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Thoracic injury metrics with side airbag: Stationary and dynamic occupants

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Abstract

Objective—Injury risk from side airbag deployment has been assessed using stationary out-of-position occupant test protocols. However, stationary conditions may not always represent real world environments. Therefore, the objective of the present study was to evaluate the effects of torso side airbag deployment on close-proximity occupants, comparing a stationary test protocol with dynamic sled conditions.

Methods—Chest compression and viscous metrics were quantified from sled tests utilizing post-mortem human specimens and computational simulations with three boundary conditions: rigid wall, ideal airbag interaction, and close-proximity airbag deployment. PMHS metrics were quantified from chestband contour reconstructions. The parametric effect of ΔV on close-proximity occupant was examined with the computational model.

Results—PMHS injuries suggested close-proximity occupants may sustain visceral trauma, which was not observed in occupants subjected to rigid wall or ideal airbag boundary conditions. Peak injury metrics were also elevated with close-proximity occupant relative to other boundary conditions. The computational model indicated decreasing influence of airbag on compression metrics with increasing ΔV . Airbag influence on viscous metric was greatest with close-proximity occupant at $\Delta V = 7.0$ m/s, at which the response magnitude was greater than linear summation of metrics resulting from rigid impact and stationary close-proximity interaction.

Conclusions—These results suggest that stationary close-proximity occupants may not represent the only scenario of side airbag deployment harmful to the thoracoabdominal region. The sensitivity of the viscous metric and implications for visceral trauma are also discussed.

Keywords

SIDE AIRBAGS; SIDE IMPACT; BIOMECHANICS; CHEST INJURY; MODELING; SLED TESTING

INTRODUCTION

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 (IIHS, 2009).

If the occupant deviates from an ideal posture, airbag deployment may pose additional injury risk during an impact event. For example, trauma to pediatric and small female occupants from frontal airbag systems has been well-documented in the literature (Durbin et al., 2003; Kirk et al., 2002). Yet, when frontal airbags were mandated during the 1990's in the U.S., frontal crashworthiness regulations (FMVSS 208) incorporated only median occupant anthropometries in ideal mid-seat postures. The collection of primarily anecdotal reports of airbag trauma following the mandate initiated an investigation of and subsequent revisions to the U.S. regulatory requirements, permitting the use of safer "second generation" airbags (Kleinberger and Summers, 1997; NHTSA, 1997). Public education efforts emphasized correct child placement in airbag-equipped vehicles. Test protocols were also developed to ascertain and limit the level of injury risk to at-risk occupants (Digges et al., 1998; Kleinberger and Summers, 1997; Lau et al., 1993; Morris et al., 1998; Plank et al., 1998). These protocols approximated injurious scenarios with stationary anthropomorphic test devices in close proximity to airbag deployment. Although this methodology neglected occupant kinematics and inertial effects prior to airbag deployment, injury metrics from stationary scenarios were matched to full scale crash tests and sled impacts (Horsch et al., 1990; Tylko and Dalmotas, 2001).

Because prior frontal airbag studies documented an increased propensity for injury to occupants in close proximity, experimental studies of side airbag risks have employed similar methods. Specifically, stationary surrogates were positioned in close proximity to SAB deployment to determine injury risks from inadvertent deployment. Computer simulation results indicated that small female occupants were not at risk of head, neck, and thorax injury from torso SAB deployment (Duma et al., 2003b; Khadilkar and Pauls, 1998). Experimental studies utilizing three- and six-year-old child test dummies identified multiple test scenarios in which injury metrics exceeded tolerance values (Pintar et al., 1999; Prasad et al., 2001; Tylko and Dalmotas, 2000). Of note was the observation that position deviation of two centimeters 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 (Duma et al., 2003a; Duma et al., 2003b; Schroeder et al., 1998). These stationary tests established the rationale of current side airbag test procedures for motor vehicles sold in the U.S. (IIHS, 2003).

