BIOMECHANICAL BASIS FOR INJURY CRITERIA USED IN CRASHWORTHINESS REGULATIONS

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ABSTRACT

A historical review of frontal and side impact regulations in the U.S.A. and Europe has been conducted. The biomechanical basis of injury criteria utilized in these regulations have been examined. It is shown that the biomechanical database utilized in the current impact regulations are based on research conducted before the mid-1980's. Research conducted since then has not had any impact on regulatory injury criteria, although proposed upgrades in the US FMVSS208 has attempted to synthesize all available data to develop tolerance levels for various size dummies.

HISTORICAL BACKGROUND: THE EARLIEST regulation utilizing full scale frontal crash testing of vehicles with test dummies is the FMVSS208 (1973). The crash test dummy used was the Hybrid II dummy, developed by General Motors in 1972, mainly to assess the integrity of lap and shoulder belt restraint system. Whereas, this dummy had the size, shape and mass distribution of a 50" percentile adult male, its biofidelic response in various parts of the dummy (e.g. head, neck, and chest) were suspect. It had limited measurement capabilities - head accelerations, spinal accelerations, and axial femur loads. Due to lack of biofidelity and limited instrumentation, the injury criteria specified in the FMVSS208 were based on head and chest accelerations, and axial femur loads.

While compliance to the FMVSS208 continued with the Hybrid II dummy for many years, substantial biomechanical impact research was initiated for establishing mechanical responses and associated injury criteria for a new generation of crash test dummies for frontal impact evaluation of vehicles. A new dummy called the Hybrid III, was developed by G.M. with substantial improvement in biofidelity over the Hybrid II dummy in all parts of the dummy. Along with improved biofidelity, measurement capabilities were substantially extended to measure more meaningful impact responses like neck loads, chest deflections, forces and moments in the femur and tibia. In 1984, this dummy was adopted as an alternate dummy to demonstrate compliance with the FMVSS208. Subsequently, the Hybrid II dummy was phased out of the FMVSS208, and the Hybrid III dummy has become the only dummy that can be used for compliance purposes. With the introduction of the Hybrid III dummy in 1984, Mertz (1) proposed a set of Injury Assessment Reference Values associated with various responses being measured in the dummy. Not all of these responses were chosen for this initial regulation. Head Injury Criteria, spinal acceleration, chest deflections, and femur loads were the only responses regulated by the FMVSS208. The FMVSS208 was further modified in 1998 to include neck loads as regulated responses.

Frontal Crash Regulations in Canada, Europe, Japan, and Australia have also accepted the Hybrid III dummy as the test device with controls over the measured dummy responses. The limits specified on the dummy responses vary somewhat from the FMVSS208. Whereas the FMVSS208 specifies controls over head, neck, chest, and femur responses, the European directive (96/79 ECE Directive) has additional controls placed on the tibia. The ECE directive for frontal impact has basically adopted all the Injury Assessment Reference Values proposed by Mertz (1, 2) with some minor modifications. The biomechanical origins and basis of these IARV's will be discussed later in this paper. It should be mentioned that these IARV's as suggested by Mertz were based on the state of the art in 1984, and spurred considerable reexamination and basic research in the biomechanics of neck, thoracic and lower limb responses and injury criteria. Further modifications of regulated injury criteria have not occurred, but are expected in future rulemaking activities.

Whereas worldwide regulatory frontal crash tests currently utilize the Hybrid III dummy, two substantially different crash test dummies are used for side impact regulatory purposes. The earliest dummy designed for side impact testing is commonly known as the USSID and became the regulatory dummy in the US FMVSS2 14 standard in 1984. At the time of the regulation, repeatability and reproducibility of the dummy design was established and the associated injury criteria was already researched. A parallel dummy development with additional measurement capability was ongoing in Europe. result&in the development and acceptance of this dummy, Eurosid I, in European side impact directive that followed a few years after the promulgation of the FMVSS214. Since the Eurosid I had biofidelity for lateral impacts in the chest and abdomen and load measurement capability in the pelvis, it is not surprising that these responses were regulated in Europe. In terms of biomechanical basis for injury criteria used with the 2 dummies, the USSID used acceleration-based criteria, Thoracic Trauma Index, and pelvic acceleration for which the dummy was designed. The Eurosid I was designed for biofidelity in chest deflection, abdominal and pelvic loads. The injury criteria adopted were deflection and load based.

To summarize, the regulatory injury criteria are highly test device specific, and are based on the biofidelity and measurement capability of the dummies, and the state of biomechanic information available at the time of the regulation was promulgated.

The Biomechanical Basis of Regulatory Injury Criteria-

The biomechanical basis for injury criteria is established by tests performed on animals and human cadaveric specimens. The tests generally include measurement of dynamic responses to known stimulus, e.g. forces, accelerations, etc., and failures, if any, in the specimens to the stimulus. In general the mechanisms of injuries are established through dose/response type of analysis. Threshold levels of injury producing stimulus are established, and the latest trend is to develop risk curves that would predict the percent of population expected to be injured versus the stimulus. In many cases voluntary exposures to known stimulus establish non-injury-producing levels.

