1

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Biomechanics of Oblique Lateral Impacts

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ABSTRACT

A majority of laboratory-driven side impact injury assessments are made using post mortem human subjects (PMHS) under the pure 90-degree lateral mode. Because real-world injuries occur under pure and oblique modes, this study was designed to evaluate injury metrics and injury mechanisms with PMHS under the latter vector. Anthropometrical data were obtained and functional x-rays were taken. Specimens were seated on a sled and lateral impact acceleration was applied such that the load vector was at an angle of 20- or 30-degrees. The study determined forces in the thorax, abdomen and pelvis; deflection profiles at the levels of the axilla and mid-sternum representing the thorax, and at the tenth rib level representing the abdomen; and accelerations at the upper and lower dorsal spine and pelvis. These response data are valuable in oblique lateral impact assessments.

INTRODUCTION

The United States side impact crashworthiness tests include impacting a stationary vehicle on the driver side with a 1,360 kg moving deformable barrier. The barrier impacts the vehicle at an angle of 27 degrees. The impact velocity of the barrier for the FMVSS 214 compliance tests is 54 km/h, and for the consumer-aiding NCAP test it is 62 km/h (CFR, 2000). Depending on the mass of the vehicle, the resulting change in velocity can range from 24 to 32 km/h. Similar energies are incorporated in Standards from other countries and other evaluations although differences exist in terms of the impacting vector and mass. Conditions such as intrusion and an obliquely directed impact vector may be more detrimental to occupant safety (Pintar et al., 2007). CIREN analysis has revealed that oblique impact has characteristic injury patterns (Pintar et al., 2007). As almost all laboratory-driven side impact injury assessments using intact PMHS have been done under the pure lateral mode, the objective of this study was to determine biomechanical metrics and chest injuries associated with this impact vector.

METHODS

Unembalmed PMHS were procured, and medical records were evaluated and screened for HIV, and Hepatitis A, B, and C. Prepared subjects were placed on a Teflon-coated bench seat, 1.3 meter in length, fixed to the platform of a deceleration sled, configured with an impacting load wall. Four plates (upper plate for measuring contact forces with the mid-thorax, middle plate for the abdomen, lower plate for the pelvis, and extremity plate for the lower extremities) were used. To simulate an oblique side impact, the abdominal

and thoracic plates of the load wall were angled 20 or 30 degrees. The test matrix included experimentation with two specimens under each oblique load vector. An accelerometer on the sled was used to obtain the change in velocity. A pyramid-shaped nine-accelerometer package was fixed to the head (Yoganandan et al., 2006). In addition, tri-axial accelerometers were fixed to the upper and lower thoracic spinous processes, T1 and T12 levels, and sacrum. Eleven load cells, two each in the thorax and abdomen, four in the pelvis, and three in the extremity, were used. Three chestbands were fixed at the level of the axilla (upper), xyphoid process (middle), and tenth rib (lower) to measure deformation-time (contours) histories. The chestband provided time-history signals of each strain gage at every 80 microseconds. Using the software provided by 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 closest gauges aligned to the 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 gauges most closely aligned with the spine and sternum. The fixed distance was set to one-half of the anteroposterior length on the non-deformed chest contour. To record accelerations of the struck-side ribcage, uniaxial accelerometers were fixed to the middle side of ribs four and eight, and the sternum. All biomechanical data were gathered according to specifications (SAE 1995). Data were mass-scaled to 75 kg (CFR, 2000; Maltese et al., 2002; Yoganandan et al., 2004). Trauma was graded based on the Abbreviated Injury Scale (AIS, 1990).

RESULTS

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 acceleration was 24 km/h. The mean peak thoracic, abdominal, and pelvic forces were 4231, 4546, and 5100 N, for the 30-degree tests, and 5326, 4103, and 6729 N, for the 20-degree tests. Mean peak compressive deflections at the upper, middle and lower levels of the chest for the 30-degree tests were 96.2, 78.4, and 74.2 mm. For the 20-degree tests, these magnitudes were 75.1, 89.9, and 73.0 mm. Mean peak upper and lower thoracic spine resultant accelerations for the 30-degree and 20-degree tests were 33.3 and 46.7 g, and 67.4 and 52.0 g.

