REVIEW OF HEAD INJURY TOLERANCE TO DIRECT IMPACT

By

Verne L. Roberts and James H. McElhaney
Highway Safety Research Institute
The University of Michigan
Ann Arbor, Michigan U.S.A. 48105

ABSTRACT

The human tolerance literature pertaining to direct head impact is reviewed and discussed. The basis for the Vienna Index, the Effective Displacement Index, the Severity Index, the Head Injury Criteria of USMVSS #208, the Wayne State Head Injury Tolerance Curve and the Maximum Strain Criteria will be examined and the correlation of these various indices with the available data demonstrated. Additionally, the prediction of injury for various impact pulses when used with the variety of available criteria will be determined.

Where appropriate, new data will be introduced to complete the existing data set. It is felt that on the basis of the currently available information regarding the limits of head impact tolerance that tolerance can be established for impacts of arbitrary direction.

INTRODUCTION

With the advent of high speed air and land transportation, engineers have become increasingly aware of the mechanical frangibility of the human body. Thus, we have seen the evolution of various isolating and load distributing devices ranging from seat belts and padded sun visors, to ejection seats, crash helmets, and acceleration couches. While there is a large amount of information available regarding the response of inanimate systems to vibration and impact, there is a comparable dearth of knowledge pertaining to the mechanical responses of biological systems. Therefore, the design of much supporting and protective equipment is often based on intuition because of the lack of information available about the mechanical behavior of the human body. In addition, such knowledge would be helpful in the treatment of injury by serving to identify the mechanism of trauma. Thus, both a rational design procedure for impact protection and a rational therapy for treatment of trauma cannot be developed until a quantitative description of the mechanical responses of the human body is obtained.

In order to properly design devices aimed at minimizing head injury in the automotive crash environment, engineers require a means of predicting potential injury or a so-called Head Injury Criteria. This criteria might be used in real accident reconstructions, car crash and sled test experiments or mathematical simulations.

The automotive crash environment encompasses a wide range of impulse durations and directions. Thus, a viable head injury criteria must provide appropriate mechanisms that realistically account for the frequently observed, but poorly documented, relations of head impact tolerance and impulse duration and direction. In addition, two distinct types of loading are observed.

1. An impact or blow involving a collision of the head with another solid object at an appreciable velocity. This situation is generally characterized by large linear accelerations and small angular accelerations during the impact phase.

2. An impulsive loading including a sudden head motion without direct contact. The load is generally transmitted through the head-neck junction upon sudden changes in the motion of the torso and is associated with large angular accelerations of the head.

It has been shown (1) that in the moderate but survivable automotive crash environment (30 mph barrier equivalent) no significant head injuries occur when the fully-belted occupant rides the crash down without head-to-vehicle contact. This does not mean that the Type Two loading described above can not produce injury, but only that the levels of linear and angular accleration required to produce head injury without contact do not occur in moderate crashes. If,however, vehicles are stiffened to provide less compartment intrusion at higher velocities, injuries of the type described by Ommaya and Hirsch (2) might become commonplace.

Thus a rational head injury criteria for current automotive design may be concentrated on the first type of loading. There is, of course, a possible defect in this rationale that concerns the possible increased potential for head injury involved with a combination of the first and second type of head loading. An attempt to reconcile this problem has lead to the development of the Mean Strain Criterion. This head injury criteria considers the total linear acceleration history of the head but assumes a single injury mechanism.

MEAN STRAIN CRITERION (MSC)

The dynamic structural characteristics of monkey skull and brain were determined over a wide frequency range by Stalnaker and McElhaney in 1971 (3). These results, reported as the change of mechanical impedance (force/velocity) with frequency, allowed the conceptual characterization of the head as two masses coupled by a spring and dashpot. The mathematically predicted dynamic response of the model agreed well with the experimental data.

Experimental impacts delivered to the heads of various size primates showed that the dynamic model postulated on the basis of vibration studies accurately predicted head-impact injuries. It was further found that for head impacts of a known magnitude, the resulting injuries could be grouped by comparing the mean strain as predicted by the theoretical model with injury levels, (where mean strain is defined as the displacement of one side of the head relative to the other, divided by the distance across the cranium). This experimentally derived head injury data for living primates formed the basis for a Mean Strain Criterion (MSC) for head injury to humans (4). Using the value of predicted strain in the Rhesus monkey head as a criterion of injury, a tolerance curve was derived with related average acceleration and time for a constant level of mean strain.

The derived tolerance curve for the subhuman primate was validated by plotting the experimental data points necessary to produce minor, but identifiable, brain injury in the living test subject for a wide range of pulse durations (Figure 1).

It was determined that the heads of several species of subhuman primates, squirrel monkeys, Rhesus monkeys, the chimpanzee and the fresh human cadaver had mechanical impedance characteristics over a broad frequency band (5 to 5000 Hertz) which were similar in shape but varied in the mechanical characteristics of mass, stiffness and damping. Using the maximum predicted strain as the basis for injury, experimental head impacts in the laboratory documented the validity of the theory and formed the basis for establishing modeling relationships through which extrapolation of the MSC to other size heads may be made. This approach was tested by comparing the MSC to human volunteer and fresh human cadaver head impacts (Figure 2). The mean tolerable head strain for humans (0.006 in./in.) was calculated from the mathematical model and the scaling technique referenced above.