If large amount of well defined accident data are available, it is possible to reconstruct accidents of varying severity with anthropomorphic test devices in crash environments. These test results can be compared to injury outcomes of accident victims to establish the levels of stimulus that produce certain levels of injuries in the real world.

Further biomechanical basis for injury criteria can be provided by theoretical, mathematical modeling studies. These studies can establish the validity of the utilized injury criteria on an engineering basis.

The following parts of this paper examines the biomechanical basis of injury criteria currently being used in governmental crash regulations and crashworthiness assessment testing worldwide.

Head Injury Criteria: The currently used worldwide regulatory criteria for controlling head injuries is commonly known as the HIC. Ever since its adoption in the FMVSS208 in 1972, this criteria has been controversial (3). It is interesting to trace the development of HIC to evaluate its biomechanical basis.

The earliest tolerance criteria for head injuries was introduced by Lissner et al. (4) in 1960 and is known as the Wayne State Tolerance Curve. This curve was based on cadaver skull fracture data, animal and human volunteer data. Further examination of the WSTC was conducted in Japan over several years of research with impact experiments on animals and human cadaveric skulls (5). Once again, scaling techniques were used to derive threshold of concussion curve, and turned out to be very similar to that originally proposed by Lissner et al. (4) and modified in the long duration regime by Patrick (6). Theoretical justification of this curve was provided through finite element modeling of the skull and the brain by Ruan and Prasad (7). In this study, iso-stress curves for the skull and the brain were developed as a function of average head acceleration and its time duration, Fig 1.

As early as in 1966, Gadd analyzed the basic biomechanical data supporting the WSTC and other animal human exposure data to propose the concept of severity index for evaluating head injury potential. The severity index was calculated by integrating the head acceleration raised to the power 2.5 (a ** 2.5) over the entire duration of the pulse with 1000 being the critical value. Early evaluation of the severity index was conducted by Hodgson et al. (8) who concluded that a good correlation existed between the severity index and degree of concussion in 29 stumptail monkeys subjected to impact, and with frontal bone fractures in cadaveric specimens. It was soon recognized that time duration of the head acceleration pulse affects the S.I., to the point that under a 1G environment, SI of 1000 is exceeded every 1000-sec. Therefore, limitation of the pulse duration for calculation of S.I. was important. Versace (9) proposed an alternative formulation of SI which subsequently was adopted in the FMVSS208 in 1972. The new formulation is commonly known as the Head Injury Criteria (HIC). Ever since its introduction, HIC has been controversial (3).

Within the deliberations of the International Standards Organization (ISO) working group on biomechanics, it was suggested that the critical value of HIC be raised to 1500 in frontal impact regulatory tests at 48 kph. The U.S. delegation rejected this proposal based on existing cadaveric data in which HIC's were measured along with skull fractures and brain injuries. The results of the U.S. delegation investigation were published in 1985 by Prasad and Mertz (10). The investigation also showed that human volunteers had undergone HIC exposures in the 1000 range with airbags without any brain injuries. These HIC durations were long (30-36ms) suggesting that limitations of the HIC duration were essential to evaluate the risk of head injuries. They suggested that calculation of HIC should be restricted to a maximum of 15ms. They also developed a head injury risk curve that correlated the possibility of head injury versus HIC. A HIC level around 1450 corresponded to a 50% risk of skull fracture, while a HIC level of 700 corresponded to a 5% risk of skull fracture. The head injury risk curve was further validated against injury producing head impact in High School football (11), Fig. (2)

Instead of limiting HIC durations to 15ms, NHTSA chose to adopt 36ms duration in the FMVSS208. However, Transport Canada adopted 15ms HIC duration, but set the HIC limit at 700.

As far as any theoretical basis for HIC is concerned, controversy continues. Many reasons are postulated, the main one being that HIC is based solely on the measurement of linear accelerations at the centre of gravity of the head. Obviously, angular accelerations also cause deformations in the brain. However, limits of angular accelerations have not yet been established for the human head. If the limit is near 13600 rad/sec^2 (12), such high angular accelerations are hardly ever seen in regulatory testing of restraint systems in frontal impacts, and may be the reason why angular acceleration limits are not specified in any regulation so far. It is currently believed that the effect of combined linear and angular acceleration on brain responses can be accounted for through detailed mathematical models of the human head.

Neck Injury Criteria: The European frontal impact regulation has adopted neck injury criteria proposed by Mertz (1) in 1984. Mertz's proposal consisted of specifying limits on peak extension and flexion moments at the occipital condyles of the dummy (head/neck junction) and time varying tension, compression and shear forces as measured by load cells in the Hybrid III dummy head. The peak extension and flexion moment limits were based on cadaver and human volunteer tests and accident reconstructions. The time varying tension forces were derived from accident reconstruction data using Volvo vehicles and the Hybrid III dummy for which real world injury data were available. The long duration end of the force-time curve (>45 msec) is based on muscle strength of a male volunteer. In terms of peak tensile force, the limit was suggested to be 3.3 kN which is in close agreement, to 3.1 kN, which was proposed by McElhaney and Myers (13) who based this conclusion on available biomechanical data.