Effects of SABs in a controlled dynamic event such as a sled test have not been delineated. Only one study was found in the literature, impacting PMHS at $\Delta V = 15$ m/s (Schroeder et al., 1998). Because thoracoabdominal trauma has been suggested as a detrimental effect of torso side airbags in real world crashes (Morris et al., 2005), it was hypothesized that crash ΔV may influence airbag-thorax interactions with close-proximity occupants. Consequently, the present study was designed to evaluate the contribution of impact velocity to thoracic injury metrics with close-proximity SAB deployment using a computational model and PMHS tests. Response was compared to a stationary test procedure currently in use.

METHODS

Thoracic biomechanical response was quantified in sled impact with and without torso-protecting side airbag. Lateral injury metrics of chest compression (C) and viscous criterion (VC) were quantified using PMHS in laboratory experiments and a computational model. Boundary conditions were categorized as rigid wall, ideal side airbag deployment, and close-

proximity deployment. Ideal SAB was defined as thorax contact at full airbag inflation. Close-proximity deployment was defined by preliminary computational simulations (Hallman et al., 2008); thorax contact occurred within four milliseconds following airbag activation.

Laboratory Experiments

This study incorporated select PMHS tests reported in part by Kuppa et al. (2003). These tests utilized rigid and ideal SAB boundary conditions at $\Delta V = 6.7$ m/s (24 km/h) and select results were reported for injury criteria development. To characterize the airbag-thorax interactions for this study, these test data were reexamined to identify effects of boundary condition on metrics and on thoracoabdominal injuries. New tests were conducted with close-proximity PMHS at similar ΔV and test setup.

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 Heidelberg-type deceleration sled (Pintar et al., 1997). 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. Tests were conducted on a deceleration sled at $\Delta V = 6.7 \pm 0.3$ m/s. After each test, necropsy was conducted to assess trauma to hard and soft tissues. Trauma scores were assigned according to the Abbreviated Injury Scale (AIS 1990), and the thoracoabdominal maximum AIS (MAIS) was determined (Copes et al., 1990).

Thoracic injury metrics were derived from two 40-channel or 59-channel chestbands at the fourth and eighth rib levels. For close-proximity SAB, a third chestband was placed at the tenth rib level. To measure chest deflection using chestband signals, a mid-thorax line was defined between the gages overlying the adjacent spinous process and mid-sternum. Deflections were quantified as the distance between the lateral chest band margin and this mid-thorax line along a vector normal to the mid-thorax line. Chest deformation rate, used to calculate VC, was obtained by numerical differentiation of chest deflection with respect to time and application of a CFC180 filter in accordance with SAE J211. Metrics obtained from this technique were correlated with injury risk in lateral impact (Kuppa et al., 2003).

Simulations

The 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 mass proportional to a fiftieth percentile male occupant. 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 (de Lange et al., 2005).

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). 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. The biomechanical response of the thorax was validated to a variety of boundary conditions (de Lange et al., 2005; Happee et al., 2000). Free pendulum tests utilized a 23.4 kg mass with ΔV between 3.3 and 9.9 m/s. Drop tests

utilized rigid contact at ΔV between 4.3 and 6.3 m/s. Sled impacts utilized rigid contact at $\Delta V = 6.7$ and 9.1 m/s. Model biomechanics consisted of force-deflection, force-time, and acceleration-time response and was compared to cadaveric tests. Model biofidelity was also evaluated using ISO TR9790 (de Lange et al., 2005), in which environmental reaction forces were validated. In all, model response was demonstrated to be a predictor of cadaveric response for impact velocities ranging $\Delta V = 3.3 - 9.7$ m/s. For this study, lateral chest deflections were quantified by a method similar to chest band methods for PMHS deflections: The mid-thorax line was defined between nodes overlying the spinal column centerline and the mid-sternum. Deflection was quantified as the normal distance between the lateral node and the mid-thorax line.

The simulated impact replicated a Heidelberg-type sled device identical to that utilized in laboratory experiments for this study (Fig. 1). 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-sagittal 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 torso SAB was simulated by housing the bag in a module within the load wall.

