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13. ABSTRACT

The Symposium on Biodynamics Models and Their Applications took place in Dayton, Ohio, on 26-28 October 1970 under the sponsorship of the National Academy of Sciences - National Research Council, Committee on Hearing, Bioacoustics, and Biomechanics; the National Aeronautics and Space Administration; and the Aerospace Medical Research Laboratory, Aerospace Medical Division, United States Air Force. Most technical areas discussed included application of biodynamic models for the establishment of environmental exposure limits, models for interpretation of animal, dummy, and operational experiments, mechanical characterization of living tissue and isolated organs, models to describe man's response to impact, blast, and acoustic energy, and performance in biodynamic environments.

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PAPER NO. 37

A MECHANICAL IMPEDANCE MODEL FOR HEAD INJURY DUE TO LINEAR IMPACTS

by

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ABSTRACT

The mechanical impedance of the human and various other primate species heads was determined over the frequency 30-5000 hertz. A simple model was developed that closely follows the observed impedance characteristics. Spring and damping constants were evaluated and comparisons between species obtained. An impact tolerance curve was computed based on the model predictions with a maximum strain criteria. Various input pulse shapes were analyzed and the effect of pulse shape and duration studied for the different species. Published values of tolerable impulses were examined and compared with the model predictions. Some discrepancies have been noted and analyzed. Particular emphasis in this work has been on linear impacts to the side of the head where angular accelerations do not predominate. Results are presented in the form of tolerable average acceleration versus pulse duration curves for various pulse shapes derived from primate head impacts. A comparison with the Wayne State tolerance curve and Eiband's work of 1959 has also been made.

INTRODUCTION

A critical factor in the design of a motor vehicle interior is the limiting deceleration-time pulse the vehicle can undergo without occupant injury. The type of tolerance data needed for this interior design is not readily available. Although much work has been done in determining tolerances for thoracic and frontal head impacts, much more work is needed to further substantiate existing data and to generate new data for other regions of the body.

The acceleration-time tolerance curve developed at Wayne State University over the past several decades is the only empirically based tolerance curve available for frontal head impacts. Although this curve is widely used as a standard for head injury, the experimental design upon which it is based has been the subject of much controversy. The original Wayne State Tolerance Curve (W.S.T.) (Lissner 1960) consists of six data points, five plotted as peak acceleration and one arbitrarily plotted as the average of the peak acceleration and the average acceleration.

A revised W.S.T. curve was introduced by Patrick (1963). This curve is a composite based on a wide variety of pulse shpaes and striker configurations. The failure criteria used was generally skull fracture. In spite of the many interpretive difficulties associated with this curve, it has been in the past the principal source of hard data for human head injury tolerance.

The work upon which this paper is based was supported in part by a contract with the Department of Transportation, National Highway Safety Bureau, #FH-11-7288 and a contract with National Institutes of Health, National Institutes of Neurological Diseases and Stroke, #PH-43-67-1136.

Several attempts to generalize the Wayne State Tolerance Curve have been made in the last four years. One such characterization of this curve was set forth by C. W. Gadd (1966), who defines the severity index as:

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

where:

a = acceleration, force, or pressure of response function producing threshold or injury of given degree.

n = weighing factor greater than 1.

t = time, seconds

The value for n was obtained from a log-log plot of the Wayne Curve and Eiband's (1969) work for spineward accelerations of seated humans. The limiting index for survival was determined from points on these curves. This severity index is tied very closely to pulse shapes in setting the survival limit.

The J Tolerance value (Slattenschek 1968) or the Vienna Institute index is the newest generalization of the Wayne Curve. In this effort, a single-degree-of-freedom model is assumed and the model constants are determined by fitting a linear differential equation to the Wayne State data. A J Tolerance value is then defined by:

$$J = \frac{X_{max}}{X_{erta}} \qquad (2)$$

where:

Xerta = tolerable amplitude from Wayne State Curve.

Xmax = maximum X generated by putting the acceleration pulse in question into the model differential equation.

thus:

$J > 1$ Not survivable

$J < 1$ Survivable

Once more the index is determined from the Wayne State Tolerance Curve.

If the dynamic response of the head is known for a wide range of frequencies and if injury can be related to this response, then a tolerance curve can be generated for all pulse durations, given only one point on the tolerance curve, provided the injury mechanism does not change over that range. This paper is an attempt to investigate this hypothesis from a model based on driving point impedance data from human and other primates.