Compressive tolerance of the neck proposed by Mertz were based on reconstruction of injury-producing high school American football "spearing maneuvers" against a mechanical tackling device with the Hybrid III dummy(14). It was an indirect way of determining dynamic axial compression tolerance limit. The peak tolerable load was 4.0 kN that dropped down to 1.1 kN if the duration of loading was 30 msec. The peak load of 4.0 kN compares to 4.8 kN + 1.3 kN mean axial load to failure in 6 cadaveric rotation constrained specimens reported-by McElhaney and Myers (13). It should be noted that Mertz's proposal called the neck injury tolerance levels as Injury Assessment Reference Values. Further, it was stated that being below these IARV's meant that significant neck injury, due to the loading condition considered, was unlikely. Subsequent biomechanical research by McElhaney and Myers (13) seem to agree with Mertz's proposal.

With further biomechanical research conducted since the introduction of the Hybrid III 50^{th}%ile male dummy and the development of a family of Hybrid III dummies, the determination of Injury Assessment Reference Values for these new dummies has taken on heightened importance. Changes to the FMVSS208 have been proposed to include child dummies, 5^{th}%ile and the 95^{th}%ile male dummies have been suggested by NHTSA (15). Scaling techniques have been used by NHTSA and the American Automobile Manufacturers Association to derive IARV's for various sized dummies and limit values in the regulatory tests have been suggested. Readers interested in following the development of these regulatory values are referred to the Docket of FMVSS208 (16).

The main departure from previous regulatory limits on peak neck responses is to consider the effects of combined loading of the neck as suggested by Prasad and Daniel in 1984 (17). It was shown that higher axial forces could be tolerated with an absence of bending moments in the neck, and higher bending moments were tolerable with an absence of axial forces in the neck. The development of combined loading criteria of the neck is being reexamined through reanalysis of combined data from two different sources - G.M. and Ford.

Chest Injury Criteria: The earliest chest injury criteria to enter a crash test regulation were specified in the FMVSS208 in the U.S.. Recognizing that the test device in the regulation was the Hybrid II dummy, which had a very stiff and non-biofidelic thorax design, chest acceleration was the only possible injury response that was meaningful to control. The chest acceleration limit placed in 30-mph rigid barrier tests was 60G's for 3msec-time duration.

This acceleration limit was based on voluntary exposures in early rocket-sled tests conducted by Col. J.P. Stapp. Stapp himself was exposed to 40G's for 100msecs without injury. Another subject had undergone 45G's for 44msec. It was assumed that for shorter durations higher accelerations could be tolerated. Considerable controversy ensued with the adoption of the acceleration based criteria. Mertz and Gadd (18) reported that there was no evidence that "even a 60G chest acceleration level would not be tolerable with an adequate restraint system" for pulse durations less than 100ms. They preferred to specify thoracic compression limit.

Early biomechanical basis was examined by Walfisch et al (19) who analyzed their cadaver database and concluded that "50G to 70G measured on a part 572 dummy seems to correspond to an acceptable level of injury for the restrained population exposed to risk of accident." However, chest acceleration was not generally accepted by the biomechanical community, and with the development of a more biofidelic dummy, the Hybrid III, chest deflection became the measurement of choice to be controlled in the Canadian and European regulations.

Substantial research was performed by G.M. in establishing impact response of the chest in dynamic blunt loading conditions. Analysis of pendulum impacts to cadaveric specimens- was conducted by Kroell, Nahum, Neathery and Mertz (20,21) who established 75mm sternal compression as being the 50% probability of AIS3 injury in blunt loading. Mertz (1,2) suggested this limit for airbag loading, but suggested 50mm sternum deflection as the limit for belt loading in the FMVSS208 tests.

Accident reconstruction studies were conducted by Mertz, Horsch, Melvin, Horn, and Viano (22,23) with Renault and Volvo restraint systems. These studies concluded that 50mm of sternal deflection produced by belt loading of the Hybrid III dummy corresponded to a 50% risk of AIS3+ thoracic injury for the population at risk in real world frontal crashes. The European and Canadian frontal impact regulations accepted this limit, but the U.S. FMVSS208 continued with 75mm of sternal deflection as the limit regardless of the restraint system utilized.

Whereas early biomechanical studies concentrated on sternal rib deflections as injury producing responses, studies utilizing anesthetized animals and human cadaveric specimens in early to mid-1980's showed that rate of compression was also an important injury producing agent. These studies were conducted by G.M. and the main researchers were Viano, Lau, and Kroell who published some of their results in 1986 (24,25) (see Figs. 3, 4). A chest injury criterion called the Viscous Criteria was proposed by Lau and Viano (25). This criteria explained heart, lung, liver soft tissue injuries. It also accounted for strain rate effects in soft tissues subjected to impact. The Viscous Criteria consisted of the product of deformation velocity (m/s) and the maximum normalized chest deflection. Numerical value of 1 for this product corresponds to a 20% risk of AIS $\geq$ 4 injury and seemed to be independent of animal species. A Viscous Criteria, now commonly referred as V*C, of 1 has been adopted in several regulations for frontal and side impacts, but has not been proposed by NHTSA even in the upcoming revised FMVSS208.