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 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., close-proximity (Marklund and Nilsson, 2002; Pyttel et al., 2007; Ruff et al., 2007), lumped-parameter models with nozzle effects have also been successfully validated to experiments (Park and Hong, 2005; Petit et al., 2003; Roychoudhury et al., 2000). Surface-to-surface contacts were defined between the airbag fabric, the rigid impact wall, and the occupant facet surface skin.

The injury risk to the thorax was evaluated with the SAB in stationary and dynamic scenarios. For the stationary test, the occupant surrogate was positioned against the SAB module with the lateral thorax exposed to maximum inflation force according to the adult test protocol specified in Section 3.3.4.5 of the *Recommended Procedures for Evaluating Occupant Injury Risk from Deploying Side Airbags* (IIHS, 2003). Injury metrics resulting from the model positioned in this stationary scenario were compared to metrics resulting from dynamic occupant conditions. To simulate dynamic sled impact, impact velocity (ΔV) was applied to the occupant toward the impact wall. Increments of 1.0 m/s from $\Delta V = 4.0 - 9.0$ m/s were chosen; each boundary condition was simulated at each ΔV . Injury metrics were quantified at each contour by methods similar to chest band analysis. PMHS metrics from rigid boundary tests were compared to the non-airbag simulations at $\Delta V = 6.0$ and 7.0 m/s. Metrics from ideal SAB tests were compared to ideal SAB simulations. PMHS metrics from close-proximity SAB were compared to close-proximity simulations at both $\Delta V = 6.0$ and 7.0 m/s.

RESULTS

A total of fourteen PMHS tests were analyzed with and without side airbag deployment. Fifteen simulations were analyzed, representing scenarios comparable to laboratory experiments as well as incremental ΔV with rigid wall and close-proximity boundary conditions.

Laboratory Results

Shown in Table 1 are subject details and resulting injuries for PMHS sled tests. With rigid boundary condition, injuries varied from no injury ($n = 3$) to AIS 4 skeletal trauma (left flail chest). With ideal airbag contact, injuries consisted of rib fractures (MAIS 2); two specimens did not sustain injury. Two specimens were subjected to close-proximity contact with airbag; both sustained visceral trauma to the spleen or kidney (left aspect).

Biomechanical injury metrics were calculated from PMHS tests using chestband contour reconstructions (Fig. 3). Deflection of the lateral border was quantified with respect to the mid-thorax line. Three lateral border locations were examined to identify peak deflection: one quarter (25%) band circumference and $\pm 5\%$ (20% and 30%) band circumference; these locations are indicated in Figure 3 by open markers. These deflections were normalized to full chest breadth to yield percent compression (Pintar et al., 1997). Peak metrics are shown in Table 2 scaled according to subject mass (Eppinger et al., 1984); other experimental measurements were reported in Kuppia et al. (2003). Although sample sizes precluded statistical comparison, specific trends were observed. Considering rigid wall boundary condition as baseline response, ideal airbag boundary condition appeared to reduce mean peak C and VC metrics. For close-proximity occupant, mean response of both injury metrics was increased compared to both ideal airbag and rigid boundary conditions.

Results from PMHS tests were compared to computational simulation results for validation purposes. Shown in Figure 3(b) are chest deformation contours computed from simulations corresponding to undeformed state, maximum deformation, and two intermediate instants. Because of predefined model constraints, i.e., nodal superposition, the most lateral contour point (25% circumference distance) always represented the location of peak metrics; metrics at this location were compared to aggregate PMHS response (Fig. 4). Individual time responses from PMHS were scaled according to subject mass (Eppinger et al., 1984) prior to aggregation. Obtained at the eighth rib level, the traces demonstrated that simulated responses from computation model were similar to PMHS response in magnitude and in morphology. Simulated response was generally within ± 1 SD from PMHS data during the loading phase of impact. Similar relationships were noted in examining the fourth rib level but are not reproduced here. Shown in Table 2 are computational peak metrics from boundary conditions similar to PMHS tests. Simulations with $\Delta V = 6.0 - 7.0$ m/s were selected as range endpoints to compare with PMHS results. Simulated rigid contact induced peak metrics within one SD of mean response obtained from PMHS tests. For ideal airbag boundary condition, the compression range obtained from varying ΔV in simulations encompassed the average response obtained from PMHS experiments; simulation VC response was within one SD. Close-proximity occupants, also compared at both $\Delta V = 6$ and 7 m/s, yielded a computational metric range which was encompassed by PMHS results. Comparing boundary conditions, simulation metrics trended similarly to PMHS tests: ideal airbag induced metrics below rigid boundary condition, and close-proximity induced metrics exceeding rigid boundary condition.