A study was undertaken to document the validity of the MSC for arbitrarily directed head impacts. Thirty carefully selected Rhesus monkeys were impacted at increasing levels in various directions (front,side,back,top and mid-front). The impact level was increased until autopsy studies indicated that an Estimated Severity of Injury (ESI) of a moderate but reversible type (Class 3) was obtained. Only closed brain injuries were considered.

The mechanical impedance was determined for five monkeys of approximately the same weight as the ones used in the impact study. These impedance curves were obtained for the top, side, rear, and front of the head (Figure 3).

The values of tolerable acceleration and impedance data were then input to the MSC model. The predicted mean strain values for each direction varied less than 7.5%. This indicates that, while widely varying accelerations are required to produce an injury level of 3 in the Rhesus monkey, the corresponding strains are approximately equal. The results of this study are given in the form of an acceleration surface for a constant strain level of 0.032 in./in. for Rhesus monkeys subjected to rigid striker impacts (Figure 4).

Preliminary studies on the fresh intact cadaver indicate that a tolerable mean strain level of 0.0061 as predicted by the MSC model may be used for arbitrary impact directions with model constants appropriate for that direction. However, a sufficient number of impact tests and driving point impedance measurements have not yet been made to verify a direct extrapolation of the Rhesus monkey data. Figure 5 shows predictions and measured values of the MSC for human head impact in the sagittal plane.

COMPARISON OF HEAD INJURY CRITERIA

The preceding discussion involved the most recent work on the Mean Strain Criterion for head impact. The following section contains the results of a series of analysis aimed at comparing various head injury criteria.

Values for various head injury indices were computed for two crash simulations.

1. Dummy resultant head accelerations in frontal automotive crash environments were used. The dummy was unbelted, and struck the windshield a clean blow. The windshield was not penetrated except a small tear in the laminate was allowed (Class A), or several small tears (Class B).

2. Resultant head accelerations from a recent series of high speed human volunteer and dummy tests at Holloman Air Force Base using airbag restraints.

Wayne State Tolerance Curve (WSTC)

The Wayne State Tolerance Curve was introduced by Lissner in 1960 (5). Originally this curve was developed from data obtained by dropping embalmed cadaver heads on to unyielding flat surfaces. Linear skull fracture was used as the criterion of injury. In 1962 Gurdjian published the Wayne State Tolerance Curve as it appears today (6). This curve was developed by combining a wide variety of pulse shapes, animal types and injury mechanisms. The failure criteria used was generally skull fracture and/or concussion, except for long pulse duration from human volunteers with no discernible injury. In 1965 Patrick et al (7) proposed that the original horizontal asymptote of 45 G's be raised to 80 G's to adjust for additional data from tests against yielding surfaces. Since that time numerous papers have been written providing supporting data for the tolerance curve (8,9,10,11,12,13).

The injury assessment is based on the average acceleration and pulse duration. A given average acceleration at a particular pulse duration which lies below the WSTC is considered to cause at most cerebral concussion without permanent after-effects, while any point which lies above the curve is considered to be dangerous to life. For single head impacts into a rigid flat surface the average acceleration and pulse duration is quite easy to determine, but for sled testing where multiple impacts are quite common the "effective" pulse, that is, the part of the pulse upon which the average is based, is not well defined. In spite of the many interpretive difficulties associated with this curve, it has been the principal source for head injury tolerance information used by the automotive safety community.

Gadd Severity Index (GSI)

Because of various interpretive difficulties associated with the use of the WST curve, C. W. Gadd introduced the Gadd Severity Index as a generalization of the Wayne curve (14,15,16). More recently the GSI has been extended for long pulse duration by means of the Eiband tolerance data and other primate sled runs. The severity index equation has the following form:

$$\text{G.S.I.} = \int_0^\tau a^n dt \qquad (1)$$

where

- a = head acceleration response function
- n = weight factor, general 2.5
- τ = pulse duration
- t = integral parameter of time

The head injury threshold severity index number was determined from comparison with the WST curve and the number 1000 recommended.

The assessment of injury hazard is obtained performing the above calculation. If this number (GSI) is greater than 1000, the acceleration pulse is considered to be dangerous to life. If this index is less than 1000, the acceleration pulse is then considered not to be life threatening.

Head Injury Criterion (HIC)

The Head Injury Criterion was first proposed by J. Versace (17) and then modified by NHTSA (18). This criterion is based on a new interpretation of the Gadd Severity Index.

Versace pointed out that because the WST curve was plotted for average acceleration, any comparison to the WST curve should be made using the average acceleration of the pulse of interest.

The question of long pulse head accelerations has posed some problems when using the S.I. to predict head injuries. In order to provide a better comparison with human volunteer tests, a head criteria has been proposed as

$$HIC = \left[\frac{\int_{t_1}^{t_2} a\,dt}{t_2 - t_1}\right]^{2.5} (t_2 - t_1) \qquad (2)$$

where

t_1 = an arbitrary time in the pulse
t_2 = for a give t_1, a time in the pulse which maximizes the HIC
a = resultant acceleration at the head center of gravity

If this index is less than 1000, the situation is considered not to be life-threatening.