It should be mentioned that measurement of deflection in tests with cadaveric specimens had been problematic in the 1970's and 1980's. The development of a device that would yield deflection and shape of the thorax in impact conditions was pursued by NHTSA. One such device, now commonly known as the "Chest Band" was reported by Eppinger (26) in 1989. Several cadaveric tests in frontal and side impact conditions have since been conducted. This device has made it possible to monitor deflections and accelerations simultaneously in test specimens subjected to impact. As a result it is now possible to determine the efficacy of deflection based and acceleration based injury criteria. This paper will examine the implication of available biomechanical data for frontal impacts.

New Data: NHTSA has tested 71 cadavers in various restraint systems over several years of research to determine mechanisms of chest injuries and associated injury criteria. The assembled data was reported by Kuppa and Eppinger (27) in the 42^{nd} Stapp Car Crash Conference, and has formed the basis for a proposed Chest Trauma Index for the FMVSS208. The reported data was analyzed for consistency by the author of this paper. Three tests in the paper were found to be erroneous reporting as to the age of the cadavers and number of rib fractures sustained by these cadavers.

Further inconsistencies were identified in calculated maximum chest deflection for some of the tests conducted with two-point belt restraint system. Out of 14 tests utilizing the two point belt restraint system, 11 tests showed an average of 39% maximum normalized chest deflection with a standard deviation of 6.28. In the other three tests, the reported normalized chest deflections are 16%, 17%, and 17%. Considering that between 12 to 15 rib fractures were observed in these tests, and no other cadaveric data exist associating such a high number of rib fractures with such low chest deflections, the estimated chest deflection data are considered to be errors in measurement and, rejected from further analysis. Similar reasoning was used to reject two other data points associating 8% and 14% deflection with massive rib fractures. For those interested in analyzing the data, the following points were rejected: RC104, ASTS103, ASTS250, ASTS259, ASTS96, and ASTS97. Three additional test data reported by Kallieris et al (28) were added to the remaining data sets, resulting in 68 data points with known maximum normalized chest deflections and T- 1 accelerations.

It was further assumed that six (6) rib fractures in cadaveric specimen testing is acceptable injury level due to previous studies (29,30) indicating that cadaveric specimens show 2 to 3 rib fractures more than living specimens. Greater than seven (7) rib fractures were assumed to be AIS3+ chest injuries. Fig 5 shows all the data for which chest deflection and T-l accelerations are known. It can be seen that only one cadaver below 30% maximum normalized chest deflection had an AIS3 injury to the chest. The age of this cadaver was 76 years. It had undergone 26% maximum normalized chest deflection resulting in 9 rib fractures. All other AIS3+ injuries were above 30% maximum normalized chest deflection regardless of T-l acceleration. Of the twenty cases of AIS O-2 injuries, 6 are at or above 30% maximum normalized chest defl ection. The AIS O-2 cases range from 15 to 78 G's T- 1 accelerations. No clear-cut demarcation between AIS-2 and AIS3+ injuries is noticed along the acceleration axis.

Based on the above observations, maximum normalized chest deflection appears to be a strong predictor of chest injuries, and maximum T-l acceleration is not.

Since the current dummy measures only central chest deflection in the sternum, regression analysis was used to predict maximum normalized chest deflection at the sternum in the cadaver tests – similar to that done by Kuppa and Eppinger. It was found that 30% maximum normalized chest deflection corresponds to 52mm of internal sternum deflection. Logit analysis of the data indicated that the above corresponded to 50% probability of AIS3+ chest injury for the average age of 58 years of the cadaver population tested.

An effort to extract age effects was made by dividing the tested cadaver population into two groups. One group consisted of all cadavers below 58 years, average age 47.5 years, and the other group consisted of all cadavers 59+ years, average age 65.5 years. It was further assumed that cadavers that did not fail in the 59 + group would not have failed if they were younger. And, the cadavers in the less than 58-year group that had failed would also have failed if they were older. The results of logit analysis showed that for average age of 47.5

years, the 50% probability of AIS3+ chest injuries would correspond to 58.3mm of internal sternum deflection. Similarly, the 50% probability of AIS3+ chest injuries would correspond to 45mm of internal sternum deflection for average age 65.5 years.

Since the Hybrid III dummy continues to be the only dummy for frontal impact regulations, the above discussions relating chest deflection to chest injuries is applicable in the worldwide regulations. However, continued usage of the two vastly different dummies for side impact regulations needs examination of the different injury criteria for side impact regulations in U.S. and the rest of the world.