Variations in response characteristics were apparent between specimens, primarily due to biological variability, inherent to PMHS experimentation (Maltese et al., 2002). The obliqueness of the impacting vector along with individual specimen anthropometry may be responsible for these observations. In both vectors, the abdomen and thoracic plates were angled to introduce the obliqueness. However, forces in these two regions did not reach their peaks prior to forces in the pelvis region. Variations in seated height of the specimen along with changes in the distribution of the body mass and subcutaneous fat acting as a padding material, especially in the thoracic and extremity regions, might be the contributing factors. It should be noted that body mass distributions were not determined during the testing protocol and any regional alterations between specimens were not accounted for in the scaling process. In order to accommodate these factors, it is necessary to determine masses of different body regions, lower extremity, chest, etc., and suitably modify the scaling process. This is considered as an extension of the present methodology.

Contents of the human thoracic ribcage and abdomen are complex, multifunctional, three-dimensional, and from biomechanical and material property perspectives, heterogeneous. 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 at 30-degree applies impact forces to ventral arterial regions engaging the common carotid artery and brachiocephalic 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 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 a force angled towards the right pedicle in the oblique case. 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., 2000). 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 responsible for the mechanisms of load transfer, tissue injury, and biomechanics.

The United States NCAP has an oblique force direction, 27-degree crab angle of the moving deformable barrier with respect to the stationary vehicle, and this is in contrast to the European test wherein the impact vector is perpendicular, i.e., the movable deformable barrier is at 90-degree to the stationary vehicle. The 90-degree vector induces pure lateral loading on the struck side with deformations of the ribcage initiating from the region of peak skeletal curvature. In contrast, an obliquely oriented vector induces antero-lateral compression of the ribcage on the struck side, and the impact force is thus transferred via a combined shear and compression mechanism at the initiating region. Frontal impact-induced chest injuries with belt-only versus combined airbag and belt loadings have used this type of concept for describing load transfer to skeletal structures and soft tissues and delineating injury mechanisms (Yoganandan et al., 1993; Yoganandan et al., 1996). The added shear component in the oblique side impact vector places demand on soft tissue structures housed within the ribcage. The hoop tension resulting as a consequence of the compressive deformation on the antero-lateral region superimposed with the tangential component is the primary difference in the internal load-sharing mechanism between the two modes of impact. These factors may explain the more aggressive nature of the oblique than the pure lateral vector; a finding recently observed in cases examined by CIREN; narrow object and oblique impacts imparting more severe injuries than pure side impacts (Pintar et al., 2007). This study reported that oblique impacts produced more unilateral fractures along with ipsilateral soft tissue trauma, a finding consistent with the present results.

CONCLUSIONS

Sled tests using the oblique vector results in primarily unilateral injuries to the chest, similar to field data reported in another publication by the authors of the present study. A conceptual discussion is presented comparing possible differences in the biomechanics of the human chest between pure lateral and oblique load vectors. These results are of value in evaluating side impact dummy biofidelity.

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DISCUSSION

PAPER: **Biomechanics of Oblique Lateral Impacts**

PRESENTER: ***Frank Pintar, Medical College of Wisconsin***

QUESTION: *Shashi Kuppa, NHTSA*
Did you see increased soft tissue injury in your oblique impacts compared to that that of the lateral impacts or were there not enough tests to make that distinction?

ANSWER: There probably weren't enough tests. But in the one I showed you where there was a considerable amount of soft tissue, we had never seen those kinds—In the 30+ tests that we've done in lateral, we had never seen that extensive types of injuries for this, which was a 15 mile-an-hour delta V. So I qualitatively say that: Yes, you probably have more soft tissue injuries. You probably get more organ injuries.

Q: What were the injuries in those vehicle crash tests that you had?

A: A lot of lung contusion. You get some spleen.

Q: No, in that particular test that you showed us.