Parametric Simulation Results

The parametric effect of ΔV on thoracic metric response with airbag was evaluated with the computational model. Simulated biomechanical response resulting from stationary occupant is shown in Figure 5. Contour at R4 (see Fig. 2) indicated the lowest peak metrics, followed

by R6 and R10. The greatest metrics were demonstrated at R8; this contour corresponded to the locus of airbag inflation. In stationary conditions, compression was found to attain between $C = 0.13$ (R4) and $C = 0.19$ (R8). The viscous metric attained between $VC = 0.3$ m/s and $VC = 0.7$ m/s at these respective contours.

The occupant model was also evaluated with close-proximity airbag and $\Delta V = 4.0 - 9.0$ m/s. Shown in Table 3 are peak metrics obtained from simulations at all impact velocities and thoracic contours. In close proximity, the greatest metric response occurred at R8 for all injury metrics and velocities. Time traces of biomechanical response are shown in Figure 6 for stationary and dynamic close-proximity occupant at all ΔV ; only the thoracic contour demonstrating peak metric for each ΔV is depicted. Occupant position was equivalent at airbag activation ($t = 0$). During the initial three milliseconds following airbag contact, injury metrics from stationary and dynamic occupants differed by less than five percent. With the progression of impact, biomechanical response of the dynamic occupant deviated from response of the stationary occupant. Peak injury metrics increased with increased impact velocity. Time at which metric peaks occurred was also affected by ΔV . During stationary deployment, peak compression occurred at $t = 17.1$ milliseconds after airbag activation; when ΔV was applied peak compression occurred within 19 – 20 milliseconds. Peak VC was attained at $t = 5.6$ milliseconds after airbag activation with stationary occupant; with applied ΔV peak occurred within 9.2 – 12.6 milliseconds after airbag activation.

Shown in Figure 7 are peak biomechanical metrics from the computational model with rigid wall and close-proximity boundary conditions. Without airbag, compression metrics ranged $C = 0.15$ ($\Delta V = 4.0$ m/s) to $C = 0.35$ ($\Delta V = 9.0$ m/s); VC ranged $VC = 0.2$ to $VC = 1.3$ m/s. With close-proximity boundary condition and dynamic impact, metrics ranged $C = 0.21 - 0.44$ and $VC = 0.8 - 1.8$ m/s. Also indicated in Figure 7 are increases in biomechanical response induced by close-proximity boundary condition, compared to rigid contact at each ΔV . Close-proximity airbag induced biomechanical response greater than rigid boundary condition at all ΔV considered. Metric increase resulting from airbag was found to be $C = 0.01 - 0.10$ and $VC = 0.5 - 0.8$ m/s over rigid contact at identical ΔV . The greatest increase in C occurred at $\Delta V = 4.0$ m/s; the greatest increase in VC occurred at $\Delta V = 7.0$ m/s.