Chest injury criteria for side impact: The U.S. side impact regulation, FMVSS2 14, utilizes the SID dummy and the Thoracic Trauma Index as the injury criteria. It has been mentioned earlier that thoracic deflection measurements were problematic in the 70's and the 80's. As a result, early biomechanical testing of animal or human cadaveric specimen utilized acceleration measurements at the thoracic spine and the rib cage with a twelve accelerometer array reported by Robbins et al. (31). It was also assumed that a dummy designed to meet biofidelity corridors based on accelerations would have adequate biofidelity for side impact testing. Such a dummy, the SID, was designed and subsequently selected in the FMVSS214. Statistical correlation between acceleration responses of T-12 and the ribs were attempted to predict AIS4+ injuries. Eppinger (32) found that the average of Peak T-12 acceleration and peak rib acceleration correlated with chest injuries. Additionally age was a major contributor to injuries and was included in the formulation of the Thoracic Trauma Index (TTI). Fig. 6 shows predicted injury risk curves for dummy TTI responses. It should be noted that acceleration based criteria was the only meaningful criteria use with this dummy since the biofidelity in deflection was doubtful and not measured in the cadaver tests.

Considerable controversy continues with the use of TTI. To further examine the validity of TTI or other injury measures twenty-six cadaver tests using "chest bands" and accelerations have been conducted and reported by Pintar et al (33). Analysis of the data indicated that normalized chest deflections, V*C and the TTI have merit as injury criteria. However, a combination of TTI and maximum normalized chest deflection was the best statistical predictor of chest injuries in lateral impacts. Obviously, there is then a need for a test device that has biofidelity for both acceleration and deflection responses.

The current European chest injury criterion is based solely on maximum deflection of the thorax. The supporting biomechanical data came from drop tests conducted at APR laboratories and reported by Stalhaker et al. (34). Interestingly, the authors found that 30% normalized compression of the thorax based on whole chest width was the tolerance for AIS3 or less. This is very similar to what was discussed earlier for frontal impacts.

Femur Injury Criteria: The earliest proposal for limits on axial compression of the femur was based on sled tests with cadavers conducted by Patrick, Kroell and Mertz in 1966. This evaluation of cadaver test data suggested a conservative threshold level of 6.23 kN for protection of the femur and hip complex. Further experiments with cadaveric specimens using various mass impactors were conducted by Powell et al (35) and Melvin et al (36) in 1974 and 1975. When all these tests were analyzed, it became clear that tolerable compressive loads depended on the duration of loading. Viano and Khalil (37) utilized a finite element model of the femur to show that time duration of impact was an important parameter, Fig. 7. The FMVSS208 adopted a peak compressive force of 10kN as the injury criteria without any time duration limitation. This level continues in all frontal impact regulations worldwide.

Tibia Injury Criteria: The tibia injury criteria currently adopted in European frontal impact regulations is based on research conducted by Mertz who suggested the Tibia Index and

tolerable axial compressive loads in the tibia. The origin of the currently use 8kN maximum compressive load is based on accident reconstruction of airbag crashes in which fractures of tibia1 condyles were observed. These injuries were independent of crash severity and were attributed to airbag/knee interactions. Mertz noted that if the load on the medial or lateral condyles exceeded 4kN fractures occurred. Since the maximum load per condyle could be only 4kN, the maximum allowable tibia compressive force could not exceed 8kN.
For the fracture of the tibia1 shaft Mertz's suggestion was to use the Tibia Index as defined below:

$$TI = \frac{M}{M_c} + \frac{F}{F_c} \leq 1 \quad \text{where} \quad \begin{array}{l} M_C = 225 \text{ N-m} \\ F_C = 35.9 \text{ kN} \end{array}$$

The critical values for compression and bending were based on earlier strength data from Yamada (38). Since, the maximum force could be only 8kN, the Tibia Index acted as a control on the maximum bending moment. Further examination of bending moment to failure data published by Nyquist (39) indicated that the critical moment of 225 Nm was too conservative for a 50^{th}%ile male. The average failure bending moment for a 50^{th} %ile male was closer to 300 Nm. As a result, the acceptable Tibia Index adopted in the European regulation was raised to 1.3.

Summary and Conclusions

An attempt at examining the biomechanical basis of injury criteria currently utilized in regulations worldwide for frontal and side impacts has been made. This historical review, although not an exhaustive one, shows that most of the injury criteria utilized in regulations today have a biomechanical basis in animal testing, human cadaveric testing, and accident reconstruction. Although most of the research upon which the regulatory criteria are currently based were conducted before the mid-eighties, subsequent research have further validated the choice of the injury criteria. It should be mentioned that the test device with its associated biofidelity of impact response has led to the adoption of injury criteria suitable of the particular device.

A substantial amount of pertinent research has followed the introduction of regulations and the biomechanical knowledge base has increased substantially. More advanced and refined test devices and measurement capabilities are under development. It is hoped that in future regulatory activities a synthesis of all existing data will be carried out and refined injury criteria will be adapted. Biomechanical researchers and regulators should note the large time lag between research and regulations.