Vienna Institute Index (JTI)

The Vienna Institute Index was introduced by A. Slattenschek (19) and is based on a single degree-of-freedom vibration model.

With the damping assumed to be critical $\beta = 1$, and two triangle acceleration pulses determined from the WST curve, the natural angular frequency value of ω = 635 rad/sec and a maximum tolerance displacement $x_{To\omega}$ = 0.092 (2.35mm) inches was obtained from the following equation:

$$\ddot{x} + 2\beta_{\omega}\dot{x} + \omega^2 x = \ddot{y}(t) \qquad (3)$$

where

x = relative displacement of brain mass to skull
$\dot{x}, \ddot{x}$ = relative velocity and relative acceleration
ω = natural angular frequency of vibration 635 rad/sec
β = viscous damping coefficient of 1.0
$\ddot{y}(t)$ = acceleration pulse measured at the head

The maximum deviation between the model and the WST curve is -4 percent. To access an impact, the amplitude x_{max} corresponding to the acceleration pulse to be analyzed is determined from the model and compared to the tolerable amplitude x_{Tolr} = 0.092 inches. A J tolerance index is then defined by:

$$J = \frac{x_{max}}{x_{Tolr}} \qquad (4)$$

where:

x_{max} = maximum x generated by the model for a given acceleration pulse

x_{Tolr} = tolerable amplitude from the Wayne State Tolerance Curve 0.092 inches (2.35mm)

According to Slattenschek, impacts with a J tolerance of $J_2 = 1$ just reach the threshold of human tolerance; values $J < 1$ at worst cause cerebral concussion without permanent after-effects, while values $J > 1$ are considered to be hazardous to life (20).

Effective Displacement Index (EDI)

The effective displacement index was introduced by J. Brinn (21) and is similar to the Vienna Institute model with changed damping and the angular frequency. New angular frequency and damping values for the Slattenschek model were determined by matching the model to the WST curve. The emphasis was placed on matching for short duration events (3-5msec pulse duration). The best fit of the model to this portion of the WST curve was found by using the model parameters ω 482 rad/sec and $\beta = 0.707$. These model parameters and the Slattenschek model were exercised for points on the WST curve and a displacement value of 0.15 inches was determined as a "Design Bogie" for human AP head impacts. Because of the unhuman like response of dummy heads, the Design Bogie is raised to a dummy Test Limit of 0.17 inches in the AP direction. When the resultant acceleration is used, a Design Bogie of 0.18 inches and a dummy Test Limit of 0.20 inches is recommended.

Revised Brain Model (RBM)

The revised brain model was introduced by W.R.S.Fan (22) and is a modification of the Vienna Institute model. Like the JTI, the RBM is a single degree-of-freedom mass-spring-dashpot model of the brain. The viscous damping coefficient for this model was estimated from published values of brain material properties. With an estimated damping coefficient of 0.4 data from the WSTC for long duration inputs, a natural angular frequency of 175 rad/sec and the theoretical tolerable brain deformation (S_d) of 1.25 inches was estimated. A tolerable brain velocity S_v was then calculated from the WST curve for short pulse duration and was found to be 135.3 in/sec.

The recommended measure of brain injury potential is $\dot{x} < S_v$ for impact pulse durations less than 20 msec and $x < S_d$ for pulse durations greater than 20 msec, as calculated from the differential equation of the Slattenschek model with revised coefficients.

RESULTS OF COMPARISON

The results of the computations of the various head injury criteria are presented in Table 2. Figure 6 shows all of the models and their constants for purposes of comparison while Figure 7 gives normalized values of the head injury indices obtained by dividing the particular computed index by the appropriate cut off value (i.e. 1500 for the SI and 1000 for the HIC).

Based on accident statistics, it is felt that head impacts of this type into the HPR windshield seldom involve serious head injury and an appropriate head injury criteria should so indicate. Study of Table 2 shows the GSI to be quite conservative in this situation. The HIC is less conservative, but still indicates four life-threatening situations in the windshield tests. The RBM, EDI and MSC all predict essentially the same injury levels for both series.

Mathematical Modeling

While the previous authors have directed their efforts primary toward the acquisition of experimental data and its explanation in terms of either lumped parameter models or mathematical correlates to published head injury data, others have concerned themselves with mathematical models of the head which define its response to impact. Both approaches can yield insight into head injury tolerance with the former providing near term tools while the later, because of their greater attention to modeling the entire system hold great hope for future evaluation of protective devices without the extensive laboratory work required at this time.

Anzelius (23) modeled the head as a fluid filled rigid shell brought suddenly to rest while Guttinger (24) solved essentially the same problem with the shell achieving a velocity from rest. Further attempts to model the head during impact lay dormant until proposed by Goldsmith (25) in his review of the physical processes of head injury. Engin (26) solved the case concerning the asymmetric response of a spherical shell filled with an inviscid fluid subjected to a local radial impulsive load. Recently Liu et al (27) extended Engin's work for the axisymetric solution for a Dirac-delta time function to the case for a finite time function. Chan (28) has developed the theory further in his study of the asymmetric response of a fluid filled shell. Benedict et al (29) solved a problem similar to that of Engin's allowing for only membrane effects while Hickling and Wenner (30) modeled the head as a two layered viscoelastic sphere subjected to axisymmetric impact. Kenner and Goldsmith (31) experimentally investigated the problem solved by the various analytical studies.

