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AFRL BIODYNAMICS
DATA BANK

DYNAMIC RESPONSE OF THE HUMAN HEAD TO $+G_X$ IMPACT

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__ABSTRACT__ - Current knowledge of the response of the human head to short-duration acceleration is largely based upon the tests of animals and human cadavers. These data have been used to develop injury-limit curves and, more recently to develop a variety of mathematical models intended to estimate the response of the head to any acceleration-time history. In order to evaluate the existing injury-limit curves and models, a test program was conducted to measure the dynamic response of the head during whole-body impact exposures. Volunteer subjects participated in 79 experimental-level tests performed under nine different impact conditions. A vertical impact tower was used to produce, $+G_X$ acceleration-time profiles with amplitudes up to 45G, velocity changes up to 15.5 ft/sec., and rise times from 1 to 23 msec. The subjects were restrained to a couch instrumented to measure impact forces and acceleration. The impact surface for the head was a 4-inch diameter individually molded fiberglass occipital headrest. The head was restrained to prevent rotation or rebound. Measured head acceleration and headrest load indicated a system with a natural frequency of approximately 100 Hertz. The response of the head was similar to that of the Maximum Strain Criterion Model described by Stalnaker, McElhaney, and Roberts in 1970.

__INTRODUCTION__ - In the United States, trauma is the leading cause of death for individuals between the ages of 1 and 44 years. Of the 160,000 deaths per year, greater than 50,000 are a result of traumatic head injuries. Additionally, there are greater than 40,000 people who are permanently disabled from head injuries. These exceedingly high incidences of death and disability due to traumatic head injury have had a tremendous socio-economic impact on the United States. Many of these injuries are preventable and/or treatable. Continued research into the mechanisms and biomechanics of head injury is, therefore, invaluable.

In-depth understanding of the mechanics and mechanisms of head injury could lead to the following: (1) valuable improvements both in the treatment of brain injuries and in the design of safety features to prevent head injuries, (2) improved data bases for computer modeling and advanced manikin development, (3) better head-injury criteria for evaluation of head-injury potential, and (4) decrease of the tremendous socio-economic burden imposed upon the United States society as a result of traumatic brain injury. Efforts to study the mechanics of head response to impact and to establish head-injury criteria were pioneered by Lissner, Gurdjian, and others in the early 1960's. They developed a curve for head-impact tolerance which became known as the Wayne State Tolerance Curve (WSTC). The original data points were obtained by impacting embalmed cadaver heads to produce a linear skull fracture. These points represented high-acceleration, short-duration impacts. The curve was later expanded to include longer duration impacts from previous studies using animal tests, human volunteer tests, and unembalmed cadaver tests. Failure criteria for the animal and cadaver tests consisted of linear skull fracture and/or concussion. Human volunteer tests were conducted up to the point of short-term cardiovascular compromise.

In 1966 Gadd [3] proposed a mathematical formulation of the Wayne State Tolerance Curve which became known as the Gadd Severity Index. This formulation was based on a log-log straight-line approximation of the Wayne State Tolerance Curve. Later, in 1971, Versance [12] presented a modified mathematical formula based on the Gadd Severity Index. This new formulation could better handle long-duration impact and multiple-peak impact. This formula became known as the Head Injury Criterion (HIC), which is now the automotive industry standard.

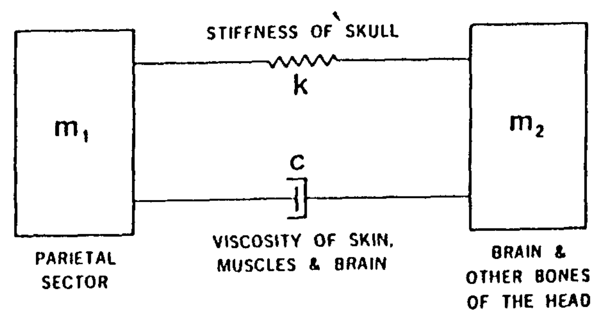
$$HIC = \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a\,dt \right]^{2.5} (t_2 - t_1)$$

t_1 = arbitrary time in acceleration pulse
t_2 = for a given t, a time in the pulse which maximized the value of HIC
a = resultant head acceleration

Many mathematical and mechanical models have been developed from the WSTC data. Several additional models have been based on more recent original experimental data.

