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Side Oblique Injury Criterion

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ABSTRACT

Previous analysis of CIREN and NASS data indicate the prevalence of anterior lateral loading to the occupant in side impact crashes. Numerous studies have examined the post mortem human subject (PMHS) response to pure lateral loading; however, the biomechanics of anterior lateral loads to the thorax are not well understood. The goal of this study is to develop a side oblique injury criterion and biofidelity requirements for modern side impact anthropomorphic test devices (ATD).

The surrogates were seated on a side impact buck with angled thorax and abdomen plates. Lateral acceleration to the buck was applied using either a deceleration-rebound sled or acceleration servo-sled with a 6.7 m/s deltaV. Chestbands placed at the axilla, mid-sternum, and lower sternum/abdomen were used to calculate the magnitude and direction of chest displacement. The direction of maximum displacement was quantified as an angle relative to a vector between the spine and sternum to yield a measure of the obliquity of applied force. PMHS Displacement data were mass scaled to 50% male subject and compared to ES2-re, Thor NT, and WorldSID.

The average maximum deflection for the PMHS oblique tests was 84.0 mm (38.0% higher than pure lateral) and an increase in the average number of rib fractures (9.3 vs. 7.0) and MAIS (3.5 vs. 2.0). The average maximum chest deflection response angle for the oblique tests was 73.1 degrees. For the ATD's the ES-RE had an average maximum deflection of 48.5 mm and response angle of 90.6 degrees, WorldSID was 83.3 mm and 87.8 degrees, and Thor NT was 60.0 mm and 91.5 degrees. Results indicate the potential for more severe injuries with anterior oblique loads and the varied response of ATD's to this input.

INTRODUCTION

A recent field study reported that conditions such as an obliquely directed load vector may be more detrimental to occupant safety than pure lateral loading (Pintar et al. 2007). An analysis of 49 cases using the Crash Injury Research and Engineering Network, CIREN, database, revealed that oblique impact has characteristic injury patterns, i.e., primarily unilateral trauma to the occupant (Pintar et al. 2007). Almost all laboratory-driven side impact injury assessments using intact PMHS have been done under the pure lateral mode, and hence, injury mechanisms and injury metrics are primarily applicable for this mode (Viano et al. 1989; Viano et al. 1989; Cavanaugh et al. 1990; Cavanaugh et al. 1994; Pintar et al. 1997; Yoganandan et al. 2007). Similar studies are lacking for oblique impact. Therefore, the objective of the present study was to determine the differences in chest deflections and injuries between the oblique and pure lateral impact vector and to compare the PMHS and ATD response to oblique loads.

METHODS

Unembalmed PMHS were procured, and medical records were evaluated and screened for HIV, and Hepatitis A, B, and C. Anthropomorphic data and pretest x-rays were obtained according established procedures (Pintar et al. 1997). Specimens were dressed in tight-fitting leotards, and a mask covered the head/face. Prepared subjects were placed on a Teflon-coated bench seat, 1.3 meter in length, fixed to the platform of a deceleration sled or acceleration servo-sled, configured with an impacting wall. A four-plate configuration, i.e., upper plate contacting the mid-thorax, middle plate the abdomen, lower plate the pelvis, and extremity plate the lower extremities, was used in the wall design. To simulate an oblique side impact, the abdominal and thoracic plates of the wall were angled. Pressurization was done according established protocols (Pintar et al. 1997; Yoganandan et al. 2004).

For the PMHS, three chestbands were fixed at the level of the axilla (upper), xyphoid process (middle), and tenth rib (lower) to measure deformation-time (contours) histories during impact. Chestbands were placed at similar anatomic positions on the ATDs. The chestband provided time-history signals of each strain gage at 12.5 kHz. Using the software provided by The National Highway Traffic and Safety Administration (NHTSA), chest deformation contours were computed at every millisecond. The computation assumed no change in the circumference of the band at any time interval. The contours were calculated by setting the two reference points on the gauges closest to midsagittal plane of the spine. Deflections were computed by finding the maximum change in length between a gauge on the struck side of the specimen and a point located at a fixed distance from the spine along a vector between the spine and sternum. The fixed distance was set to one-half of the anteroposterior length on the non-deformed chest contour. The angle of the maximum displacement vector to the spine-sternum vector was calculated to further quantify the oblique response of the surrogate (figure 1).