Another series of investigators have modeled the response of the head to impact for the case where the pulse duration is long relative to the transit time of a pressure wave through the container. Gross (32) in 1958 developed a glass, fluid filled model and compared the results of cavitation induced in the fluid with his analysis. Unterharnscheidt and Sellier (33) also performed extensive studies of this nature utilizing analytical and experimental models extending their work to animals subjects to confirm their analysis. Concurrently Lindgren (34) developed an extensive study of mechanical inputs to a model of the head when struck from different directions with varying boundary conditions. Kopecky and Ripperger (35) extended the modeling treatment to include a deformable fluid filled cylinder and confirmed the previous results (12) indicating that the location of the nodal point is a function of the container deformation.

DISCUSSION

The driving point impedance studies indicate that the mechanical response of the primate head may be approximated as a two-mass system. Injury levels for blunt impacts have been related to the compression of a spring in the simple two-mass model. The MSC model indicates decreasing tolerance with impulse duration for pulses of approximately 10 times the resonance period or less. For pulses longer than 10 times the resonance period, a quasi-static response is indicated that is unaffected by further increases in pulse duration. The prediction is quite similar to that produced by EDI, JTI and the RBM, but considerably different from the SI and HIC, which indicate that a quasi-static response is never obtained. Obviously, as with all simple models of complex phenomena, extrapolation of model predictions beyond the range of validation or to new situations is dangerous and should be done with caution. The MSC model has been developed for blunt impacts where the amount of bone and scalp in contact with the impactor approximates that loaded by the coupling clamp during the impedance tests. When the loads are applied to large sections of the head or through the neck, many of the arguments used in the model development do not apply. In addition, it is probable that the injury mechanisms change considerably with these different types of loading, and a single mechanism model, as are all the ones discussed in this paper, would be inadequate.

No treatment of this subject would be complete without acknowledging the numerous studies which point out the combination of linear and angular motion of the brain substance in contributing to the head injury process. However the existence in this symposium of a separate paper on head injury in the absence of impact will undoubtedly cover this material in its fully expanded form.

REFERENCES

1) McElhaney, J. H., V. L. Roberts and J. W. Melvin, "Biomechanics of Seat Belt Design," Proc. 16th Stapp Conf., 1972.

2) Ommaya, A. K. and A. E. Hirsch, "Protection of the Brain From Injury During Impact: Experimental Studies in the Biomechanics of Head Injury," AGARD Conference Pre-Print No. 88 on Linear Acceleration (Impact Type), 1969.

3) Stalnaker, R. L., J. L. Fogle and J. H. McElhaney, "Driving Point Impedance Characteristics of the Head," J. of Biomechanics, Vol.4,No.2, pp. 127-139, March 1971.

4) McElhaney, J. H., R. L. Stalnaker, V. L. Roberts and R. G. Snyder, "Door Crashworthiness Criteria," Proc. 15th Stapp Conf., Paper 710864, p.39, New York: Society of Automotive Engineers, Inc., 1971.

5) Lissner, H. R., M. Lebow and F. G. Evans, "Experimental Studies on the Relation Between Acceleration and Intracranial Pressure Changes in Man," Surgery, Gynecology and Obstetrics,111:329-338, 1960.

6) Gurdjian, E. S., H. R. Lissner and L. M. Patrick, "Protection of the Head and Neck in Sports," Journal of American Medical Association 182 pp. 509-512, November 1962.

7) Patrick, L. M., H. R. Lissner and E. S. Gurdjian, "Survival by Design - Head Protection," Proc. 7th Stapp Conf., Society of Automotive Engineers, Inc., New York, pp. 483-499, 1965.

8) Gurdjian, E. S., V. R. Hodgson, L. M. Thomas and L. M. Patrick, "Significance of Relative Movements of Scalp, Skull, and Intracranial Contents During Impact Injury of the Head," Journal of Neurosurgery, Vol. 29, pp. 70-72, 1968.

9) Hayashi, T., "Study of Intracranial Pressure Caused by Head Impact," (2nd Report), Journal of the Faculty of Engineering, University of Tokyo, Vol. XXX, No.2, 1969.

10) Hodgson, V. R. and L. M. Patrick, "Dynamic Response of the Human Cadaver Head Compared to a Simple Mathematical Model," Proc. 12th Stapp Conf., 1968, Society of Automotive Engineers, Inc., New York, pp.280-301, 1968.

11) Hodgson, V. R. and L. M. Thomas, "Testing the Validity and Limitations of the Severity Index," Proc. 14th Stapp Conf., Nov. 17-18, 1970, Paper 700901, Society of Automotive Engineers, Inc., New York, pp.169-187,1970.

12) Roberts, V. L., V. R. Hodgson and L. M. Thomas, "Fluid Pressure Gradients Caused by Impact to the Human Skull," Proceedings of the Human Factors Conference, ASME, Paper No. 66-HUF-1, 1966.

13) Hodgson, V. R. and L. M. Thomas, "Comparison of Head Acceleration Injury Indices in Cadaver Skull Fracture," Proc. 15th Stapp Conf. pp.190-206, 1972.

