IMPROVED SIDE IMPACT PROTECTION: THE DEVELOPMENT OF INJURY ASSESSMENT FUNCTIONS

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ABSTRACT

The objective of the ISIP Project has been to develop a methodology to allow vehicle designers to optimize safety systems of vehicles in side impacts. This optimization was based on the minimization of the cost of injury or Harm.

To form the link between the safety system protective capability in a crash and the cost of injury to the occupant required the development of a series of lateral impact Injury Assessment Functions (IAFs). These IAFs had to be able to predict the risk of injury, in AIS, for each of the major body regions of the occupant. The injury predictions were used to derive Harm for the crash and were based on the responses of a human surrogate, the BioSID.

This paper describes the development of these lateral injury IAFs from the analysis of cadaver test data.

INTRODUCTION

Vehicle design optimization for improved occupant safety in side impacts has been the goal of a collaborative research program sponsored by the Australian Research Council, Fildes et al, 2001. goal. This paper describes the development of lateral injury IAFs from the analysis of cadaver test data to fill the gaps in our current knowledge of lateral impact IAFs.

Table 1.
The Body Region Priorities for Side Impact Injury in Australia on the Basis of Harm

Priority	Body Region	Harm
1	Head	47%
2	Thorax (inc hlsk)	22%
3	Lower Extremities	9%
4	Upper Extremities	6%
5	Spine	4%
6	Face	4%
7	Pelvis	3%
8	Shoulder	2%
9	Abdomen	2%
10	Neck	2%
		100%

Note: hlsk is the heart, liver, spleen and kidneys.

The starting point for this work was a review of available injury tolerance data in lateral impacts, Gibson et al, 2000. The overall body region priorities for side impact injuries based on Harm are presented in Table 1. These priorities together with the availability of data determined the focus of IAF development for this paper.

The injury assessment functions derived here associate an injury level for a specific body region, in terms of the Abbreviated Injury Scale, AIS, AAAM, 1990 with a specific criteria or dummy response. Various approaches were used to develop the IAFs for lateral impacts:

- the head, lateral head impact test data was analysed with respect to the Head Injury Criteria (HIC) employing an approach similar to that used for frontal impacts by Mertz et al,, 1994,
- the thoracic, shoulder, abdominal and pelvic IAFs used the methodology developed by the ISO Road Vehicles Committee Working group as a starting point, ISO, 1999.

The later approach consists of using available cadaver side impact test data and equating the injuries recorded to specific dummy responses measured in identical tests. IAFs have been derived for the shoulder, thorax, abdomen, and pelvis. The BioSID was the base dummy for this analysis and the IAFs are therefore, already matched to BioSID responses.

Each IAF forms a bridge between the response of the occupant surrogate, the dummy, and the prediction of the Harm for a specific side impact crash. The dummy response may be derived by mathematical simulation or measured from a side impact test or crash reconstruction.

IAF DERIVATION

The analysis of injury data has always been the subject of some controversy. The methods of analysis fall into two broad classes:

1. Ordinal methods which rank the data by a particular injury indicator and assume that it is the prime influence upon injury including the Mertz-Weber method, certainty groups, and logistic analysis of certainty group data; and,
2. Non-ordinal methods (primarily logistic regression analysis using a variety of methods to group data).

Non-ordinal grouped logistic regression analysis produces IAFs that predict higher probabilities of injuries at low values and less for higher values. Non-ordinal methods usually indicate a higher probability of injury at low levels than seem warranted by the experimental evidence.

The ordinal methods however, assume that thresholds of injury and certain injury have been determined and that there is a known dominant factor predicting injury. These assumptions result in IAFs that indicate a smaller spread in the tolerance to injury amongst the general population. Both methods tend to intersect at the 50 percent risk of injury threshold.

This paper employs a pragmatic approach to the development of IAF curves. The final curves are, where possible, averaged area logistic curves based upon the averaged area of an ordinal analysis (a Mertz-Weber curve) and a non-ordinal analysis (logistic regression).

The averaged area curve is an attempt to overcome:

1. the exaggerated estimates of injury provided by non-ordinal analyses at minor impact severities when using sparse data, and

2. the exaggerated probabilities of injury that the ordinal methods produce for tests that are skewed towards the threshold of injury rather than the threshold of certain injury.

An advantage of this method is that an indication of the reliability of the final curve is provided by the divergence between the curves used to derive it. The steps involved in deriving an averaged area curve in logist form are presented in the appendix.

Lateral Head Impact IAFs

The Source of Data used to determining the probability of skull fracture or brain injury from lateral head impacts with respect to HIC were 58 cadaver tests. These tests included:

1. 28 impactor tests
   - 10 by Walsh, 1985,
   - 2 by Rizetti, 1997, and
   - 16 by McIntosh, 1994;
2. 8 pedestrian-car form impact tests by Walsh, 1985; and,
3. 22 drop tests by Got, 1978 and 1983.

These tests included lateral impacts against rigid and padded surfaces by helmeted and unhelmeted cadavers. The severity of injury was re-coded wherever possible with AIS-90 and the impact severity assessed using the Head Injury Criteria, HIC. The data was unadjusted for cadaver age as no correlation between the probability of injury and age was found.

Three outlying data points were excluded from any further analysis. For the remaining scores, it was apparent that the helmet tests were far more likely to reveal no discernible injuries with only 5% of helmeted compared to 57% of unhelmeted cadaver tests showing no signs of injury for impacts with a HIC score above 900.

The shortage of data meant that the helmet tests could not be excluded. It was found that the MAIS versus HIC scores for helmet tests were similar to that for unhelmeted tests providing the 8 non-injury helmet tests with impacts of above 900 HIC were discarded. These tests were therefore also censored prior to further analysis of the data leaving the 47 data points shown below.

