J. W., Horsch, J. D., McCleary, J. D., Wideman, L. C., Jensen, J. L., & Wolanin, M. J. Assessment of airbag deployment loads with the small female hybrid III dummy. *37th Stapp Car Crash Conference,* San Antonio, TX, USA. 1993.

Morris, A., Welsh, R., Thomas, P., & Kirk, A. Head and chest injury outcomes in struck-side crashes. *2005 International IRCOBI Conference,* Prague, Czech Republic. 2005. 73-83.

Morris, C. R., Zuby, D. S., & Lund, A. K. Measuring airbag injury risk to out-of-position occupants. *The 16th International Technical Conference on the Enhanced Safety of Vehicles Proceedings,* Windsor, Ontario, Canada. 1998. 1036-1043.

National Highway Traffic Safety Administration. *Actions to reduce the adverse effects of airbags: Final regulatory evaluatio*n. Washington, D.C.: U.S. Department of Transportation. 1997.

Park, W., & Hong, S. A study on the modelling technique for the passenger out-of-position simulation. *SAE 2005-01-1298*. 2005.

Petit, P., Trosseille, X., Baudrit, P., & Gopal, M. Finite element simulation study of a frontal driver airbag deployment for out-of-position situations. *Stapp Car Crash Journal,* 47, 2003. 211-241.

Pintar, F. A., Yoganandan, N., Hines, M. H., Maltese, M. R., McFadden, J., Saul, R., Eppinger, R. H., Khaewpong, N., & Kleinberger, M. Chestband analysis of human tolerance to side impact. *41st Stapp Car Crash Conference,* Lake Buena Vista, FL. 1997. 63-74.

Pintar, F.A., Yoganandan, N., Maltese, M.R., Samaha, R.R., & Eppinger, R.H. Three-year-old out-of-position side airbag studies. *43rd Stapp Car Crash Conference,* San Diego, CA. 1999. 25-38.

Plank, G. R., Kleinberger, M., & Eppinger, R. H. Analytical investigation of driver thoracic response to out-of-position airbag deployment. *42nd Stapp Car Crash Conference,* Tempe, AZ. 1998.

Prasad, A. K., Samaha, R. R., & Louden, A. E. Evaluation of injury risk from side impact air bags. *17th International Technical Conference on the Enhanced Safety of Vehicles,* Amsterdam. 2001.

Pyttel, T., Floss, A., Thibaud, C., & Goertz, C. Realistic simulation models for airbags and humans-new possibilities and limits of FE simulation. *IJ Crash,* 12(5), 2007. 481-492.

Roychoudhury, R., Sun, D., Hamid, M., & Hanson, C. 5th percentile driver out of position computer simulation. *SAE 2000-01-1006*. 2000.

Ruff, C., Jost, T., & Eichberger, A. Simulation of an airbag deployment in out-of-position situations. *Vehicle System Dynamics,* 45(10), 2007. 953-967.

Schroeder, G., Kallieris, D., Tschaeschke, U., Scheunert, D., Schuetz, J., & Zobel, R. Are side airbags dangerous in certain seating positions? *1998 International IRCOBI Conference,* Goeteborg, Sweden. 1998. 447-483.

Tanavde, A. S., Khandelwal, H., Lasry, D., Ni, X., Haug, E., Schlosser, J., & Balakrishnan, P. Airbag modeling using initial metric methodology. *SAE 950875*. 1995.

Tylko, S., & Dalmotas, D. Assessment of injury risk to children from side airbags. *44th Stapp Car Crash Conference,* Atlanta, GA. 2000.

Tylko, S., & Dalmotas, D. Static out-of-position test methodologies: Identifying a realistic worst case for small stature female drivers. *17th International Technical Conference on the Enhanced Safety of Vehicles,* Amsterdam, Netherlands. 2001.

Viano, D. C. Biomechanical responses and injuries in blunt lateral impact. *33rd Stapp Car Crash Conference,* Washington, D.C. 1989. 113-142.

Viano, D. C., Lau, I. V., Andrzejak, D. V., & Asbury, C. Biomechanics of injury in lateral impacts. *Accident; Analysis and Prevention,* 21(6), 1989a. 535-551.

Viano, D. C., Lau, I. V., Asbury, C., King, A. I., & Begeman, P. Biomechanics of the human chest, abdomen, and pelvis in lateral impact. *Accident; Analysis and Prevention,* 21(6),1989b. 553-574.

Weber, P. R., Cassatta, S. J., Sochor, M. R., Faust, D. P., Fakry, S., Watts, D., Mock, C., & Wang, S. C. Lateral air bag performance in CIREN field studies. *SAE 2004-01-0331*. 2004.

Yoganandan, N., Pintar, F. A., & Gennarelli, T. A. Evaluation of side impact injuries in vehicles equipped with side airbags. *2005 International IRCOBI Conference,* Prague, Czech Republic. 2005. 97-108.

Zaouk, A. K., Eigen, A. M., & Digges, K. H. Occupant injury patterns in side crashes. *SAE 2001-01-0723*. 2001.