14) Gadd, C. W., "Criteria for Injury Potential," in, Impact Acceleration Stress. A symposium held at Brooks Air Force Base, Texas, 27-29 November 1961. National Academy of Science/National Research Council (1962). Publication 977.

15) Gadd, C. W., "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard," Proc. 10th Stapp. Conf., University of Minnesota, Minneapolis, pp.164-174. Society of Automotive Engineers, Inc., New York, SAE Paper No. 660793, 1966.

16) Gadd, C. W. and L. M. Patrick, "System Versus Laboratory Impact Tests for Estimating Injury Hazard," Society of Automotive Engineers, Inc., New York. SAE Paper, No. 680053, 1968.

17) Versace, J., "A Review of the Severity Index," Ford Technical Report, No. S-71-43, November 12, 1971.

18) Department of Transportation National Highway Traffic Safety Administration (49 CFR Part 571) [Docket No. 69-7; Notice 17], "Occupant Crash Protection Head Injury Criterion."

19) Slattenschek, A. and W. Tauffkirchen, "Criterical Evaluation of Assessment Methods for Head Impact Applied in Appraisal of Brain Injury Hazard, In Particular in Head Impact on Windshields," International Automobile Safety Conference Compendium, 1970, Paper 700426, Society of Automotive Engineers, Inc., New York, pp.280-301, 1968.

20) Slattenschek, A., W. Tauffkirchen and G. Benedikter, "The Quantification of Internal Head Injury by Means of the Phantom Head and the Impact Assessment Methods," Proc. 15th Stapp Conf., pp.742-766,1972.

21) Brinn, J. and S. E. Staffeld, "Evaluation of Impact Test Accelerations: A Damage Index for the Head and Torso," Proc. 14th Stapp Conf., November 17-18, 1970, Paper 700902, Society of Automotive Engineers,Inc., New York, pp. 188-202, 1970.

22) Fan, W. R. S., "Internal Head Injury Assessment," Proc. 15th Stapp Conf., pp. 645-665, 1972.

23) Anzelius, A., "The Effect of an Impact on a Spherical Liquid Mass," Acta. Path. Microbiol. Scand. Suppl. 48, pp.153-159, 1943.

24) Guttinger, W., "Der Stosseffekt aufeine Flussigkeitskugel als Grundlage einer Physikalischen Theorie der Entstehung von Gehirnverletzungen. Z. Naturf. A5, pp.622-628, 1950.

25) Goldsmith, W., "The Physical Processes Producing Head Injury," Proc. Head Injury Conf., Lippincott, pp. 350-382, 1966.

26) Engin, A. E., "The Axisymmetric Response of a Fluid-Filled Spherical Shell to a Local Radial Impulse-A Model for Head Injury," J. of Biomechanics, Vol.2,No.3, pp.325-341, 1969.

27) Liu, Y. K., H. S. Chan and J. A. Nelson, "Intracranial Pressure Wave Propagation in Head Impact," Proc. Summer Computer Sim. Conf., pp.984-994, 1971.

28) Chan, H. S., "The Asymmetric Response of a Fluid-filled Spherical Shell-A Mathematical Simulation of Head Injury," Ph.D. Dissertation, Tulane U., New Orleans, La., 1971.

29) Benedict, J. V., Harris, E. H. and D. H. Von Rosenberg, "An Analytical Investigation of the Cavitation Hypothesis of Brain Damage," J. of Basic Engr., Vol.92, No.3, pp. 597-603, 1970.

30) Hickling, Robert and M. L. Wenner, "Mathematical Model of a Head Subjected to an Axisymmetric Impact," J.of Biomechanics,Vol.6,No.2,pp.115-132, 1973.

31) Kenner, V. H. and W. Goldsmith, "Impact on a Simple Physical Model of the Head," J. of Biomechanics, Vol.6,No.1, pp.1-12, 1973.

32) Gross, A. G., "Impact Thresholds of Brain Concussion," J. Am. Med., Vol. 29, pp. 725-732, 1958.

33) Unterharnscheidt, F. and K. Sellier, "Closed Brain Injuries: Mechanics and Pathomorphology," Proc. Head Injury Conf., Lippincott, pp.321-341, 1966.

34) Lindgren, Sten O., "Experimental Studies of Mechanical Effects in Head Injury," Acta Chir. Scand. Suppl. 360, 1966.

35) Kopecky, J. A. and E. A. Ripperger, "Closed Brain Injuries: An Engineering Analysis," J. of Biomechanics, Vol.2, No.1, pp. 29-34, 1969.