A: There were all unilateral rib fractures. We had actually a lung contusion in that cadaver test. You're talking about the cadaver test?

Q: Yes.

A: We had a lung contusion in that cadaver test. We had pelvic fractures. We had unilateral rib fractures.

Q: One of the questions I had was your testing method that has two plates meet at one point. That's not what you would find in a crash test, in a vehicle crash test, per se and I'm wondering: Does that have any effect because what you're essentially doing is confining the chest into that narrow oblique area? And that would have impacted, you know—Essentially not allowing your chest to move around like it may be able to do in an actual crash condition.

A: I'm not sure I understand, but, like you're saying it gets squeezed down in the corner?

Q: Yes.

A: We were concerned about that, as well. But first of all, I think that that is actually very realistic to what these pole tests—If you actually look at the pole test cadaver, it gets squeezed between the door and the seatback and that's where a lot of the deformation comes from so you do get that squeezing. But in the sled test, you don't actually get it to—it doesn't actually slide into the corner.

Q: No, but you've done these contours in chest bands where you are already aligning so you really don't know that very well. But my point is that the seat actually does deform, to some extent.

A: The seat deforms, correct.

Q: Whereas you don't—Here you have two rigid plates.

A: Well—yes, so it is a different condition. Yes, but at some point, you have to translate that into a sled environment and this is our first attempt at it.

Q: It's a good attempt.

A: I mean I think what we're assuming in the sled environment is basically that a lot of the load comes from the door coming in as opposed to just the boundary condition of the shoulder being on the seat. And my first estimation is that that seems to be fairly good.

Q: And the last: The kink that you saw in the vehicle crash test, was that from the seatbelt?

A: That is actually—I mean the chest actually got squeezed. The chest band actually kinked.

Q: Why?

A: Because the chest was actually squeezed. It changed volume. And as you know, the chest band doesn't—the chest band analysis doesn't change volume so the chest band actually came off of it. The chest band actually came off the chest and kinked.

Q: And you had a 32-gage band?

A: What's that?

Q: 32-gage band?

A: No, that was a 59.

Q: Thank you.

Q: *Bruce Donnelly, VRTC NHTSA*
Shashi just stole my question, but let me ask a little further. Do you think, if you compare the accelerometer data from the sled test to the car crash, we could get an idea of the kinematics of the subject and whether or not it actually does pack into that acute angle?

A: Yes. That's exactly what we're doing now. So we're going back to the vehicle tests. We've run a few more of the cadaver tests. We were concerned, also, with the relative timing of when the chest hits—when the chest hits the door versus the pelvis. And so, that's why we went to the vehicle test so that we could, using the accelerometers, know when those events actually happen. So that's what we're doing now.

Q: Maybe get a feel for whether that thorax is actually—

A: Was actually squeezing—

Q: Moving away the way a crash test—

A: Yes. So that's what we're doing now and we will go back and have to probably modify the sled test setup so that the timing is similar.

Q: *Richard Kent, UVA*
That's good work, Frank. A traditional definition of oblique impact would involve changing the velocity vector and I think you haven't done that here, right? The velocity vector is the same. It's a lateral vector. It's just sort of the normal to the surface is now—

A: Yes.

Q: Oblique. So, I'm just wondering why that instead of just changing the velocity vector because it does create a very complex loading environment like we've been talking about with wedging and stuff—Is that because you think that that's a better representation of what's going on due to penetration or just what exactly was the thinking behind this specific definition of oblique?

A: Two things: One is because I think it is probably closer to what I think is going on, at least in these pole tests. But, we also wanted to be able to compare it to the previous 90° tests. So we wanted to try to get it as close to those 90° tests so that we could make that comparison. If you change it too much, then it's hard to make those comparisons, as well.

Q: So how do you control, for example, where the first contact is, which would be highly dependent on the cadaver?

A: That's exactly what we're struggling with now because you can move that wall in and out relative to the other one, and that's why we went to these real vehicle tests to see what the actual timing is in a vehicle. So after we have all those vehicle tests analyzed, then we can go back and alter the sled configurations so that it's more realistic.