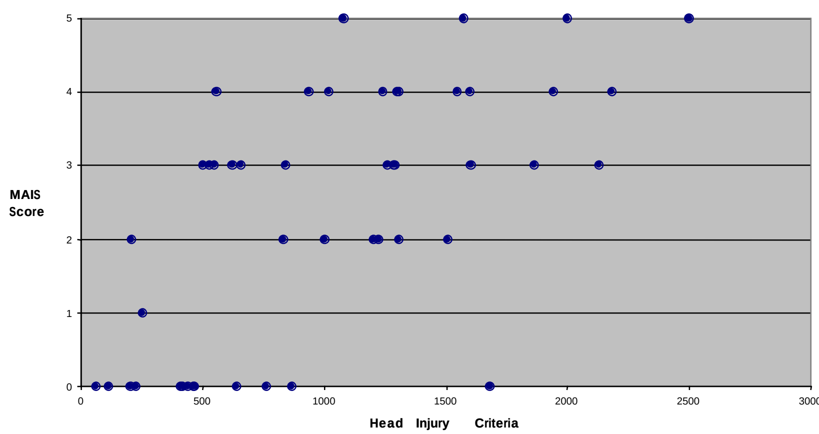

Figure 1. Censored brain & skull MAIS scores Vs HIC in lateral head impacts

An examination of available acceleration versus time curves indicated that the head accelerations were typically within a duration of 15ms.

Skull and Brain Injury Analysis of the scatter plot of these AIS scores for suggests that AIS 1 and 2 skull and brain injuries are under reported in cadaver tests. The signs for AIS 1 and 2 head injuries, such as headaches, dizziness, loss of consciousness, amnesia or lethargy, are not present in cadaver tests. The absence of normal vascular pressure and the post-trauma survival time also result in errors in the estimation of injuries of all severity.

With the cadaver data underestimating the likelihood of brain injuries, particularly minor ones, an IAF for AIS 1+ brain or skull injury was not developed.

Regression Analyses were performed using a variety of techniques. The first involved standard logistic regressions based upon appropriate (usually quintile) groups. Mertz-Weber curves using threshold estimates based on linear regressions of the thresholds of injury, were also formed. The final IAFs are based upon the averaged area logist curve that passes through the intersection of these curves.

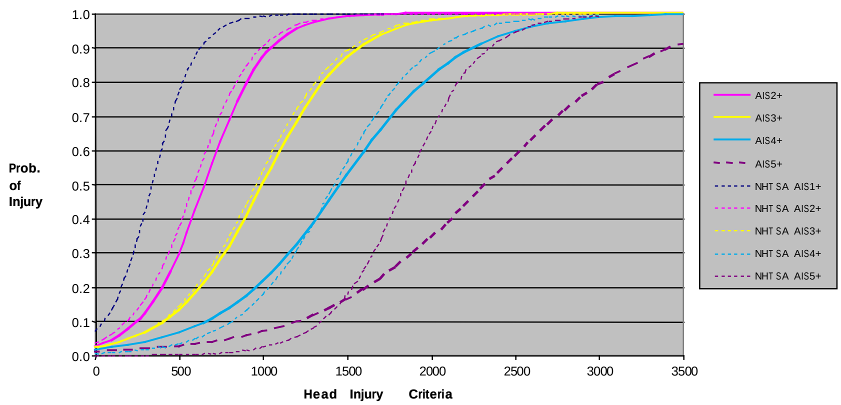

Figure 2. Probability of brain or skull injury vs HIC using averaged area logist curves compared with curves for frontal impacts from NHTSA

The averaged area curves are similar to the frontal impact curves from Kanianthra et al, 1996. The JARI tolerance curves for lateral impacts indicate very small differences between lateral and frontal impacts to the head for impacts of a duration of more than 8ms, Gibson et al. 2000.

The curves are reasonably consistent for AIS2+, 3+ and 4+ injuries. The AIS5+ curve is represented as a dashed line because the different methods of analysis produced widely divergent curves.

Hard Thoracic IAFs

The Source of Data used for the bony thoracic injury probability curves was 162 cadaver tests. The tests included:

1) 45 impactor/pendulum tests:
   - 16 by Viano et al, 1989;
   - 17 by Talantakite, 1998;
   - 12 by Robbins and Lehman, 1979; and
   - 6 by Chung, 1999.
2) 91 sled tests:
   - 17 by Cavanaugh, 1990 and 1993;
   - 36 by Kallieris, 1990 and 1994;
   - 26 by Pintar, 1997; and
   - 12 by Robbins & Lehman, 1979;
3) 26 drop tests by Tarriere et al, 1979.

The 26 sled tests by Pintar were excluded because the impacted surface was lower than in other sled tests and equivalent BioSID tests were not available.

Ninety of the remaining 136 tests were able to be matched, according to the padding used and the sled speed, impactor kinetic energy or drop height with BioSID dummy tests. These BioSID tests included:

1) pendulum tests conducted by:
   - Viano et al, 1995; and
   - Chung et al, 1999.
2) sled tests by:
   - Harigae et al, 1991; and
   - Cavanaugh et al, 1995.
3) drop tests by Harigae et al, 1991.

Adjustments to Data were required for both the injuries recorded and the dummy responses. Where the number of fractured ribs (NFR) could be obtained (irrespective of the cadaver's age) the results were adjusted for age according to the following formula used by the ISO, 1999.

$$NFR_{age\ adjusted} = NFR + (45\text{-}Age) \times 0.2 \qquad 1.$$

Then the number of fractured ribs was adjusted again to take into account the greater strength of the living, Viano and Lau, 1986:

$$NFR_{live} = NFR_{age\ adjusted} - 2 \qquad 2.$$

Finally the live adjusted scores were then converted into Abbreviated Injury Scales (AIS) scores.

In some tests the only data available was the total number of rib fractures (TRF) rather than the number of fractured ribs. A relationship between the number of fractured ribs and the total number of rib fractures was derived from tests where both data were recorded, providing restrictions to cadavers age were applied. Impactor data being used for impactor tests and, sled test data for sled and drop tests.

These relationships were used to estimate the number of rib fractures where only the TRFs were known. This score was then adjusted for age and living and converted into AIS scores as described above.

The maximum chest deflection and V•C scores of the BioSID dummy were then adjusted for differences between the kinetic energy of the impactor or sled velocity used in BioSID and cadaver tests in the following manner (drop test adjustments were not required):

Impactor Adjustment

$$def_{adjusted} = def \times (KE_{cadaver\ imp}/KE_{BioSID\ imp}) \quad 3.$$

$$V{\bullet}C_{adjusted} = V{\bullet}C \times (KE_{cadaver\ imp}/KE_{BioSID\ imp}) \quad 4.$$

Sled Test Adjustment

$$Def_{adjusted} = def \times \left({}^{V_{cadaver}}/{}_{V_{BioSID}}\right)^2 \quad 5.$$

$$V{\bullet}C_{adjusted} = V{\bullet}C \times \left({}^{V_{cadaver}}/{}_{V_{BioSID}}\right)^2 \quad 6.$$

A scatter plot of the final scores is shown below.