One model of particular interest is the Maximum Strain Criterion (MSC) model proposed by Stalnaker and McElhaney in 1970 [11]. Unlike many other models the MSC model was not based on the Wayne State Tolerance Curve but was derived from Stalnaker's and McElhaney's experimental data. They conducted vibration tests on both human and monkey cadaver skulls as well as anesthetized monkey skulls to determine the driving-point impedence of the head. From the observed impedence characteristics they developed a simple two-degree-of-freedom, lumped-parameter model (Figure 1).

In their experiment particular emphasis was placed on linear accelerations where angular accelerations were minimized. Thus, the model applies to direct translational head impacts and not to angular acceleration of the head. This model provides a method to analyze the effects of different acceleration waveforms as well as to predict a continum of injury levels. Figure 2 shows the MSC model response for different acceleration waveforms as well as a comparison to the WSTC.



MAXIMUM STRAIN CRITERION HEAD MODEL

FIGURE 1

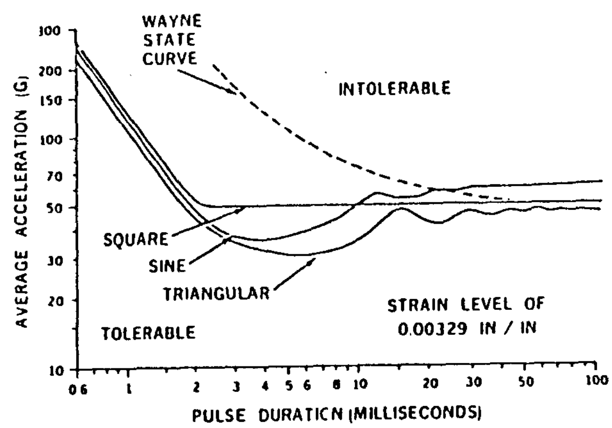


FIGURE 2

OBJECTIVE/HYPOTHESIS - The overall objective of our experiment was to measure head inertial response to $+G_x$ impact accelerations. Specifically, we wanted to do the following: (1) compare dynamic head response data to the response of existing head injury models, (2) provide a data base for model verification and/or model development, and (3) define the natural frequency of the head from dynamic-response data.

The hypothesis tested was that the magnitude of the dynamic response of the head will change as the rise of a half-sine acceleration pulse decreases below

20 msec. The null hypothesis was that there is no difference in the dynamic response of the head as the acceleration rise time decreases.

METHODS - In our experimental effort we used human volunteers to study the response of the head to translational impact. The testing conditions investigated are shown in Table 1.

TABLE 1: TEST CONDITIONS

TEST MATRIX CELL	TIME TO PEAK G (MSEC)	ACCELERATION (G)	VELOCITY CHANGE (FT/SEC)
O	1.0	45	1.7
N	2.3	25	1.5
M	3.9	25	3.2
J	5.2	20	3.1
P	6.0	20	3.6
R	8.3	20	4.6
L	11.4	20	5.7
T	16.0	15	6.8
I	23.0	20	15.5

The order of presentation of the test conditions was randomized for cells O, N, M, L, and I. The four additional test conditions J, P, R, and T were later added in a non-random fashion to assist in the interpolation of the data. To minimize the potential of injury to each subject, the tests were conducted at presumed sub-injury impact acceleration levels. The waveform of a typical acceleration-time profile is shown in Figure 3.

The volunteer subjects (13 men) were active duty officers and enlisted personnel at Wright-Patterson Air Force Base. Prior to participation, the subjects were required to meet stature, weight, and sitting-height criteria. The subjects also completed a medical screening more rigorous than a flying class II physical.

The experiment was carried out at the Harry G. Armstrong Aerospace Medical Research Laboratory (AAMRL) using a vertical impact tower (IMPMODE). The test assembly (including seat, restraint system, and instrumentation) was mounted to the impact carriage of the IMPMODE. This carriage was raised to a specified height and was allowed to fall freely along vertical rails onto an elastomeric decelerator at the base of the tower. The acceleration profile is determined by the type and number of the elastomeric decelerators as well as the drop height. The test seat was of generic design. An individually-molded fiberglass head support was made for each subject. This provided a rigid contact surface area of approximately 80 cm^2 to the occipital area of the head. The subjects were restrained at the arms, legs, waist, chest, and head to prevent body movement. Using this test set up, rotational head movement was minimized (See Figure 4).