TABLE 1 RESULTS OF RHESUS MONKEY HEAD IMPACTS AND IMPEDANCE TESTS

Direction of Head Impact	Acceleration (G's)	Pulse Duration (msec)	Model Constants				Mean Strain
			w_1	k	c	w_2	ε
Front	1800	3.6	.051	39,000	1.6	1.1	0.032
Side	1500	2.8	.040	33,000	2.1	1.0	0.032
Top	980	7.0	.030	18,000	1.2	0.9	0.032
Back	1000	3.4	.035	20,000	2.9	1.1	0.032

TABLE 2 SUMMARY OF HEAD INJURY INDICE COMPARISONS

	Pulse I.D.	Accel. Pulse Duration (msec)	Accel. Pulse Peak (g's)	GSI	HIC Duration (msec)	HIC Aver. Accel. (g's)	HIC	JTI	RBM	EDI	MSC
	Sine	10	100	458	7.8	83	415	0.853	-	0.146	.0039
	Triang	10	100	286	5.7	72	247	0.691	-	0.118	.0045
	Square	10	100	1000	10.0	100	1000	1.026	-	0.172	.0061
Dummy Windshield Class A	22	128	162	1170	40	49	680	0.749	0.909	0.120	.0037
	23	105	207	1609*	4	125	702	1.087*	0.905	0.180	.0053
	24	187	144	922	33	54	704	0.718	0.950	0.123	.0027
	25	188	109	717	28	52	555	0.683	0.937	0.111	.0039
	26	182	111	825	43	45	597	0.764	0.934	0.122	.0028
	41	187	248	2080*	38	61	1082*	1.117*	0.890	0.182	.0056
	42	211	290	3066*	2	254	2057*	1.608*	1.283*	0.269*	.0065*
	43	202	150	917	46	47	716	0.668	0.741	0.109	.0041
	44	250	117	1154	32	59	841	0.839	0.863	0.137	.0033
	45	250	111	825	18	66	644	0.901	1.098	0.147	.0037
Dummy Windshield Class B	11	300	418	2229*	36	75	1273*	1.018*	1.373*	0.164	.0046
	12	155	151	1020	50	46	701	0.721	0.909	0.115	.0030
	13	153	174	1194	24	68	903	0.812	1.087	0.130	.0028
	21	105	150	1275	15	61	438	0.922	0.940	0.153	.0038
	31	88	85	395	43	31	232	0.603	0.685	0.100	.0018
	32	62	58	400	31	42	355	0.527	0.733	0.088	.0024
	47	250	98	1577*	34	69	1360*	0.913	0.116	0.146	.0060
Holloman Human Air Bag Tests	51	162	80	1246	26	60	718	0.795	1.051	0.130	.0047
	53	167	66	915	12	49	544	0.666	0.858	0.107	.0038
	54	189	56	697	36	41	392	0.568	0.782	0.092	.0029
	55	169	75	1249	30	55	682	0.759	0.980	0.122	.0043
	57	154	76	1324	26	61	765	0.764	0.997	0.122	.0043
	58	173	79	1212	24	63	763	0.798	1.059	0.128	.0044
	59	187	78	1224	28	59	751	0.783	1.031	0.126	.0041
	5B	159	78	1446	30	61	875	0.754	1.011	0.120	.0044
	5C	154	67	1077	126	29	563	0.666	0.866	0.109	.0038
Dummy Air Bag	5A	147	78	1305	124	34	848	0.683	0.892	0.110	.0042
	52	158	65	987	122	29	546	0.628	0.819	0.102	.0036
	56	144	71	1394	124	34	832	0.716	0.975	0.116	.0041

*Exceeds tolerable value of applicable criterion.

TABLE 3 COMPARISON OF NORMALIZED HEAD INJURY INDICES

	TEST I.D.	SI	HIC	JTI	RBM	EDI	MSC
Dummy Wind-shield Tests Class 'A'	22	.780	0.680	0.749	0.727	0.600	0.607
	23	1.073	0.702	1.087	0.724	0.900	0.869
	24	.614	0.704	0.718	0.760	0.615	0.443
	25	.478	0.555	0.683	0.750	0.555	0.639
	26	.550	0.597	0.764	0.747	0.610	0.459
	41	1.386	1.082	1.117	0.712	0.910	0.918
	42	2.043	2.057	1.608	1.026	1.345	1.066
	43	.611	0.716	0.668	0.593	0.545	0.672
	44	.769	0.841	0.839	0.690	0.685	0.541
	45	.550	0.644	0.901	0.878	0.735	0.607
Dummy Wind-shield Tests Class 'B'	11	1.485	1.273	1.018	1.098	0.820	0.754
	12	.680	0.701	0.721	0.727	0.575	0.492
	13	.746	0.903	0.812	0.870	0.650	0.459
	21	.850	0.438	0.922	0.752	0.765	0.623
	31	.263	0.232	0.603	0.548	0.500	0.295
	32	.267	0.355	0.527	0.586	0.440	0.393
	47	1.051	1.360	0.913	0.893	0.811	0.983
Hollo-man Air Bag Tests Human	51	.831	0.718	0.795	0.841	0.722	0.770
	53	.610	0.544	0.666	0.686	0.594	0.623
	54	.465	0.392	0.568	0.627	0.511	0.475
	55	.832	0.682	0.759	0.784	0.677	0.705
	57	.882	0.765	0.764	0.798	0.677	0.705
	58	.808	0.763	0.798	0.847	0.711	0.721
	59	.816	0.751	0.783	0.825	0.700	0.672
	5B	.964	0.875	0.754	0.809	0.666	0.721
	5C	.718	0.563	0.666	0.693	0.605	0.623
Hollo-man Air Bag: Dummy	52	.658	0.546	0.628	0.655	0.566	0.590
	56	.929	0.832	0.716	0.780	0.644	0.672
	5A	.870	0.848	0.683	0.714	0.611	0.689