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Fig. 1- Experimental and Theoretical Tolerance Curves

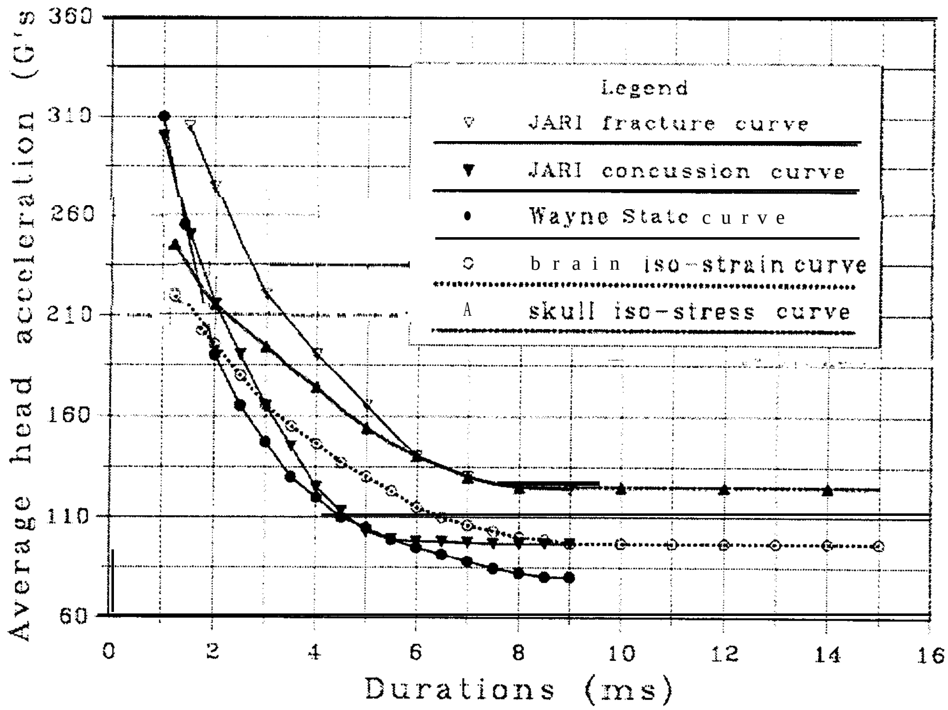

Fig. 2 - reconstruction of Football Fatalities

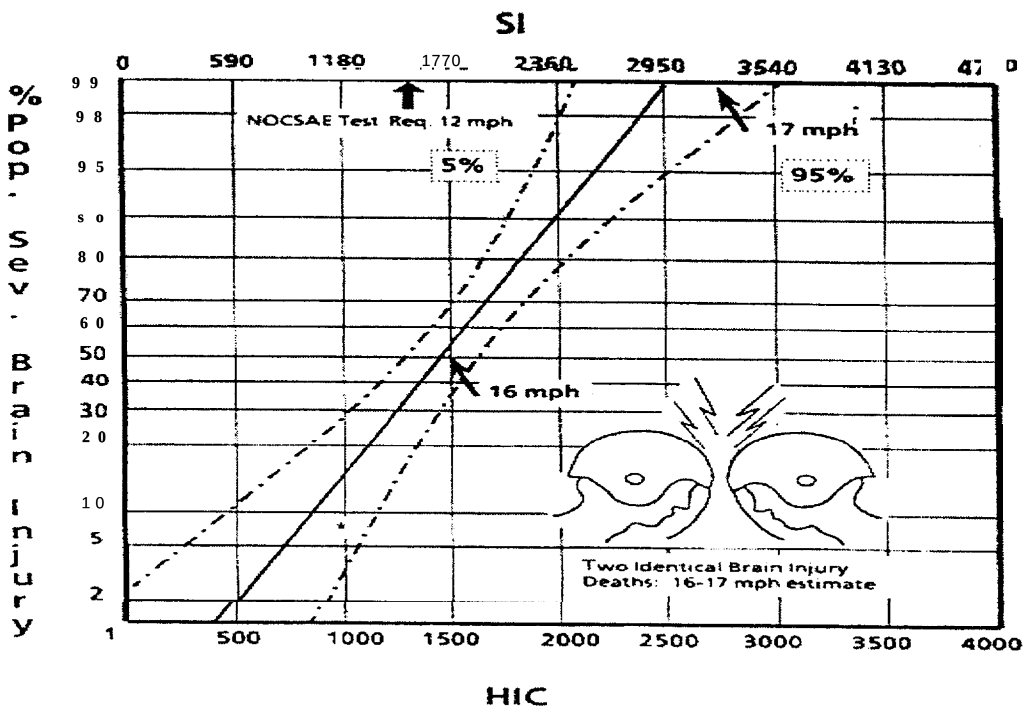

Fig. 3 – Chest Injury Mechanisms (Lau and Viano)