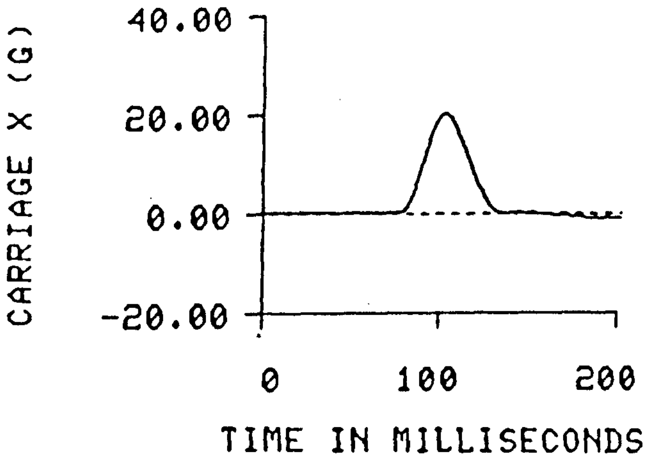


Figure 3

Measured parameters included head loads and acceleration of the carriage, headrest, and head. The left-hand coordinate reference system for acceleration (+X anterior, +Z cephlad) was used for data analysis. Data were evaluated using the Wilcoxon Paired-Replicate Rank Test and Regression Analysis. the Wilcoxon technique was selected to compare peak values of measured parameters and to establish statistical significance of observed trends in the data. This analytical approach established each subject as his own control, thereby reducing the effects of biological variability among subjects.

The 95th percentile confidence level, assuming a two-tailed test, was chosen as the level of statistical significance.

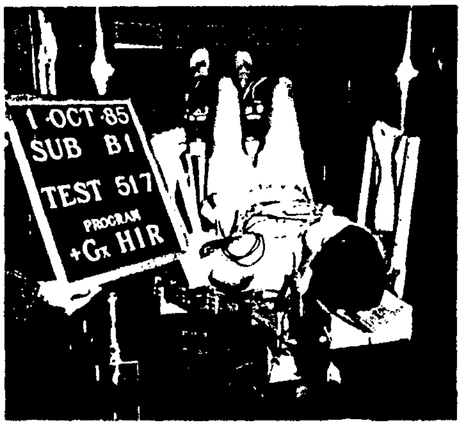


Figure 4

RESULTS - Seventy-nine experimental-level tests were performed under nine different impact conditions. Impact rise times (time-to-peak acceleration or approximately one-half the acceleration pulse duration) ranged from 1 to 23 msec with acceleration amplitudes up to 45 G and velocity changes up to 15.5 ft/sec. Results from the measured parameters in this experimental effort are given in Table 2 and graphically represented in Figures 5 and 6.

Wilcoxon analysis of the data plotted in Figure 5 indicates statistically significant differences among test cells for all combinations except cell comparisons L-I, L-N, and I-N. Wilcoxon analysis of the data plotted in Figure 6 indicates significant differences for all cell comparisons. These results are given in Table 3.

Regression analysis of the head-load data was accomplished to determine if statistically significant correlations exist between measured head loads and acceleration, velocity, or head mass. No statistically significant correlation was found among any of these parameters.

DISCUSSION - Several interesting features are evident in the graphs of the experimental data (Figures 5 and 6). First, the natural frequency of the head under our experimental conditions is approximately 100 Hz. As can be seen in Figure 5, the head maximally amplifies the input acceleration approximately one and one-half times, occuring at a rise time of 5 msec. Below the rise time of 2.3 msec significant attenuation of the input acceleration occurred. Above rise times of 8 msec the head response returned to an approximate 1:1 input / output ratio. The head-load data shown in Figure 6, demonstrated a similar response but did not quite return to a 1:1 ratio. This can be explained by the additional loading effect of the lower frequency torso / neck system. As the rise time increases, loads transmitted to the head through the neck from the torso / neck area increase. Thus, the measured head loads continued to exceed the theoretical loads at longer rise times.

TABLE 2: RESPONSE PARAMETERS

CELL	HEAD ACCELERATION / HEADREST ACCELERATION		MEASURED HEAD LOAD / COMPUTED LOAD*	
	MEAN	S.D.	MEAN	S.D.
O	0.62	0.05	0.58	0.06
N	0.95	0.13	0.83	0.11
M	1.37	0.16	1.50	0.10
J	1.53	0.13	1.56	0.10
P	1.23	0.13	1.50	0.09
R	1.01	0.07	1.35	0.10
L	0.95	0.06	1.38	0.13
T	0.92	0.03	1.28	0.10
I	0.97	0.11	1.16	0.13

*Computed loads were determined for each test by multiplying the individuals head mass times the headrest acceleration.