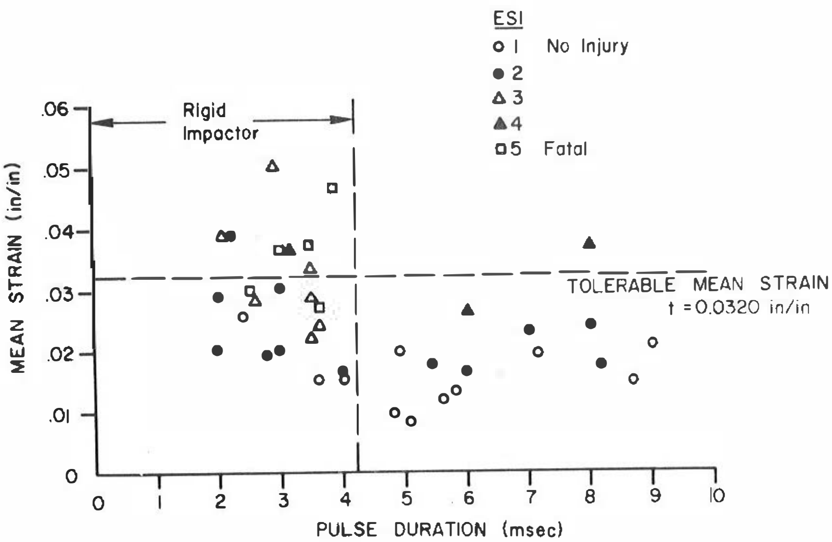

Figure 1. MSC Strain Levels for Rhesus Head Impacts Variable Direction and Pulse Duration.

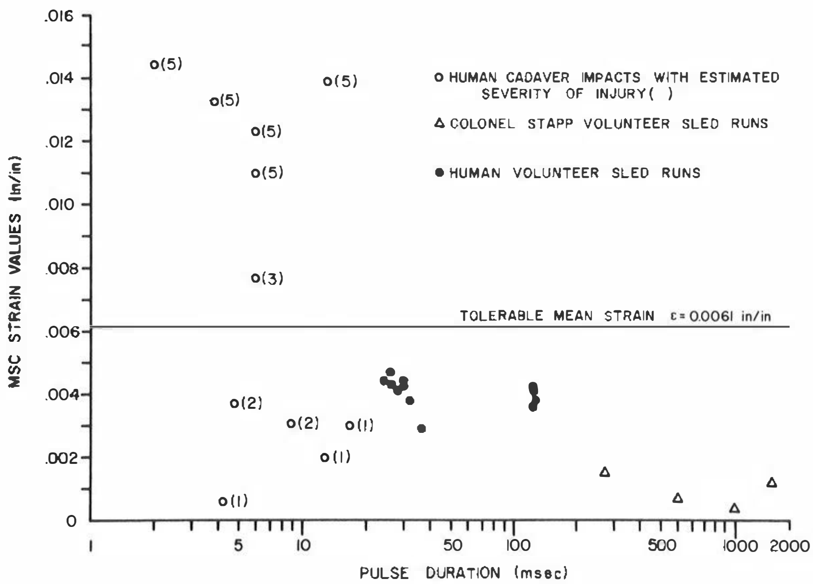

Figure 2. Mean Strain Criterion for Humans vs Pulse Duration.

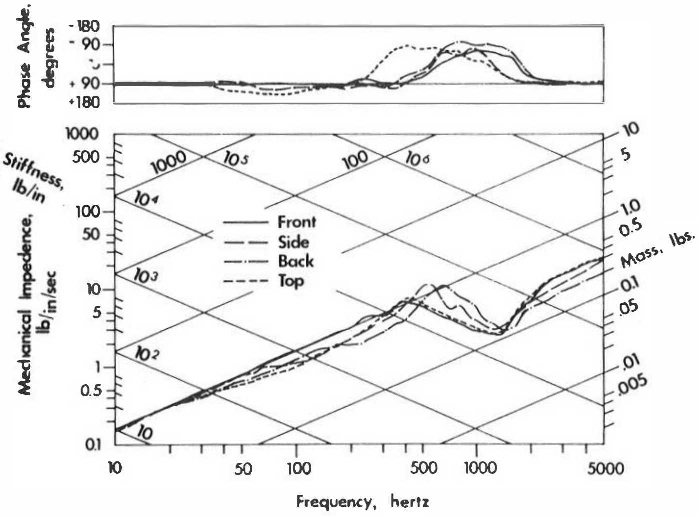

Figure 3. Mechanical Impedance of Rhesus Monkey Head.

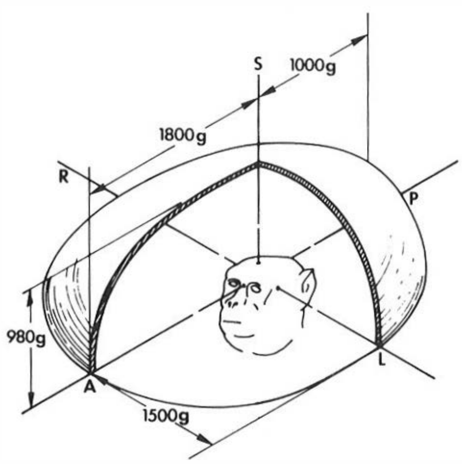

Figure 4. Critical Acceleration Surface for Rhesus Monkey, Injury Index = 3.