DISCUSSION

The purpose of this study was to examine the effect of dynamic occupant on thoracic injury metrics with close-proximity torso side airbag interaction. Sled tests utilizing PMHS were analyzed to examine thoracoabdominal injuries and evaluate lateral thoracic injury metrics with rigid and airbag boundary conditions. A computational model was utilized to evaluate the parametric effect of ΔV in an orientation utilized in stationary test protocols for injury risk assessment of torso airbags. It should be noted that this side airbag orientation was examined in a previous modeling exercise in which a worst-case deployment was determined (Hallman et al., 2008). Limited PMHS sample sizes were evaluated, but observed PMHS trends were in agreement with computational simulations. Further, observations regarding the parametric effect of ΔV with close-proximity occupants are relevant and have not been reported in the literature.

Sled tests conducted with PMHS may suggest side airbag influences on injury and local response. With rigid boundary contact, only skeletal injuries were observed during post-test necropsy. With ideal side airbag interaction, injuries were less severe with fewer rib fractures observed (mean = 1.3 vs. 4.8). In close-proximity to airbag deployment, both subjects sustained visceral (splenic or renal) trauma not present in other boundary conditions. However, findings should be interpreted with caution as the performance of side airbags in the field depends on many variables: airbag design, impact velocity, tank test results, vehicle side structure, etc.

Interpretation is further complicated by biological diversity within these subject sets despite mass-scaling techniques (Eppinger et al., 1984). It is worthwhile to note that at this laboratory, this ΔV has heretofore not resulted in splenic trauma; kidney trauma was produced only with abdominal offset (Kuppa et al., 2003; Pintar et al., 1997; Yoganandan et al., 1997; Yoganandan et al., 2008). These results may suggest that close-proximity airbag deployment influences visceral trauma risk. Occupants subjected to close-proximity airbag boundary condition also demonstrated generally increased injury metrics with respect to ideal side airbag interaction and with respect to rigid boundary condition. Rigid and ideal side airbag boundary conditions were included in injury metric development identifying C and VC to be good predictors of AIS 4+ injury (Kuppa et al., 2003); therefore these biomechanical metrics were considered correlated with injury risk.

The MADYMO model was used in this study to facilitate a parametric analysis. The model has been utilized in other studies, and further geometric details have been described (de Lange et al., 2005; Happee et al., 2000). In these studies, the model was demonstrated to be a predictor of cadaveric response with multiple boundary conditions. These included sled impacts, drop impacts, and free pendulum impacts with impact velocity ranging $\Delta V = 3.3 - 9.7$ m/s. Cadaveric test results used for this study were stratified into rigid and side airbag boundary conditions, and compression and viscous metrics resulting from these experiments were compared to those resulting from computational simulations. From this perspective, modeling and simulation processes serve as an additional validation of the fiftieth percentile occupant model. As shown in Table 2 and Figure 4, the present simulation results agreed well with PMHS findings. Although model contour morphology differs in complexity compared to PMHS response (Fig. 3), resulting biomechanical metrics agree well with calculated PMHS corridors (Fig. 4). Yet, it should be noted that close-proximity airbag results represent only two PMHS tests. This limitation should be considered when interpreting the results of this study.

This occupant model was used in conjunction with a new side airbag model. Because a side airbag was not available with the current version of MADYMO, an existing generic airbag model 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 realistic for a torso-interacting 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 side airbags (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 specific airbag designs. 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 and Nilsson, 2002; Pyttel et al., 2007; Ruff et al., 2007). Yet, recent studies have successfully validated lumped-parameter analysis of computational airbag models with close-proximity occupants (Park and Hong, 2005; Petit et al., 2003; Roychoudhury et al., 2000). Furthermore, comparisons between computational and PMHS results obtained in this study (Table 2) suggested that the effect of this analysis assumption may be limited.

The impact velocity of $\Delta V = 4.0 - 9.0$ m/s chosen in the parametric 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 airbag to evaluate compression and viscous metrics resulting from close-proximity airbag boundary condition. As previously stated, these biomechanical metrics were shown to correlate with thoracic injury risk in PMHS experiments (Kuppa et al., 2003; Viano, 1989); biomechanical tolerances of C = 0.30 or VC = 1.0 m/s have been defined to quantify 25% risk

of AIS 4+ thoracic injury (Cavanaugh et al., 1990; EuroNCAP, 2004; Lau and Viano, 1986; Pintar et al., 1997; Viano et al., 1989b).