Defining the acceleration amplification and attenuation characteristics of the human head is crucial for the establishment of improved injury criteria and the development of more effective head protection equipment. Knowledge of the natural frequency of the head is especially important since acceleration pulses at or near this frequency or accelerations containing significant energy at this frequency will be more likely to cause injury for a given acceleration level. In particular, protective equipment should be designed to attenuate energy near this frequency.

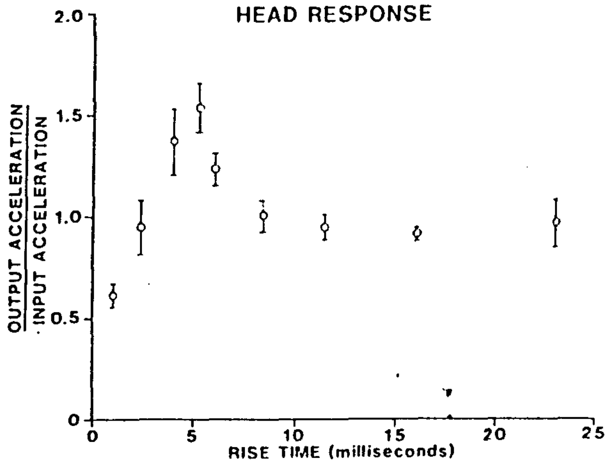


Figure 5

The response characteristics of the human head measured during our experiment are similar to a simple mechanical system. The MSC model, a two-degree-of-freedom model, closely predicted the type of response observed in our experimental effort, as can be seen by comparing Figures 5 and 6 with the MSC model response shown in Figure 7. (Note the lack of this type of response with the WSTC) shown in Figure 2.

There is, however, a small shift of approximately 2 msec in rise time where the maximal amplification occurs in our experimental data compared to the MSC model prediction. This difference is most likely explained by the difference in our experimental methods. Mc Elhaney's and Stalnaker's input and output signal devices were mounted to bone; our input and output signals were transmitted through soft tissue. Secondly, McElhaney and Stalnaker used unembalmed cadavers for testing; we used human volunteers in our experimental testing. Despite these differences the overall response predicted by the MSC model was very similar to the overall rsponse observed in our experimental data. These results suggest that the MSC model may be a reasonable basis for a head injury criterion for translational acceleration impacts.

The inaccuracy of the HIC (based on the Wayne State Tolerance Curve) has been the subject of numerous papers and

TABLE 3: RESULTS OF WILCOXON ANALYSIS#

CELL	HEAD ACCELERATION / HEADREST ACCELERATION	MEASURED HEAD LOAD / COMPUTED LOAD
L-I	NS*	.01
L-O	.01	.01
L-N	NS	.01
L-M	.01	.02
I-O	.01	.01
I-N	NS	.01
I-M	.01	.01
O-N	.01	.01
O-M	.01	.01
N-M	.01	.01

Values listed represent the level of statistical significance

*(NS) indicates no statistical significance

#Analysis was not performed for Cells P, R, J, and T since they were not randomized.

meetings. Thus, there is a recognized need for improved head-injury criteria. Alan Nahun and John Melvin in their book, The Biomechanics of Trauma, suggest new injury criteria for head injury. The new criteria separates rotational and translational acceleration injuries. For translational criteria they suggest using the Maximum Strain Criterion model. The MSC model has at least one strong advantage over the HIC model in that it allows for a graded prediction of injury potential. The capability of the MSC model to predict a graded response allows for comparison with other injury scales, such as the Abbreviated Injury Scale (AIS) and the Glasgow Coma Scale.

The inadequacies of the HIC to evaluate translational and especially angular acceleration head-injury potential must be corrected by further experiment-based efforts to develop accurate head-injury criteria. Models such as those proposed by Nahun and Melvin need to be validated. More extensive data bases and detailed information are needed to support the sophisticated finite-element models that have been proposed. A concerted effort is needed from various disciplines and organizations to validate and/or develop more adequate head-injury criteria.