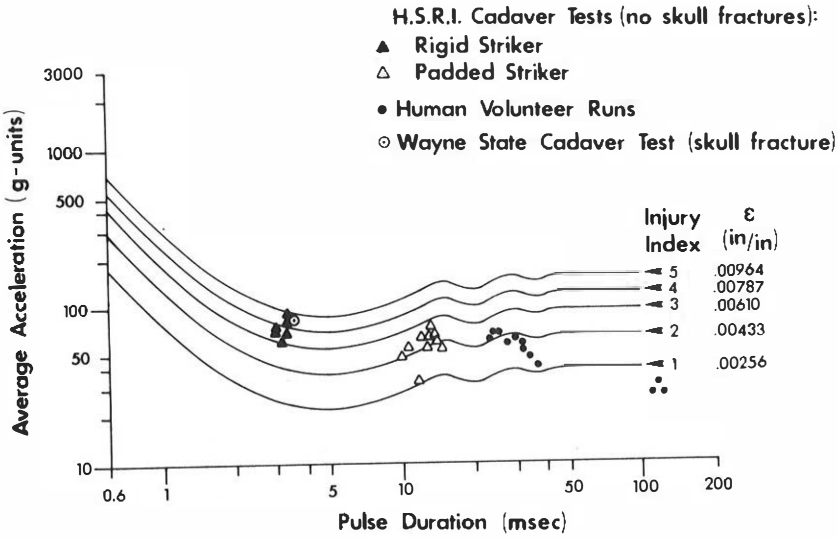

Figure 5. Mean Strain Criterion for Humans, Sagittal Plane Loading.

SI	HIC	JTI	RBM	EDI	MSC
SEVERITY INDEX (GADD)	HEAD INJURY CRITERION (VERSACE & NHTSA)	J-TOLERANCE INDEX (SLATTENSCHEK)	REVISED BRAIN MODEL (FAN)	EFFECTIVE DISPLACEMENT INDEX (BRINN)	MAXIMUM STRAIN CRITERION (STALNAKER)
Weighted Impulse of a(t) $SI = \int_0^T [a(t)]^{2.5} dt$ Time in seconds Acc. in g-units	Weighted Impulse of a(t) Let $\bar{a}_{12} = \frac{\int_{t_1}^{t_2} a(t)dt}{(t_2 - t_1)}$ $HIC = \left\{ \max_{\substack{all \\ t_1, t_2}} \left[\bar{a}_{12}^{2.5} (t_2 - t_1) \right] \right\}$, $0 < t_1 < t_2 \leq T$	a(t), k, c, m, x(t) $\omega_n = \sqrt{k/m}$ (rad/sec) $\beta = c/c_c$ $\omega_n = 635.$ $\beta = 1.0$	a(t), k, c, m, x(t) $\omega_n = 175.$ $\beta = 0.4$	a(t), k, c, m, x(t) $\omega_n = 482.$ $\beta = 0.707$	a(t), m_1, k, c, m_2, x(t) $m_1 = 0.6$ (lbs) $m_2 = 10.0$ (lbs) $c = 2.0$ (lb sec/in) $k = 50000.$ (lb/in)
$SI_{tol} = 1000$	$HIC_{tol} = 1000$	$J = \frac{x_{max}}{0.0925}$, $\frac{in}{in}$ $J_{tol} = 1.0$	T < 20ms: $\dot{x}_{tol} = 135.3 \frac{in}{sec}$ T > 20ms: $x_{tol} = 1.25$ in	x_{tol} = A-P / RES. HUMAN: 0.15 in / 0.18 in DUMMY: 0.17 in / 0.2 in	$\varepsilon = x_{max}/L$ HUMAN: L = 5.75 in (A-P) $\varepsilon_{tol} = 0.0061$ in/in

Figure 6. Summary of Head Injury Criteria.

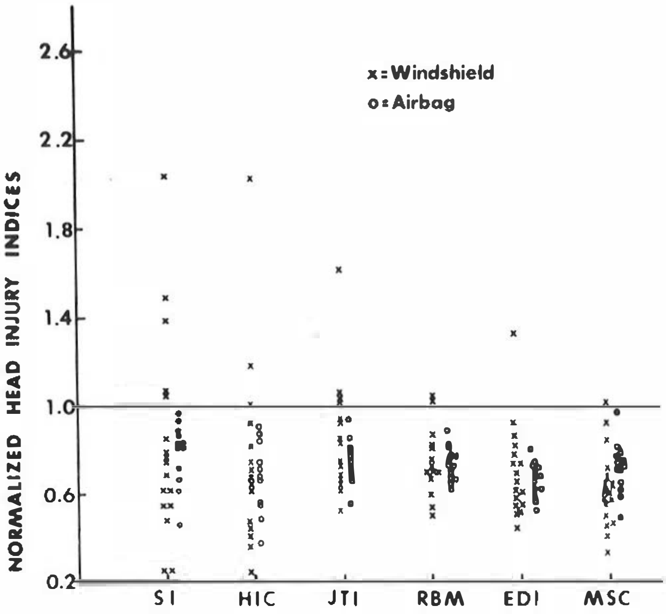

Figure 7. Comparison of Head Injury Criteria.