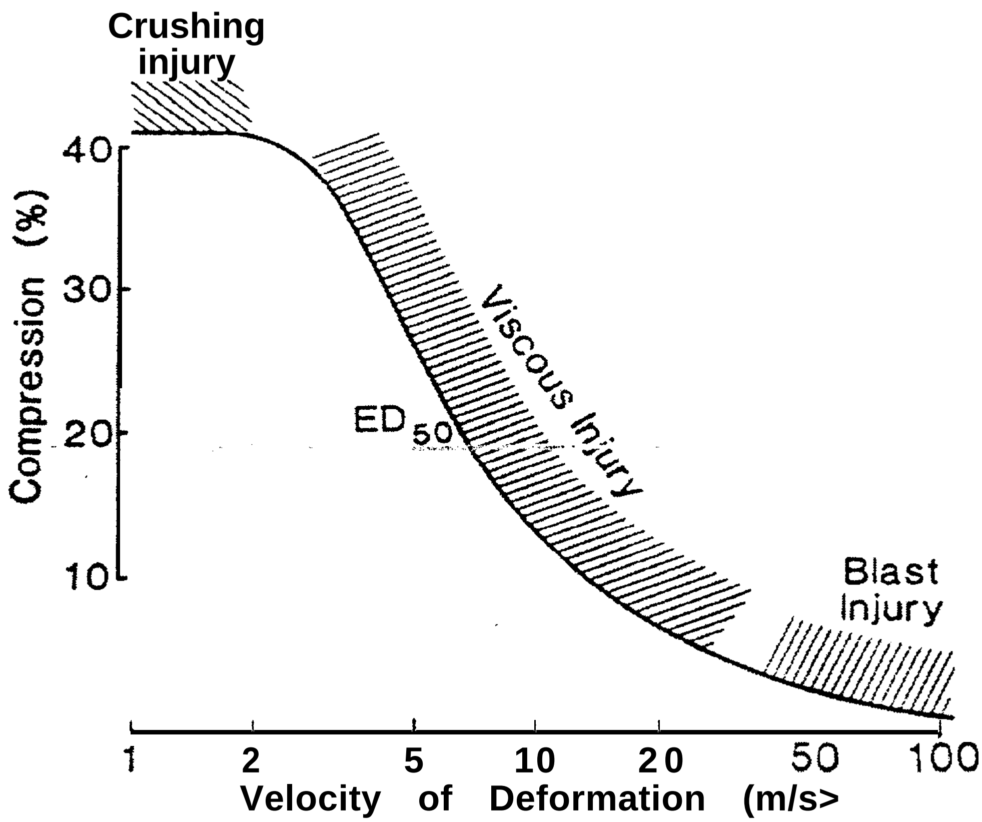

Fig. 4 - Risk of Injury Versus Viscous Criteria (Lau and Viano)