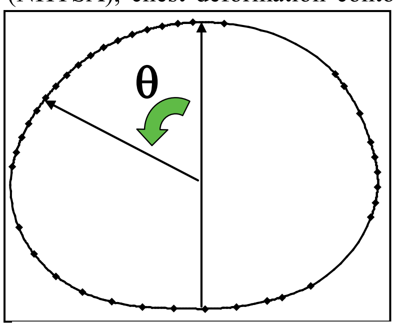

Figure 1: Angle of maximum deflection

PMHS chest deflections were obtained by mass scaling techniques (Eppinger 1976). Following the test, specimens were palpated, a clinical-type examination for stability was performed by the clinical personnel, x-rays were obtained, and a detailed autopsy was conducted. Resulting trauma was graded based on the Abbreviated Injury Scale (AIS 1990).

RESULTS

Anthropometric data were such that the average age was 55 years, stature was 173 cm, and total body mass was 59 kg. The mean acceleration was 13 g (standard deviation 0.1 g). The corresponding mean change in velocity was 24 km/h. Mean peak deflections and mean response angle at the upper, middle and lower levels of the chest for the PMHS, ES2-RE, Thor NT, and WorldSID are shown in table 1. Note that only one chestband was used for the WorldSID tests and was placed at the mid-thorax level. Average response angle was 73.1 degrees for the PMHS, 90.6 for the ES2-re, 91.6 for Thor NT, and 87.8 for WorldSID.

Table 1: Peak displacements and response angle

	Upper		Middle		Lower	
	Displacement	**Angle**	**Displacement**	**Angle**	**Displacement**	**Angle**
	mm	Degrees	mm	Degrees	mm	Degrees
PMHS	87.8	74.9	85.2	71.1	78.9	73.4
ES2-RE	34.5	98.3	52.2	89.5	58.7	84.2
Thor NT	49.4	96.3	64.5	89.4	66.0	89.0
WorldSID	n/a	n/a	83.3	87.8	n/a	n/a

PMHS oblique data are compared to flat wall tests conducted at the same velocity in figure 2. The overall average maximum displacement for the oblique tests was 84.0 mm and 60.9 mm for the flat wall tests.

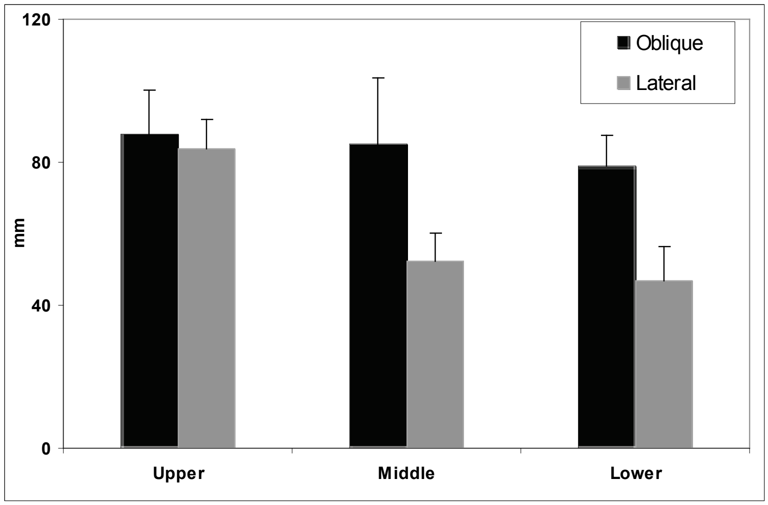

Figure 2: Peak average chest deflection

Table 2 summarizes bony and soft tissue-related injuries identified following the test. Average MAIS and rib fractures were 3.5 and 9.3 respectively. Rib fractures were predominately unilateral with soft tissue injuries occurring in one of the specimens.