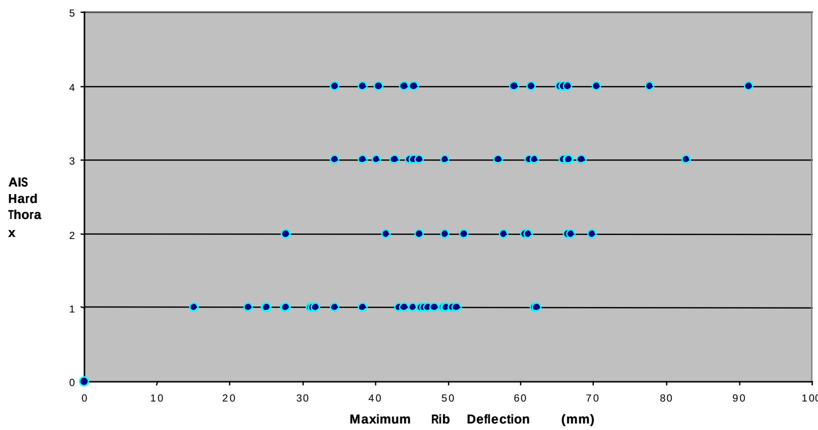

Figure 3. AIS Hard Thorax Vs Biosid Rib Deflections

Regression Analysis of the aggregated total of 90 data points shown above was performed to obtain the IAFs for the probability of AIS 1+, 2+, 3+ and 4+ hard thorax injuries from BioSID rib deflections in lateral impacts. BioSID deflections had a far greater correlation with hard thorax injuries than did BioSID V•C values.

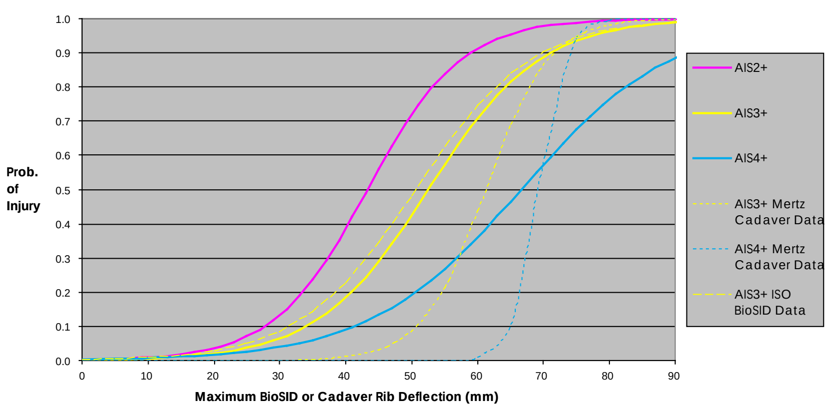

Figure 4. Comparison of IAFs for hard thorax injury vs rib deflection in lateral impacts (ISIP BioSID vs Mertz cadaver and ISO BioSID curves)

A method of analysis similar to that employed for the brain and skull injury assessment functions was used.

The different methods of statistical analysis were reasonably consistent and varied in the usual fashion with the Mertz method curve being slightly steeper than the logistic curve. The divergence found between the Mertz and logistic AIS 4+ injury curves casts some doubt upon the final averaged area AIS 4+ curve.

A graph illustrating the IAF for the probability of injury versus BioSID deflections is presented above and compared with BioSID response and cadaver IAFs derived in previous research. It should be noted that the divergences between the dummy and cadaver curves is to be expected given the greater stiffness of the dummy thorax, Gibson et al, 2000. The IAF for the risk of AIS 3+ injuries versus BioSID deflections derived by the ISO, 1999, is in agreement with IAF for AIS 3+ injuries derived in this study.

Soft Thoracic IAFs

The Source of Data for soft thorax injury are as detailed in the hard thorax injury section. The 26 sled tests by Pintar et al, 1997, were again excluded for lack of matching dummy tests. Ninety of the remaining 136 tests were then matched to BioSID dummy tests of similar impact velocities and identical padding as per the hard thorax injury analysis. Matching tests with identical rather than similar padding was required because differences in padding significantly reduced the correlation between thorax injuries and maximum V•C in the BioSID.

Additionally, a further 6 tests were excluded because of a lack of soft thorax injury data. Finally, low speed impacts of less than 4.5m/s were excluded because impactor tests by Viano, 1989, suggest that these impact speeds cause crush rather than viscous injuries and would therefore be better predicted by deflection rather than V•C. This left 62 tests for the soft thorax analysis - 39 impactor, 25 sled and 11 drop tests.

Censoring the matched data increased the correlation between the probability of receiving soft thoracic injuries of a given level of severity versus V•C using logistic regressions from approximately 0.7 to 0.85. Censoring the data also resulted in more realistic

IAFs because they indicate more reasonable probabilities of injury at 0m/s V•C. A scatter plot of the data is provided below.

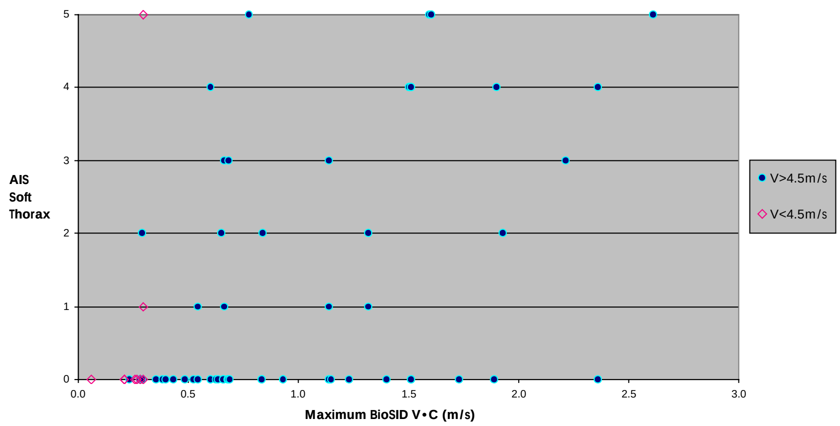

Figure 5. AIS soft thorax Vs BioSID V•C in lateral impacts

Adjustments to Data were made with respect to the viscous criteria, maximum V•C, of the matched dummy tests. Injuries were coded according to AIS scores but included injuries to the spleen, liver and kidneys. Correlations between the injury severity and maximum V•C was significantly improved by adjusting for differences in the kinetic energy between the cadaver and BioSID tests by using the following formulae:

Impactor Tests

$$\text{Adjusted V•C} = \text{BioSID V•C} \times \frac{\text{Cadaver Impactor KE}}{\text{BioSID Impactor KE}} \quad 7.$$

Sled and Drop Tests

$$\text{Adjusted V•C} = \text{BioSID V•C} \times \frac{(\text{Cadaver Impact Speed})^2}{(\text{BioSID Impact Speed})^2} \quad 8.$$

An age adjustment was applied to the AIS scores for internal injury with negative scores zeroed and scores above 5 lowered to 5. The age adjustment formula is:

$$\text{Adj. AIS}_{\text{soft thorax}} = \text{rounded}[\text{AIS}_{\text{soft th}} - 0.008\,(\text{Age}-45)] \quad 9.$$

Regression Analysis of the age adjusted internal thorax scores illustrates the relationship between the adjusted soft thorax injury score and adjusted V•C. Logistic regression using quintile groups reveals that there is a stronger correlation between AIS 3+/4+, and V•C than AIS 2+ injuries of the soft thorax and V•C.

The curves generated by logistic regression still indicated an improbably high risk of injury at 0 m/s V•C. This is a tendency of logistic regressions in this area of work that is partly due to the method of analysis upon small numbers of non-threshold tests. The same process that was employed in the analysis of the head injury data was used to overcome this problem.

The generated curves are compared with the ISO AIS 3+ curve below. The AIS 3+ curve has a similar overall shape but is shifted left indicating a lower threshold to injury. The IAFs for impact speeds greater than 4.5m/s, are presented below. A dashed AIS 5+ curve is used because it is based on only 4 AIS5+ injury points.

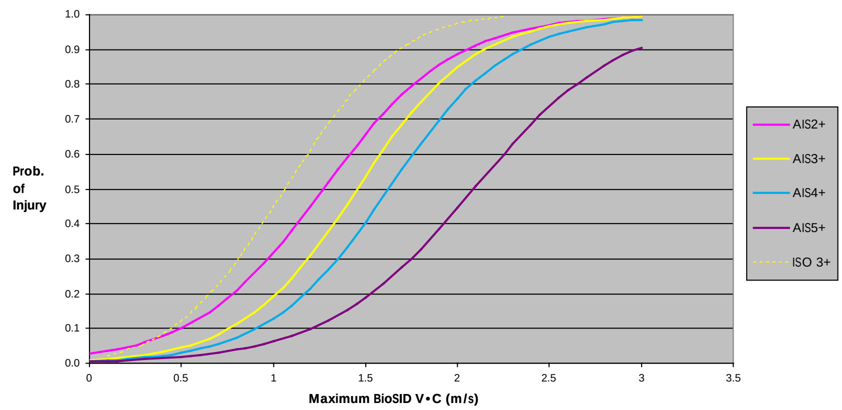

Figure 6. Comparison of soft thorax injury vs V•C IAFs in high V•C lateral impacts (averaged area Vs ISO 3+ cadaver curves)

Shoulder Injury IAFs

The Source of Data for the shoulder analysis was obtained from 39 cadaver sled tests. Only sled tests were used as there were no dummy tests to match to the cadaver drop or pendulum tests. The tests included:

1) 12 tests by Robbins and Lehman, 1979;
2) 10 tests by Kallieris et al, 1981; and,
3) 17 tests by Cavanaugh et al, 1993.

The Robbins and Lehman tests were excluded from the analysis because the shoulder injuries reported were dramatically lower than in the other test series. Twenty of the remaining 27 tests were then matched to BioSID tests with impacts of similar speed.

Adjustments to the Data were made for cadaver age and differences in the impact speeds of the cadaver and the dummy test to which it was matched using the following formulae:

$$\text{'T1' Acceleration}_{\text{speed adjusted}} = \text{'T1' Acc.} \left({}^{V_{cadaver}} / {}_{V_{BioSID}} \right)^2 \quad 10.$$

$$\text{'T1' Acceleration}_{\text{age and v adjusted}} = \frac{\text{'T1' Acc.}_{\text{speed adjusted}}}{(1 - 0.004(\text{Age} - 45))} \quad 11.$$

The scatter plot of resulting 15 data points is shown below, Figure 7.

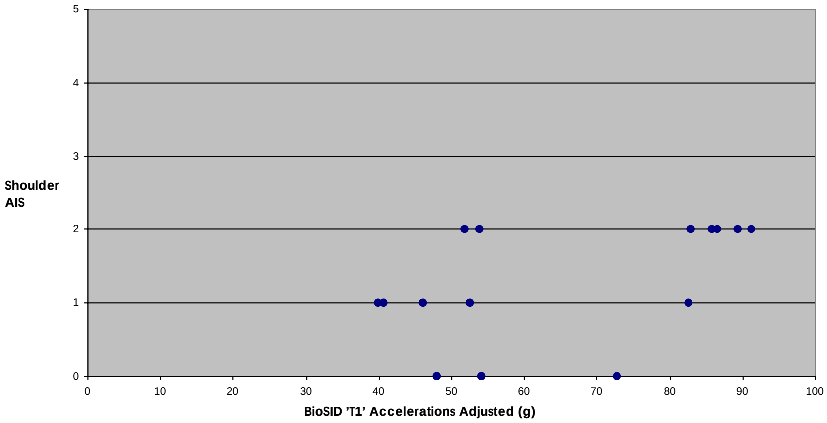

Figure 7. Shoulder AIS vs BioSID 'T1' accelerations in lateral impacts

Regression Analyses based upon this number of points are statistically unreliable. The data was analysed using Mertz-Weber method, logist regressions of certainty group data based on 5g intervals and certainty point analyses (see appendix). Each of these techniques was compared with a two group logist analysis to determine if there was any consistency between the curves derived from each approach.