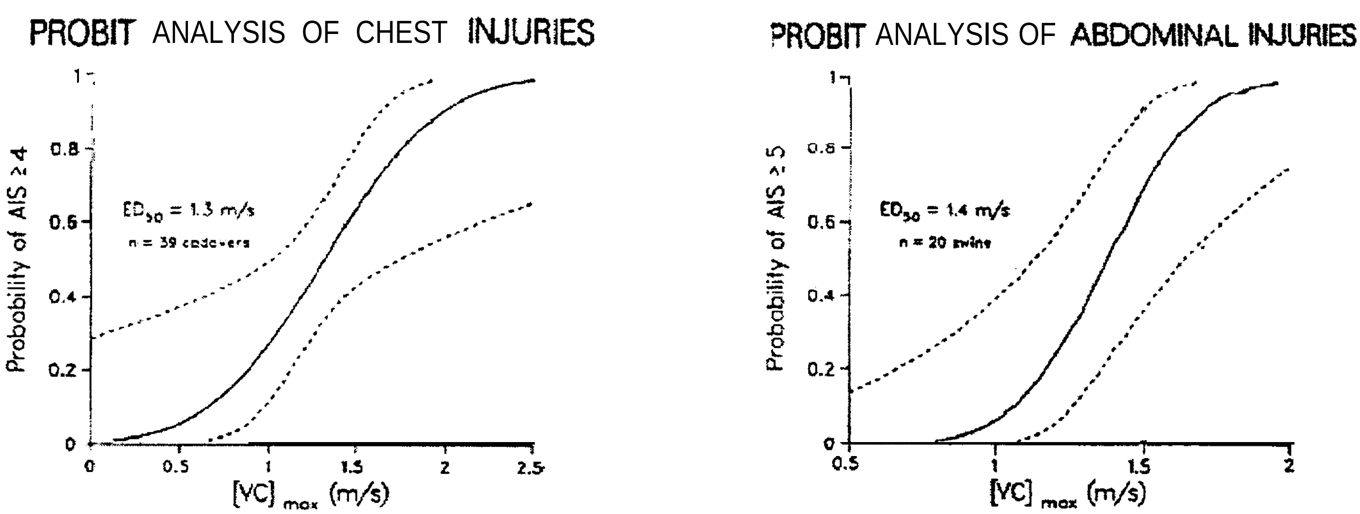

Fig. 5 - Normalized Deflection and T- 1 Acceleration for Cadaveric Specimens

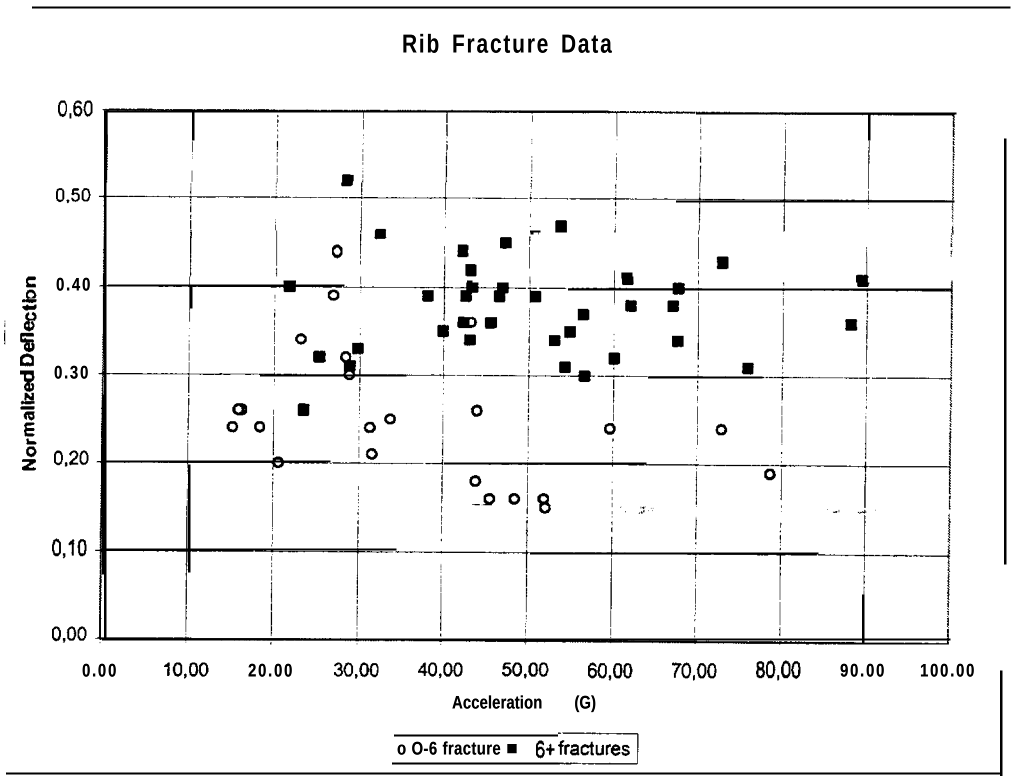

Fig. 6 - Predicted Thoracic Risk Curves Versus Dummy TTI Responses
(Hackney et al. 3 1"' Stapp Car Crash Conf.)

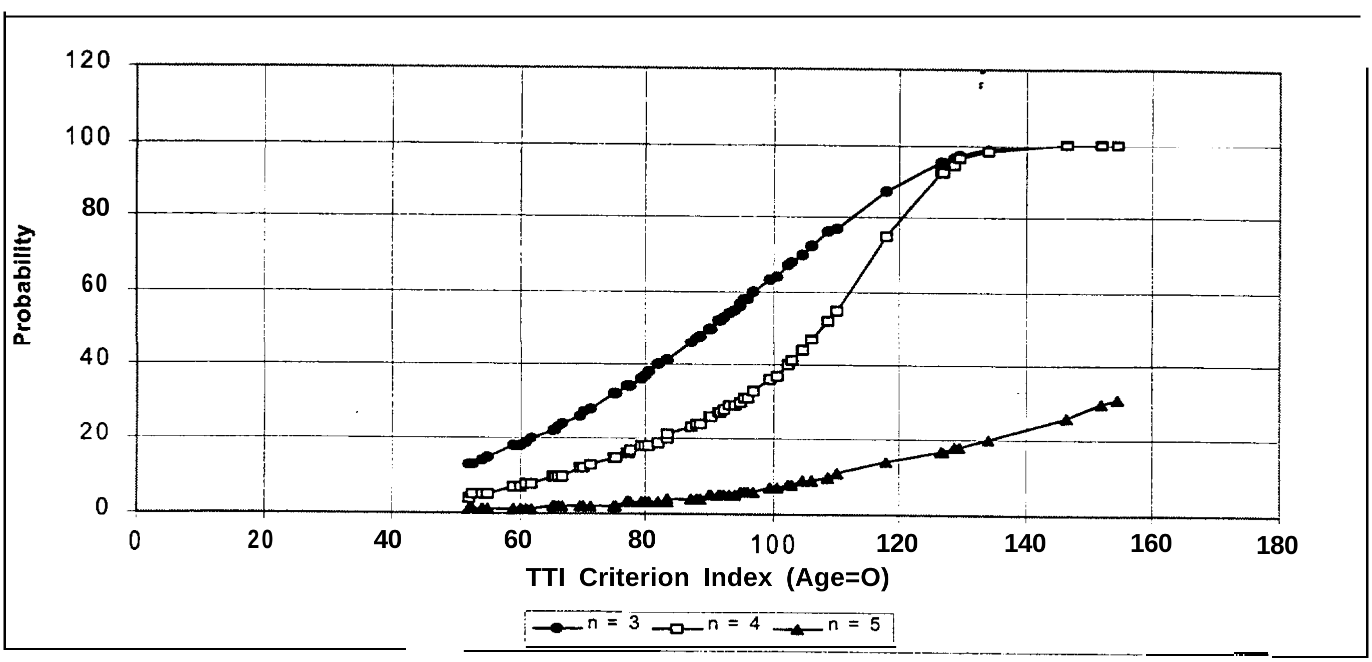

Fig. 7 - Theoretical and Experimental Femur Fracture Data
(SAE SD-4 12. 1976)