Table 2: PMHS injury summary

Test	Injuries	MAIS
PMHS 1	Fractured ribs: 3,4,5,6,7,9,10	3
PMHS 2	Fractured ribs: 2,3,4,5,6,7,8,9 (rib 5 free floating); Diaphram laceration; Lung contusion	4
PMHS 3	Fractured ribs: 1,4,5,8; Diplaced left ulna fracture	3
PMHS 4	Fractured ribs: 1,2,3,4,5,7,8 (left); 5 (right); Left distal humeral fracture	4

CONCLUSIONS

The average maximum displacement of the PMHS in oblique tests were higher at all three levels of the thorax compared to pure lateral tests. The pattern of maximum displacement was also relatively uniform across the thorax in the oblique tests, while the pure lateral tests demonstrated less deflection at the middle and lower regions. Injuries in the oblique tests were in general more severe and included a soft tissue injury not seen in any of the pure lateral tests. It is well known that the contents of the human thoracic ribcage and abdomen are complex, multifunctional, three-dimensional, and from biomechanical and material property perspectives, heterogeneous. Conceptually, an oblique impact, at the same severity and to the same level of the chest, engages the same internal organ differently, compared to the pure lateral vector. For example, at the upper thoracic region, the pure lateral vector directly loads regions dorsal to the subclavian artery while an oblique vector applies impact forces to ventral arterial regions engaging the common carotid artery and brachio-cephalic vein. The former vector introduces postero-anterior load transfer to these tissues, in contrast to antero-posterior load transfer by the oblique vector. The ribcage is loaded with direct compression at its most lateral region by the pure lateral loading vector. This is in contrast to the angulated compression at the antero-lateral region by the oblique vector. The anterior regions of the thoracic vertebral body sustains lateral shear in the pure loading case, whereas, it resists antero-lateral shear in the oblique case. The spine is weaker in antero-posterior than lateral shear because of the posterior complex. At an inferior level, while the aorta is protected by the stomach in the pure lateral loading vector, in the oblique vector case, the major vessel is protected by the relatively smaller left lobe of the liver and its articulations (Yoganandan et al. 2001). Similar regional load transfer mechanisms are apparent as the impact vector traverses caudally. Purely anatomical considerations with respect to the impact vector in addition to functional and constitutive differences are likely to play in the mechanisms of load transfer, tissue-specific injury, and biomechanics.

The peak displacements for the ATD's occurred primarily about the line perpendicular to the spine sternum vector indicating a pure lateral thorax response. The PMHS maximum deflection, however, was along a more antero-lateral vector than the dummies. To further evaluate the biofidelity of the ATD's, the deflection was computed along a PMHS equivalent vector (the chestband deflection at the averaged response angle from the PMHS tests). The resulting displacements were up to 23.2% less for the ES2-re, 34.0% less for the Thor NT, and 33.6% less for the WorldSID. Therefore a hypothetical sensor placed on a dummy corresponding to the anatomical location of the PMHS equivalent vector would record deflections less than the pure lateral instrumentation for this loading scenario. The increased severity in injury in anterior oblique loads and the relative insensitivity of the current family of ATD's to this load highlight the need for the development of a side injury criterion.

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REFERENCES

AIS 1990. The Abbreviated Injury Scale. Arlington Heights, IL, American Association for Automotive Medicine.

Cavanaugh, J. M., T. J. Walilko, A. Malhotra, Y. Zhu and A. I. King 1990. Biomechanical response and injury tolerance of the thorax in twelve sled side impacts. Stapp Car Crash Conf, Orlando, FL.

Cavanaugh, J. M., Y. Zhu, Y. Huang and A. I. King 1994. Injury and response of the thorax in side impact cadaveric tests. Biomechanics of Impact Injury and Injury Tolerances of the Thorax-Shoulder Complex. S. Bachaitis. Warrendale, PA, SAE, Inc.**:** 949-972.