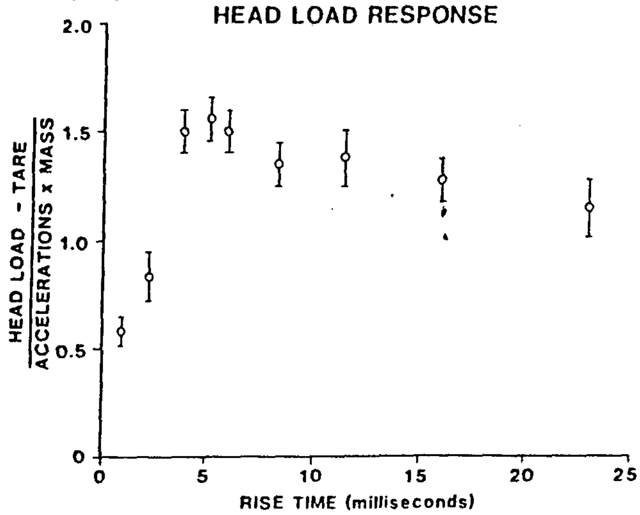


Figure 6

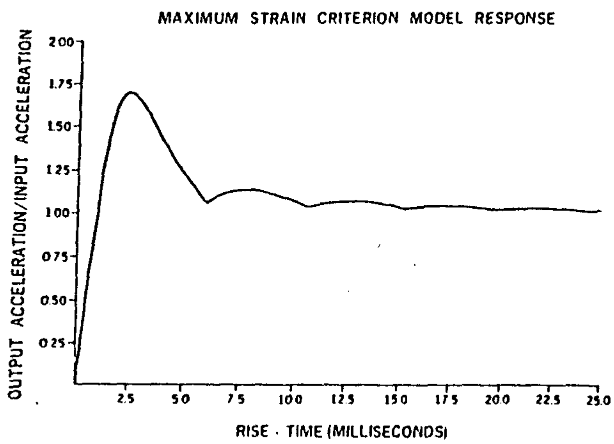


Figure 7

CONCLUSIONS - Wilcoxon Paired-Replicate Rank test analysis supports at greater than the 95 percentile confidence level rejection of the null hypothesis that there is no difference in the dynamic response of the head as the acceleration rise time decreases. Our experimental results were closely predicted by the MSC model. Data analysis for the dynamic response of the head demonstrated a natural frequency of 100 Hz. Maximal amplification of the input head acceleration occurred at a rise time of 5 msec, with attenuation at lower rise times, and a 1:1 response above this rise time. Modifying the MSC model to reflect these experimental results would lend further credence to use of the MSC model as a head-injury criterion for translational acceleration impacts, as suggested by Nahun and Melvin. The amplification of the input acceleration demonstrated by our measurements of head response has significant implications in safety-equipment design and head-injury tolerance.

NOTES AND ACKNOWLEDGEMENTS - *This research effort was conducted under the auspices of the Human Use Review Committee of the Harry G. Armstrong Aerospace Medical Research Laboratory and in accordance with Air Force Regulation 169-3, "Use of Human Subjects in Research, Development, Test and Evaluation.'*

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BIOGRAPHIES

Captain Mark D. Salerno is a flight surgeon assigned to the Biomechanical Protection Branch, Biodynamics and Bioengineering Division of the Harry G. Armstrong Aerospace Medical Research Laboratory, Wright-Patterson AFB, Ohio. Captain Salerno has a Bachelor of Arts in Mathematics from the University of Virginia and a Doctor of Medicine from the Medical College of Virginia. He is currently doing research on the biodynamic responses of human subjects to impact accelerations and on the protection provided by advanced restraint systems.

Mr. James Brinkley is the Chief of the Biomechanical Protection Branch. He has been continuously involved in research at the AAMRL since his graduation from the Ohio State University in 1958. His research activities are centered on the development of design standards for aircraft emergency escape systems, space vehicles and personnel-protection equipment for automotive as well as aircraft crash. He has authored or co-authored more than 70 journal articles and technical reports as well as two book chapters. His numerous awards include the Air Force Exceptional Civilian Service Medal, the GEICO Public Service Award, the Eric Liljencrantz Award of the Aerospace Medical Association and the SAFE Award.

Captain Mary Ann Orzech is a flight surgeon assigned to the Biomechanical Protection Branch. Captain Orzech received a medical degree, a M.S. degree in Biomedical Engineering and a B.S. degree in physics from Case Western Reserve University. She also received a M.A. degree in Marine Science from the College of William and Mary. She is presently working on an M.S. degree in Aerospace Medicine from Wright State University. Her current areas of research include evaluation of the human response to acceleration environments, escape systems, and physiological responses to suspension in fall protection equipment.