As expected, C and VC increased with increasing ΔV in all boundary conditions: without airbag and with close airbag proximity (Figs. 6 and 7). In Figure 7, aforementioned injury tolerances are also indicated by dotted line (EuroNCAP, 2004; Pintar et al., 1997; Viano, 1989). Under the stationary condition tested, metrics did not exceed injury tolerances. This scenario, i.e., out-of-position testing, represents the likelihood an airbag may induce injury during an inadvertent “no crash” deployment. During dynamic impact without side airbag, both metrics exceeded or nearly achieved tolerance when $\Delta V > 7$ m/s. During impact with close-proximity airbag deployment, metrics achieved or exceeded biomechanical tolerance at approximately $\Delta V \geq 5$ m/s; these metrics were also greater than metrics from stationary close-proximity at all ΔV . Therefore, injury tolerances were exceeded between $\Delta V = 5.0$ and ~ 8.0 m/s only with airbag deployment. This velocity range represents a scenario in which side airbag deployment may be detrimental rather than beneficial to the thoracoabdominal region. Although quantitatively specific to this simulation only, similar circumstances may exist for forms of torso-interacting airbags in real-world crashes. Because the airbag did not induce metrics in excess of tolerances under the stationary out-of-position test, dynamic test scenarios may also be relevant to the propensity for side airbags to induce thoracoabdominal injury.

Additionally, an evaluation of the normalized biomechanical response indicated that VC may be a more sensitive parameter to close-proximity airbag deployment. Shown in Figure 8 is the difference in metrics between close-proximity injury metrics and rigid boundary condition after normalizing to stationary biomechanical response. Examining compression metrics, influence of close-proximity airbag deployment decreased monotonically as a function of ΔV : linear fit to compression response demonstrated $R^2 > 0.997$. In contrast, the influence of close airbag proximity on the viscous metric was not linear; the effect of the airbag boundary condition increased up to $\Delta V = 7.0$ m/s. At this ΔV , normalized metric increase was greater than 1.0, indicating that the airbag increased the metric by a magnitude greater than that which resulted from the stationary out-of-position test. In other words, the collective effect on VC response of close-proximity airbag and dynamic rigid impact exceeded the linear summation of their independent effects. Although further cadaveric testing must confirm this behavior to be independent of the model, this observation may be explained by a theoretical interpretation of VC. Mathematically, peak VC is proportional to the peak energy storage rate of the thorax (Wang, 1989). While the total energy imparted by the close-proximity airbag deployment and the dynamic rigid impact (at specified ΔV) may not have increased beyond linear summation, their combined energy was imparted without an increase in duration. From another perspective, the airbag imparted energy mostly prior to substantial rigid wall interaction and effectively “preloaded” the thorax. Without wall interaction (stationary deployment), this event represented the highest rate of energy storage, as shown by Figure 6 in which all C and VC curves were within 5% during initial airbag interaction regardless of ΔV . With increased ΔV , kinematic energy was imparted to a “preloaded” thorax, requiring greater energy storage rates. The maximum airbag effect at $\Delta V = 7.0$ m/s may have resulted from an optimized counteraction between near-constant preload energy, i.e., identical airbag deployment, and increasing kinematic energy. As ΔV increased, the airbag energy decreased in proportion to total impact energy; at ΔV beyond those investigated here, severity of impact may render airbag effects irrelevant. Again, further cadaveric testing must confirm this behavior.

Because simulations identified a local VC maximum at $\Delta V = 7.0$ m/s, this may represent another possible scenario for out-of-position testing. However, performance variability in individual airbags and dependency on vehicle and occupant characteristics preclude generalization of this scenario to specific side airbags in the current automotive fleet. Performance evaluation is further complicated by the lump-parameter airbag simulation

method discussed previously. With this in mind, these findings may demonstrate the merit of further thoracoabdominal injury risk assessments to close-proximity occupants in realistic dynamic crash scenarios. 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 than C response to close-proximity torso side airbag deployment, visceral (soft tissue) trauma risk may be more sensitive than skeletal (hard tissue) trauma to airbag deployment under the conditions examined. Injuries observed in close-proximity PMHS (Table 1) may also suggest this relationship, but sample size limits a robust conclusion.