This analysis revealed that the most reliable IAFs would be generated by using average area logist functions based on the Mertz-Weber method and logist regressions of certainty group data. The later were used rather than the standard logist analysis because of the lack of data points.

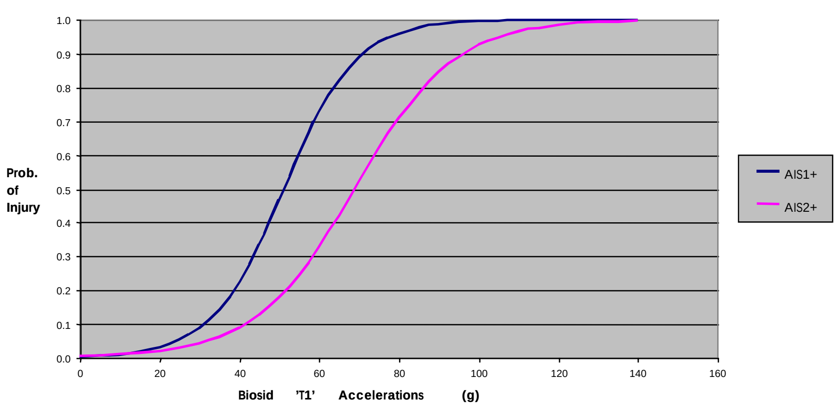

Figure 8. Shoulder injury probability vs BioSID 'T1' accelerations in side (averaged area curves of logist certainty point & Mertz

The risk of a given severity of injury to the shoulder with respect to BioSID 'T1' accelerations are illustrated above. 'T1' accelerations are used as the predictor of injury because this was the only data available consistently from dummy shoulder impact tests. The limited number of data points used to derive these curves mean they are simply guides to the risk of injury in the absence of more reliable data.

The BioSID lacks a centrally mounted spine and so the shoulder injury risk curves would be improved if they were a function of BioSID rib 1 deflections

Abdominal Injury IAFs

The Source Data used to assess the likelihood of injury for the abdomen was limited to lateral impactor tests to the abdomen and lateral abdominal drop tests against a rigid or padded protrusion that simulated an armrest. No sled tests were used for two reasons. First, it was impossible to distinguish between injuries that resulted from lateral impacts to the chest or abdomen in sled tests against a flat wall. The second reason was that no cadaver tests were found that matched those of BioSID dummies against walls with an abdominal offset.

The data for this analysis included:

1) 14 cadaver pendulum tests conducted by Viano et al, 1989; and,
2) 11 drop tests onto armrest forms by Walfisch et al, 1980.

Of these 25 tests, 12 pendulum tests were matched to dummy pendulum tests performed by Viano et al, 1995, 3 drop tests were matched to dummy drop tests performed by Harrigae et al, 1991, and 3 to tests by Bendjellal et al 1991. This left a total of 18 matched tests. Unfortunately Harrigae did not report V•C or rib deflections for any of his dummy drop tests and Bendjellal did not report maximum V•C values either. This left only 12 abdominal injury versus V•C data 15 abdominal injury versus rib deflection data points.

Adjustments to Data were again made for age and living in the rib fracture data and for differences in the kinetic energy of the impact between the cadaver and BioSID tests. The soft Tissue injury scores however, were unadjusted. The drop test heights matched exactly and therefore required no adjustment in the dummy response.

The scatter plots for rib, organ and MAIS of rib and organ injuries versus maximum 4-5th BioSID rib deflections and maximum V•C are presented in the two graphs above. They suggest that there **may** be a strong relationship between internal injuries and V•C in lateral impacts however there are insufficient points to determine the nature of this relationship.

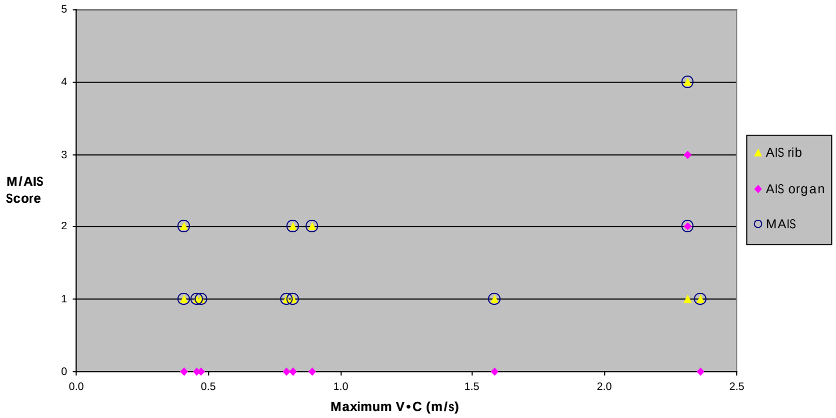

Figure 9. AIS Abdominal Injury Levels by Maximum V•C in Lateral Impacts

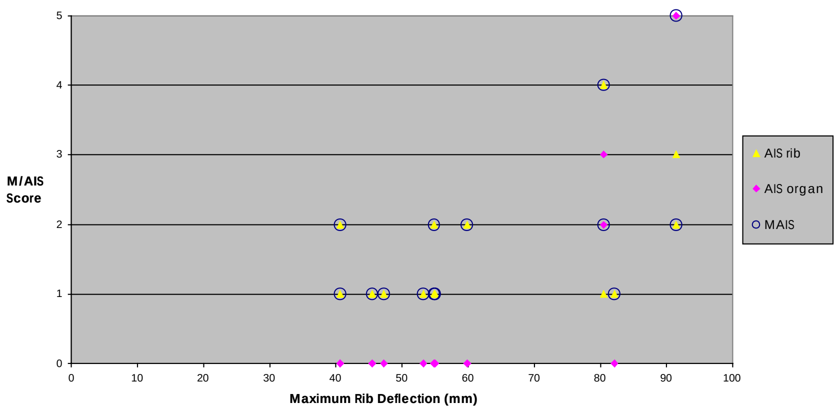

Figure 10. M/AIS abdominal injury levels vs maximum 4-5th BioSID rib deflections in lateral impacts

The dominance of rib fractures in determining the MAIS scores and the spread of data points suggests that the most constructive regression analysis that could be performed would be for probable MAIS injury levels versus maximum 4/5th BioSID rib deflections. This variable also provides three extra data points for analysis than maximum V•C.