Dakin, G. J., R. A. Arbelaez, J. M. Nolan, D. S. Zuby and A. K. Lund 2003. Insurance Institute for Highway Safety side impact crashworthiness evaluation program: Impact configuration and rationale. Proceedings of the 18th International Technical Conference on the Enhanced Safety of Vehicles Washington, DC.

Eppinger, R. H. 1976. Prediction of thoracic injury using measurable experimental parameters. Sixth Int'l Conf on Experimental Safety Vehicles, Washington, DC.

Kuppa, S. and R. H. Eppinger 1998. Development of an improved thoracic injury criterion. Stapp Car Crash Conf, Washington, DC.

Maltese, M. R., R. H. Eppinger, H. H. Rhule, B. R. Donnelly, F. A. Pintar and N. Yoganandan 2002. "Response corridors of human surrogates in lateral impacts." Stapp Car Crash J **46**: 321-51.

Pintar, F. A., N. Yoganandan, A. Sances and R. Eppinger 1996. Instrumentation of Human Surrogates for Side Impact. Stapp Car Crash Conference, Albuquerque, NM.

Pintar, F. A., N. Yoganandan, M. H. Hines, M. R. Maltese, J. McFadden, R. Saul, R. Eppinger, N. Khaewpong and M. Kleinberger 1997. Chestband analysis of human tolerance to side impact. Stapp Car Crash Conf, Lake Buena Vista, FL.

Pintar, F. A., D. J. Maiman and N. Yoganandan 2007. Occupant dynamics and injuries in narrow-object side impact. Experimental Safety of Vehicles, Lyon, France.

Rhule, H. H., M. R. Maltese, B. R. Donnelly, R. H. Eppinger, J. K. Brunner and J. H. t. Bolte 2002. "Development of a new biofidelity ranking system for anthropomorphic test devices." Stapp Car Crash J **46**: 477-512.

SAE (1995). Instrumentation for impact test-Part 1-electonic Instrumentation-SAE J211/1. Society of Automotive Engineers, Warrendale, PA.

Viano, D. C., I. V. Lau, D. V. Andrzejak and C. Asbury 1989. "Biomechanics of injury in lateral impacts." Accid Anal Prev **21**(6): 535-51.

Viano, D. C., I. V. Lau, C. Asbury, A. I. King and P. Begeman 1989. "Biomechanics of the human chest, abdomen, and pelvis in lateral impact." Accid Anal Prev **21**(6): 553-74.

Yoganandan, N., F. A. Pintar, D. Skrade, W. Chmiel, J. M. Reinartz and A. Sances, Jr 1993. Thoracic biomechanics with air bag restraint. Stapp Car Crash Conf, San Antonio, TX.

Yoganandan, N., R. M. Morgan, R. H. Eppinger, F. A. Pintar, D. A. Skrade and A. Sances, Jr. 1995. "Thoracic deformation and velocity analysis in frontal impact." J Biomech Eng **117**(1): 48-52.

Yoganandan, N., R. M. Morgan, R. H. Eppinger, F. A. Pintar, A. Sances, Jr. and A. Williams 1996. "Mechanisms of thoracic injury in frontal impact." J Biomech Eng **118**(4): 595-7.

Yoganandan, N., F. A. Pintar and M. R. Maltese 2001. "Biomechanics of abdominal injuries." Crit Rev Biomed Eng **29**(2): 173-246.

Yoganandan, N. and F. A. Pintar 2005. "Deflection, acceleration, and force corridors for small females in side impacts." Traffic Inj Prev **6**(4): 379-86.

Yoganandan, N., F. A. Pintar, B. D. Stemper, T. A. Gennarelli and J. A. Weigelt 2007. "Biomechanics of side impact: injury criteria, aging occupants, and airbag technology." J Biomech **40**(2): 227-43.