These results may suggest that visceral organs in close proximity to the SAB during deployment such as spleen and kidney may be more susceptible to trauma in close-proximity side airbag scenarios. Recognizing that the present study is primarily a combined 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 (Braver and Kyrychenko, 2004; McGwin et al., 2003). Analysis of U.K. National Accident Data has also suggested possible detrimental effects from torso side airbag deployment (Morris et al., 2005). Yet, more recent GES and FARS analysis (1999–2004) suggested mortality risk reductions for occupants in SAB-equipped vehicles (McCartt and Kyrychenko, 2007). Cases studies, which first identified frontal airbag risks in the previous decade, have not yet identified consistent injury patterns which may have resulted from SAB deployment (Baur et al., 2000; Dalmotas et al., 2001; Hallman et al., 2009; Kirk and 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 injury in real-world side airbag deployment cases. Although this study was limited by the availability of a rigorously validated coupled fluid-structure side airbag model, analyses suggest that broader trends may exist in which torso SAB deployment could theoretically induce injury in actual crashes. However, injury risk may also be present for occupant orientations not examined in this study. For example, stationary methods have been utilized to examine injury risk to supine occupants (Duma et al., 2003b) as well as rearward-facing pediatric occupants (Pintar et al., 1999). In light of results presented here, future work should examine these occupant orientations with coexistent lateral ΔV . Laboratory analysis of dynamic scenarios have received only cursory treatment in the literature (Schroeder et al., 1998). Therefore, further work will include more PMHS tests with dynamic close-proximity airbag 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 torso side airbag deployment (Hallman et al., 2009).

CONCLUSIONS

A combined experimental-computational approach was undertaken to comparatively evaluate lateral thoracic injury metrics between stationary and dynamic scenarios of close-proximity airbag deployment. The side airbag utilized in this analysis did not exceed biomechanical tolerance in stationary out-of-position testing; therefore injury risk was low with inadvertent deployment. During dynamic close-proximity interaction, airbag exhibited monotonically

decreasing influence on the compression metric. Airbag effects on VC exhibited local maximum at $\Delta V = 7$ m/s, at which peak VC exceeded linear summation of rigid impact and close-proximity airbag effects. These results suggested that VC response, and therefore visceral trauma, may be more sensitive to close-proximity torso airbag. This study may be limited by PMHS sample sizes, robust model validation, and use of a singular side airbag scenario. Yet, dynamic conditions identified in this study may represent another possibly injurious out-of-position scenario that cannot be immediately recognized through stationary testing. Future work should address this scenario by continued field performance analyses and laboratory testing.

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Fig. 1.
Model and sled environment

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

Fig. 3. Chest contours reconstructed from (a) chestband measurements and (b) computational model. Contours are shown undeformed, at instant of maximum deflection, and at two intermediate times. Open circle indicates deflection computation points.

Fig. 4.

Aggregate injury metric response from rigid wall boundary condition at the eighth rib level from PMHS ($\Delta V = 6.7 \pm 0.3$ m/s) and computational occupant models ($\Delta V = 6.0$ and 7.0 m/s): (a) chest compression, (b) deflection velocity, and (c) viscous metric

Fig. 5. Biomechanical response in time domain from simulation of stationary close-proximity occupant as measured by the (a) compression and (b) viscous metric at the four thoracic levels

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

Fig. 7.

Simulation thoracic peak biomechanical response to ΔV with and without SAB measured by the (a) compression metric and (b) viscous metric. Numerical values indicate metric increase induced by close-proximity boundary condition relative to rigid contact.