Regression Analyses of MAIS injury versus maximum rib deflection scores were based upon a logistic analysis of triadic groups to obtain an estimate of the probability of MAIS 2+ injuries as a result of lateral abdominal impacts. The Mertz method of deriving injury curves could not be used because of the lack of clear upper and lower thresholds of injury.

The curve for MAIS 2+ injuries is compared with the ISO, 2000 AIS 3+ curve below. The lack of data points makes the curve statistically unreliable. It does however, resemble that derived by the ISO for MAIS 3+ abdominal injuries.

Walfisch's drop tests total rib fracture scores were converted into estimates of the number of fractured ribs. The actual or estimated number of fractured ribs in each test was then adjusted for age and living and converted into AIS rib fracture scores as per the hard thorax analysis. The maximum of the rib and soft tissue AIS scores was then used for further analysis.

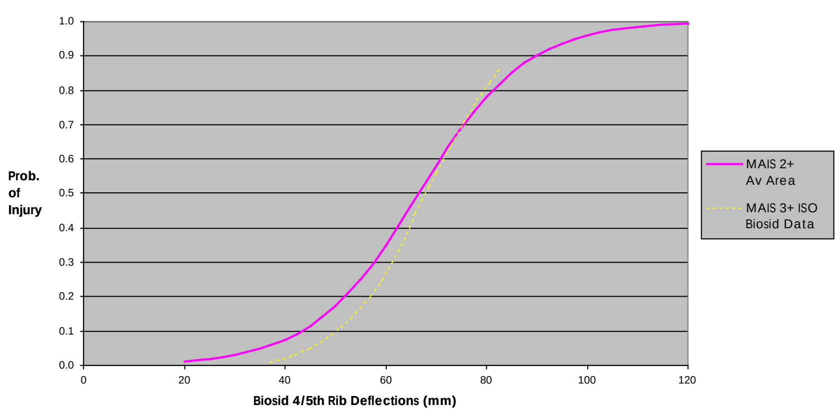

Figure 11. Comparison of abdominal MAIS Vs BioSID rib deflection IAFs in lateral abdominal impacts

In the impactor tests it was necessary to adjust the BioSIDs maximum V•C and 4-5th rib deflections for differences in the kinetic energy between the cadaver and dummy impactors.

Pelvic Injury IAFs

The Source of Data was 182 cadaver tests. The tests included:

1) 129 impactor tests with
   - 63 by Cesari, 1980,
   - 21 by Nuscholtz et al, 1986,
   - 11 by Bouquet et al, 1998,
   - 14 by Viano, 1989 and
   - 20 by INRETS during 1992-94;
2) 27 sled tests with
   - 10 by Kallieris et al, 1981 and
   - 17 by Cavanaugh et al, 1990; and
3) 26 drop tests by Tarriere et al, 1979.

The ten sled tests by Kallieris et al, 1981 were excluded because no pelvic injuries were reported.

Seventy three impactor tests were then matched to similar dummy tests conducted by Kanianthra et al, 1991, Harigae et al, 1991, Viano, 1989 and Bendjellal et al, 1991. Five sled tests were matched to dummy tests by Harigae, Kanianthra and Bendjellal and 12 drop tests to dummy tests by Harigae. This gave a total of 90 matched tests.

Injuries to the pelvis in the cadaver tests appear to be closely correlated to the impact velocity in impactor tests. The BioSID impactor tests on the other hand, showed that the measures of impact severity on the BioSID dummy are more closely related to kinetic energy rather than impact force. This lead to some difficulties in matching likely injury outcomes to BioSID indicators of pelvic injury where the effective mass of the impacting mass is unknown.

Adjustments to Data were made for cadaver age and differences between the cadaver and BioSID tests. The measured outcomes of the BioSID were first adjusted for the differences in kinetic energy in impactor tests and for differences in velocity squared in sled tests (no adjustments were required for the drop tests). The adjustment formulae are listed below

Impactor Tests

$$\text{Pelvic Acc}_{adj} = \text{Pelvic Acc}_{BioSID} \times \frac{KE_{cadaver\ impactor}}{KE_{dummy\ impactor}} \quad 12.$$

Sled Tests

$$\text{Pelvic Acc}_{adj} = \text{Pelvic Acc}_{BioSID} \times \left({}^{V_{cadaver}}/{}_{V_{BioSID}}\right)^2 \quad 13.$$

The final step was to adjust these outcomes for variations in the age of the cadavers, ISO, 1999.

$$\text{Pelvic Acc.}_{final} = \frac{\text{Pelvic Acc}_{speed\ adjusted}}{(1 - 0.004(Age - 45))} \quad 14.$$

The scatter plot for the adjusted data is illustrated below. The two outlying data points shown as circles were excluded from the analysis.

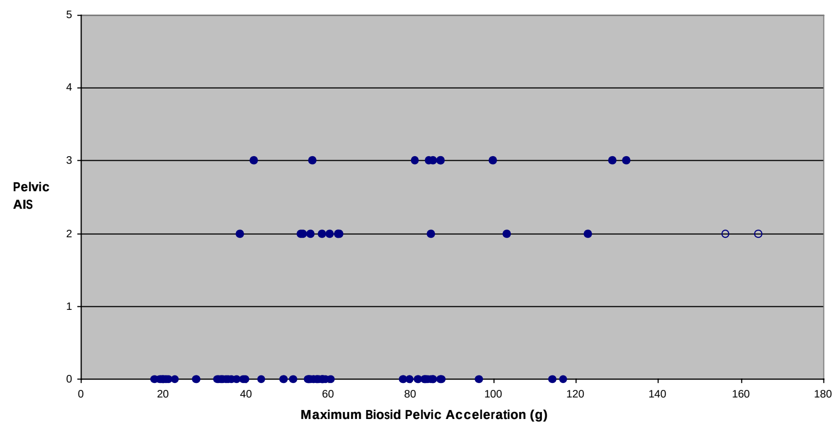

Figure 12. Pelvic AIS vs pelvic acceleration in lateral pelvic impacts

Regression Analyses were conducted to derive an averaged area curve based on Mertz' method and logistic regressions based upon hexile groups. Each method of statistical analysis generated similar IAFs for AIS 2+ and somewhat divergent curves for AIS 3+ injuries. This indicates that the AIS 2+ injury curve is more reliable.

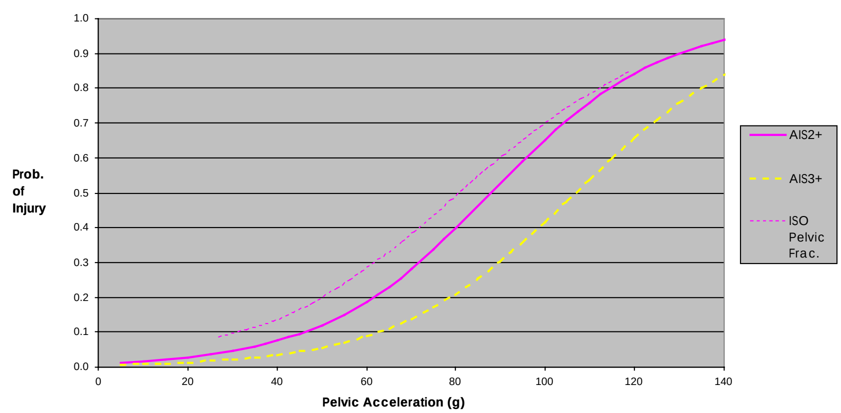

Figure 13. Pelvic injury probability vs BioSID pelvic accelerations in lateral pelvic impacts

The final curves are illustrated above, and compared with the ISO, 1999 curve for pelvic fracture. The AIS 2+ injuries in this report are equivalent to the ISO pelvic fracture curve.

SUMMARY

Lateral impact IAFs have been developed here for the head, thorax, shoulder, abdomen and pelvis. These IAFs form a bridge between the response of BioSID dummy and the prediction of Harm for the specific side impact crash. They extend the number of lateral IAFs available for head, thorax and pelvic injuries and provide new ones for the shoulder.

REFERENCES

AAAM: Abbreviated Injury Scale - AIS 90, Association for the Advancement of Automotive Medicine, 1990.

Bendjellal F, Fuld G, et al: EUROSID-1 and BIOSID Impact Response Characteristics vs ISO Biofidelity Requirements, 13th ESV Conference Proceedings, Paris, 1991.

Bouquet R, Ramet M, et al: Pelvis Human Response to Lateral Impact, 16th ESV Conference Proceedings, Windsor, 1998.

Cavanaugh J et al: A Comparison of the Biomechanical Responses of SID, BIOSID and Cadavers in Side Impact, Proceedings of WSU Biomechanics Symposium, Detroit, 1995.

Cavanaugh JM, Waliliko TJ, et al: Injury and Response of the Thorax in Side Impact Cadaveric Tests, 34th Stapp Car Crash Conference Proceedings, SAE Paper 902307, 1990.

Cavanaugh JM, Walilko TJ, et al: Biomechanical Response and Injury Tolerance of the Pelvis in 12 Sled Side Impact Tests, 34th Stapp Car Crash Conference Proceedings, SAE Paper 902305, 1990.

Cesari D and Ramet M: Pelvic Tolerance and Protection Critera in Side Impact, 26th Stapp Car Crash Conference Proceedings, SAE Paper 821159, 1982.

Chung J, Cavanaugh JM, et al: Thoracic Injury Mechanisms and Biomechanical Responses in Lateral Velocity Pulse Impacts, 43rd Stapp Car Crash Conference Proceedings, San Diego, SAE Paper 99SC04, 1999.

Fildes B, Sparkes, L et al: Improved Side Impact Protection (ISIP) in Australia: Overview of a Collaborative Approach, Unpublished Paper for ESV Conference Proceedings, 2001.

Gibson T, Fildes B, et al: Improved Side Impact Protection: A Review of Injury Patterns, Injury Tolerance and Dummy Measurement Capabilities. MUARC Report Number 147, Monash University, May, 1999.

Got C, Patel A, et al: Results of Experimental Head Impacts on Cadavers: The Various Data Obtained & Their Relations to Some Measured Physical Parameters, SAE Paper 780887, 1978.

Got C, Guillon F, et al: Morphological & Biomechanical Study of 146 Human Skulls Used in Experimental Impacts, in Relation with the Observed Injuries 27th Stapp Car Crash Conference Proceedings, SAE Paper 831619, 1983.

Harigae T, Ohsaki K, et al: Evaluation of Impact Responses of the Eurosid-1 and BioSID, 13th ESV Conference Proceedings, Paris, 1991.

ISO: Lateral Collision Protection-Performance Criteria Expressed in Biomechanical Terms. ISO TC22/SC12/WG6 Document N517, November, 1999.

Kallieris D, Mattern R, et al; Quantification of Side Impact Responses and Injuries, SAE Paper 811009, 1981.

Kallieris D, Mattern R, et al; The Biofidelity of EUROSID 1 and BIOSID, 36th Stapp Car Crash Conference, Washington, SAE Paper 922518, 1992.

Kanianthra JN, Wilke TJ, et al: Comparative Performance of SID, BIOSID and EUROSID in Lateral, Pendulum, Sled and Car Impacts, 13th ESV Conference Proceedings, Paris, 1991.

Kuppa S: personal communication, 1st January, 2000

Lau IV and Viano DC: How and When Blunt Injury Occurs - Implications to Frontal and Side Impact Protection, 32nd Stapp Car Crash Conference Proceedings, Atlanta, SAE Paper 881714, 1988.

McIntosh, AS: Pedal Cycle Helmet Effectiveness UNSW PhD Thesis, July, 1994.

Mertz H: Procedure for Normalising Impact Response Data, SAE Paper 840884, 1984.

Mertz HJ and Irwin AL; Brain Injury Risk Assessment of Frontal Crash Test Results, SAE Paper 941056, 1994.

Mertz HJ, Prasad P and Nusholtz G, Head Injury Risk Assessment for Forehead Impacts, SAE International Congress and Exposition, Detroit, SAE Paper 960099, 1996.

Nusholtz GS and Kaiker PS: Pelvic Stress, Journal of Biomechanics, Vol. 19, No. 12, 1986.

Pintar FA, Yoganandan, N, et al: Chestband Analysis of Human Impact Tolerance to Side Impact', 41st Stapp Car Crash Conference Proceedings, SAE Paper 973320, 1997.