Fig. 8.
Normalized metric response to close-proximity airbag, i.e., rigid metric subtracted from close-proximity metric

Table 1

Cadaveric injuries from side impact with and without side airbag

Test #	Age (yrs)	Weight (kg)	Gender	Test Condition	Skeletal MAIS	Visceral MAIS	Description
1	27	73	M	rigid wall	0	0	No injury
2	55	76	M	rigid wall	3	0	7 rib fx (Left aspect), plural puncture
3	67	74	F	rigid wall	0	0	No injury
4	86	67	M	rigid wall	3	0	9 rib fx (Left aspect)
5	45	63	F	rigid wall	0	0	No injury
6	56	64	F	rigid wall	4	0	11 rib fx (Left aspect), flail chest
7	73	50	F	rigid wall	3	0	4 rib fx (Left aspect)
8	51	61	M	rigid wall	3	0	7 rib fx (Left aspect)
9	61	65	F	ideal SAB	0	0	No injury
10	60	100	M	ideal SAB	0	0	No injury
11	61	72	M	ideal SAB	2	0	3 rib fx (Left aspect)
12	70	99	M	ideal SAB	2	0	2 rib fx (Right aspect)
13	81	58	F	close-proximity	4	3	18 rib fx (Left aspect), flail chest, 2 rib fx (Right aspect), 3 splenic lacs.
14	64	87	M	close-proximity	0	2	kidney lac.

Table 2

Peak injury metrics from PMHS and simulations in dynamic scenario

Test Condition	C	VC (m/s)
rigid wall	0.26	1.3
rigid wall	0.24	1.4
rigid wall	0.21	0.8
rigid wall	0.23	0.8
rigid wall	0.22	1.0
rigid wall	0.27	1.2
rigid wall	0.21	2.0
rigid wall	0.21	0.6
Model ($\Delta V = 6 - 7$ m/s)	0.23 – 0.27	0.5 – 0.7
ideal SAB	0.15	0.2
ideal SAB	0.23	1.5
ideal SAB	0.28	1.2
ideal SAB	0.16	0.9
Model	0.19 – 0.25	0.3 – 0.6
close-proximity	0.23	0.9
close-proximity	0.29	3.1
Model	0.30 – 0.32	1.2 – 1.5

Table 3

Peak metrics from simulations at all contours and impact velocities

ΔV (m/s)	B.C.	Metric	R10	R8	R6	R4
Stationary	Close-proximity	C	0.18	0.19	0.16	0.13
		VC (m/s)	0.4	0.7	0.4	0.3
4	Rigid	C	0.11	0.12	0.14	0.15
		VC	0.2	0.2	0.2	0.2
	Close-proximity	C	0.24	0.25	0.22	0.19
		VC	0.8	0.8	0.6	0.6
5	Rigid	C	0.13	0.15	0.17	0.19
		VC	0.2	0.3	0.3	0.3
	Close-proximity	C	0.27	0.27	0.23	0.20
		VC	0.9	1.0	0.7	0.6
6	Rigid	C	0.16	0.18	0.21	0.23*
		VC	0.4	0.4	0.4	0.5*
	Close-proximity	C	0.29	0.30*	0.25	0.22
		VC	1.2	1.2*	0.9	0.7
7	Rigid	C	0.19	0.21	0.24	0.27*
		VC	0.5	0.5	0.6	0.7*
	Close-proximity	C	0.31	0.32*	0.27	0.22
		VC	1.5	1.5*	1.1	0.8
8	Rigid	C	0.21	0.24	0.28	0.31
		VC	0.6	0.7	0.9	1.0
	Close-proximity	C	0.32	0.34	0.29	0.23
		VC	1.6	1.6	1.2	0.8
9	Rigid	C	0.24	0.27	0.31	0.34
		VC	0.8	0.9	1.1	1.3
	Close-proximity	C	0.34	0.35	0.30	0.24

ΔV (m/s)	B.C.	Metric	R10	R8	R6	R4
		VC	1.7	1.8	1.3	1.0

* Denotes values appearing in Table 2.