Ridella SA and Viano DC: Determining Tolerance to Compression and Viscous Injury in Frontal and Lateral Impacts, 34th Stapp Car Crash Conference Proceedings, SAE Paper 902330, 1990

Rizetti A, Kallieris D, et al: Response & Injury Severity of the Head-Neck Unit During a Low Velocity Head Impact, 21st IRCOBI Conference Proceedings, Hanover, 1997.

Robbins D, Lehman R, et al: Prediction of Thoracic Injuries as a Function of Occupant Kinematics, 7th ESV Conference Proceedings, Paris, 1979.

Talantakite Y, Bouquet R, et al: Human Thorax Behaviour for Side Impact – Influence of Mass and Velocity, 16th ESV Conference Proceedings, Windsor, 1998.

Tarriere C, Walfisch G, et al: Synthesis of Human Tolerance Obtained from Lateral Impact Simulations, 7th ESV Conference Proceedings, Paris, 1979.

Viano D: Biomechanical Responses and Injuries in Blunt Lateral Impact, 33rd Stapp Car Crash Conference Proceedings, Washington, SAE Paper 892432, 1989.

Viano DC, Lau IV, et al: Biomechanics of the Human Chest, Abdomen and Pelvis in Lateral Impact, 33rd AAAM Conference Proceedings, Baltimore, 1989.

Viano D, Fan A, et al: Biofidelity and Injury Response in Eurosid 1 and BioSID, 39th Stapp Car Crash Conference Proceedings, San Diego, SAE Paper No. 952731, 1995.

Walfish G, Fayon A, et al: Designing a Dummy's Abdomen for Detecting Injury in side Impact Collisions, 5th IRCOBI Conference Proceedings, Birmingham, 1980.

Walsh MJ, Kelleher BJ, et al: Head & Neck Injuries in Human Cadavers from Lateral Impacts, 10th ESV Conference Proceedings, London, 1989.

APPENDIX

Derivation of Injury Probability Curves

The steps in developing the various curves used in this paper are to develop:

A) Mertz-Weber Ordinal Logistic Curves based on the Mertz-Weber method (Mertz, 1984) is derived by:

1) arranging the data in ascending order of the injury indicator and assigning an ordinal value from 1 to n,

2) determining the lower and upper threshold of injury for AIS 1+, 2+, 3+, 4+ and 5+ levels by removing outlying data points and taking the lowest value associated with the injury level being investigated and the highest value associated with a lesser levels of injury,

3) assuming the injuries are normally distributed about the mean of these thresholds the probability of injury at each threshold is calculated using the formulae:

$$p(\text{lower threshold}) = (1- 0.3)/(n + 0.4) \quad \text{A1.}$$

$$p(\text{upper threshold}) = 1- p(\text{lower threshold}) \quad \text{A2.}$$

with n being the number of data points between and including the two thresholds,

4) plotting the curve on a normal probability axis by and drawing a line between the two points. Alternatively the appropriate normal curve z-score for the injury probability calculated at each threshold can then be found in normal distribution tables and the z-scores calculated at regular probability intervals.

B) Non-ordinal Logistic Curves by:

1) arranging the data in ascending order of the injury indicator,

2) forming quintile, hexile or other groups,

3) finding the probability of injury and average value of the injury indicator for each group, and

4) performing a logistic regression for the probability versus average injury indicator value data points and determining the value of the logistic parameters A_n and B_n.

C) Area Averaged Logistic Curves by finding the parameters A_a and B_a for the logistic curve that passes through the point of intersection and divides the area of the two previous curves equally using the formulae:

$$A_a = \frac{2A_oA_n}{(A_o + A_n)} \qquad \text{A3.}$$

$$B_a = (A_n - A_a)(x_i) + B_n \qquad \text{A4.}$$

or

$$B_a = (A_o - A_a)(x_i) + B_o \qquad \text{A5.}$$

where x_i is the injury indicator value at which the curves intersect and is given by

$$x_i = \frac{-(B_o - B_n)}{(A_o - A_n)} \qquad \text{A6.}$$

An example of curves derived by this process is shown below.

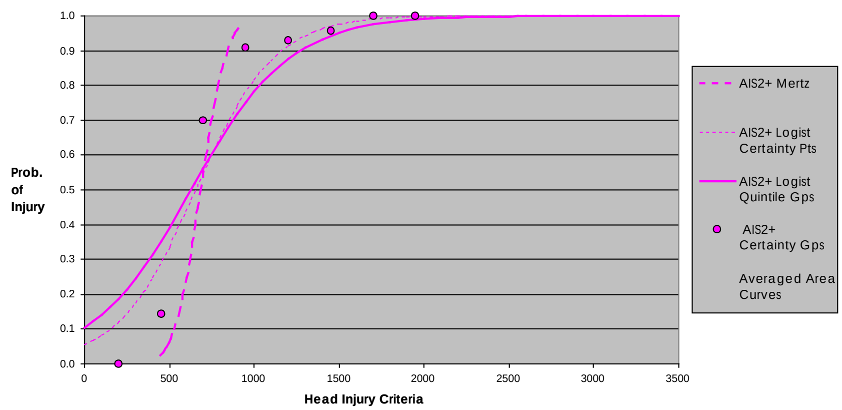

Figure A1. Probability of MAIS 2+ brain and skull injury vs HIC in lateral head impacts using various analysis techniques

Alternative Methods of Analysis

Certainty Point or Certainty Group Points represent an alternative to the above approaches. Certainty points are based on an ordinal approach that generates curves that are generally similar to those derived by the Mertz method. The primary disadvantage of this analysis technique is that it will always indicates that injury is certain to occur at the highest test value regardless of the number and range of test conditions.

Certainty group curves are derived by:

1) arranging the data in ascending fixed injury indicator intervals and assigning an ordinal value from 1 to n, and

2) assigning a probability of injury for each interval from 0 to 1 by dividing the number injured in equal or lower HIC intervals by the total of the number of non-injured in equal or higher HIC test intervals plus the number of injured in equal or lower HIC tests. An example formula for probabilities of fracture is:

$$p(i) = \frac{\sum_1^i frac}{\sum_1^i frac + \sum_i^n non-frac} \qquad \text{A7.}$$