METHOD

The mechanical driving point impedance of human and other primate heads was determined over a frequency range of 30-5000 hertz utilizing the following experimental design.

The primates were anesthetized and a 10 millimeter (mm) circular hole was cored 0.25 inch above the ear canal on the side which was attached to an electromagnetic shaker. The loading fixture was then fastened to the skull at this site. On the opposite side of the skull a similar hole was made and a miniature accelerometer attached (Figure 1).

The sub-human primates skull via the load cell was rigidly attached to the platen of a 200-pound electromagnetic shaker (Figure 2). The bodies were supported in a sling hammock. A servo controller was then set to apply a sinusoidal constant amplitude acceleration of either 10 or 20 g's to the head (Figure 3). In addition, an accelerometer was placed on the free side of the head and the transmitted acceleration recorded. (Stalnaker and McElhaney 1970) A sweep oscillator drove the shaker system over a 30 to 5000 Hz cycle range, while an automatic on-line analogue impedance computer was used to convert the force-time and acceleration-time information into a phase and impedance

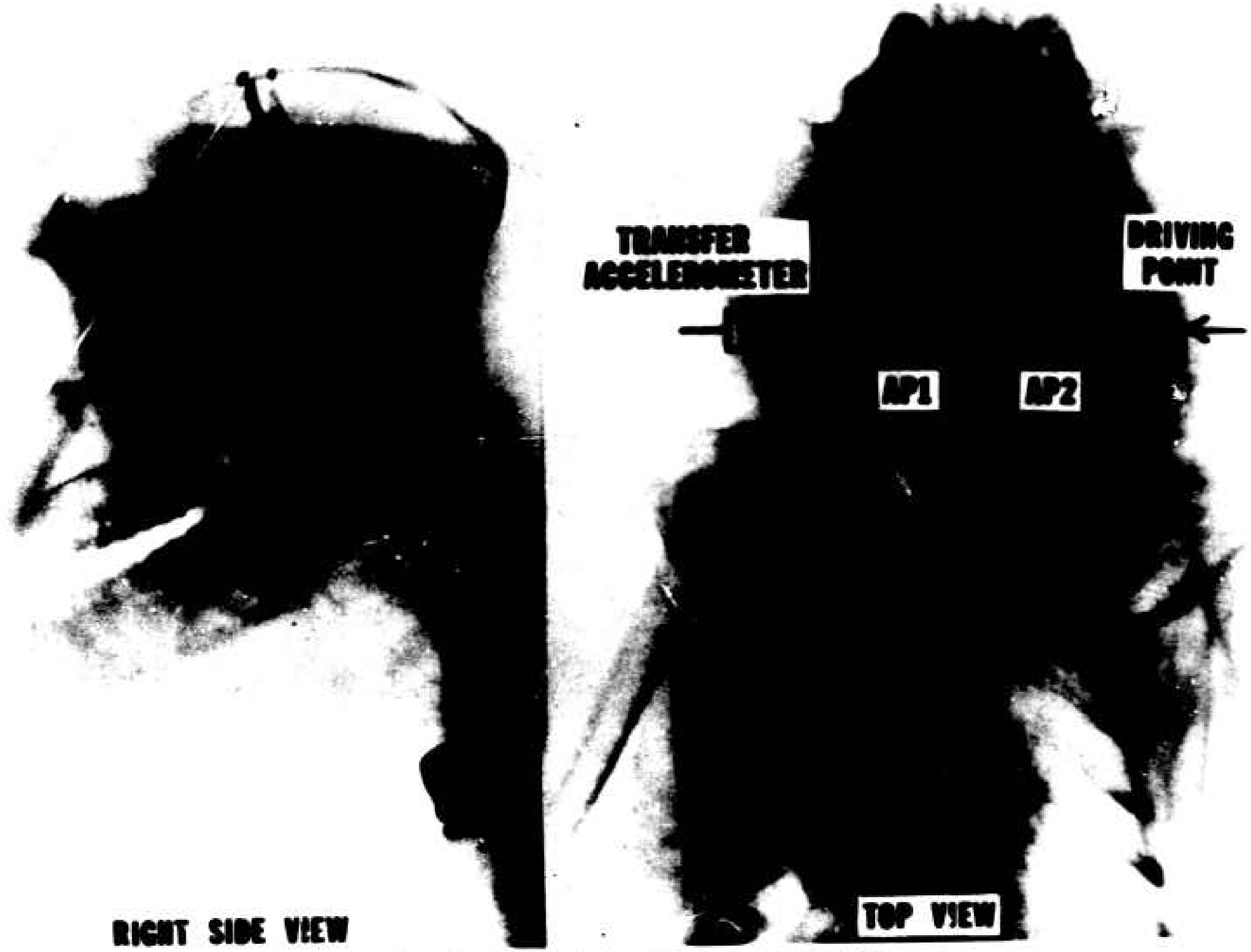

Figure 1. X-ray of Monkey Showing Attachment Point to Shaker

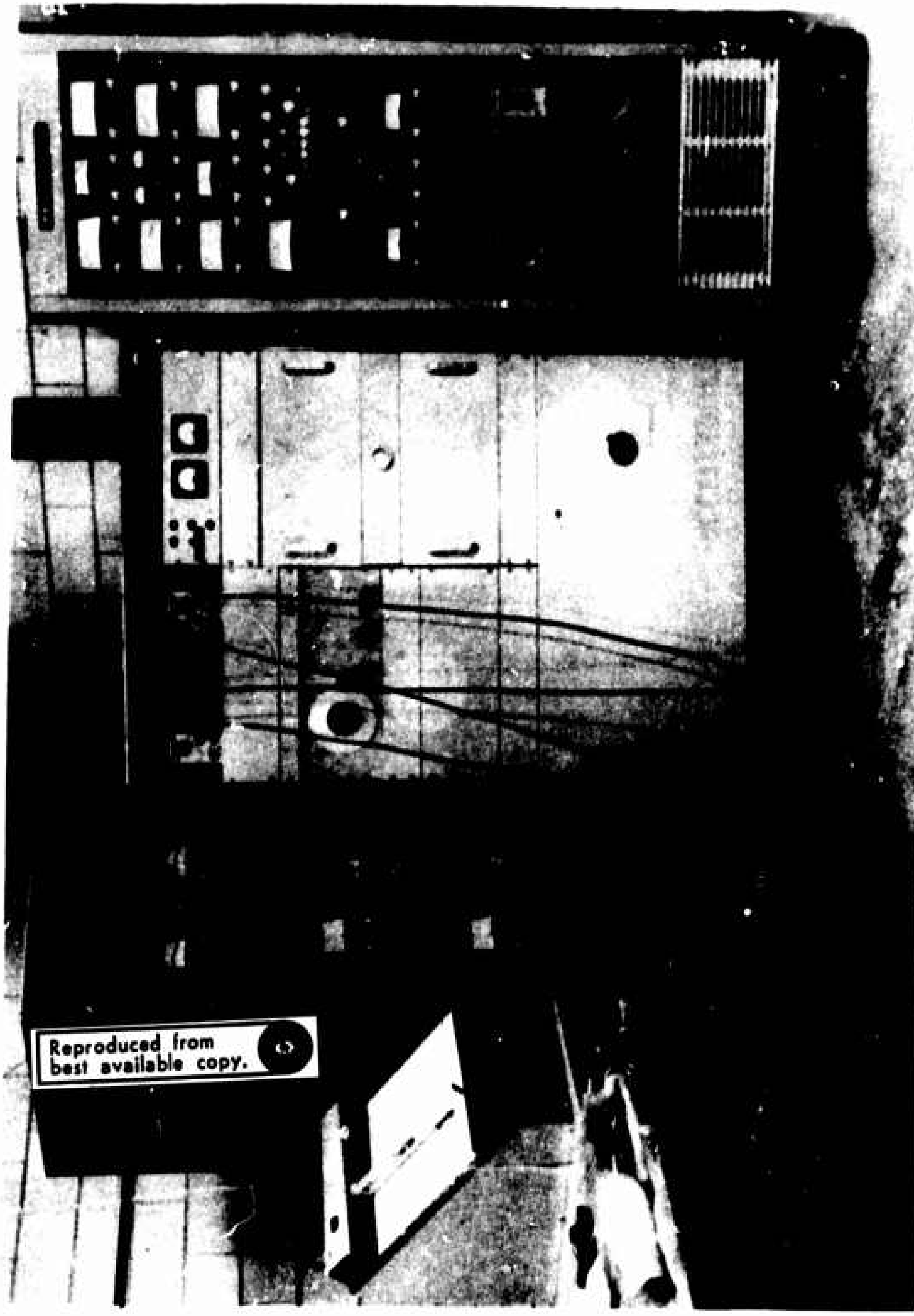

Figure 2. Mechanical Impedance Test Set-Up

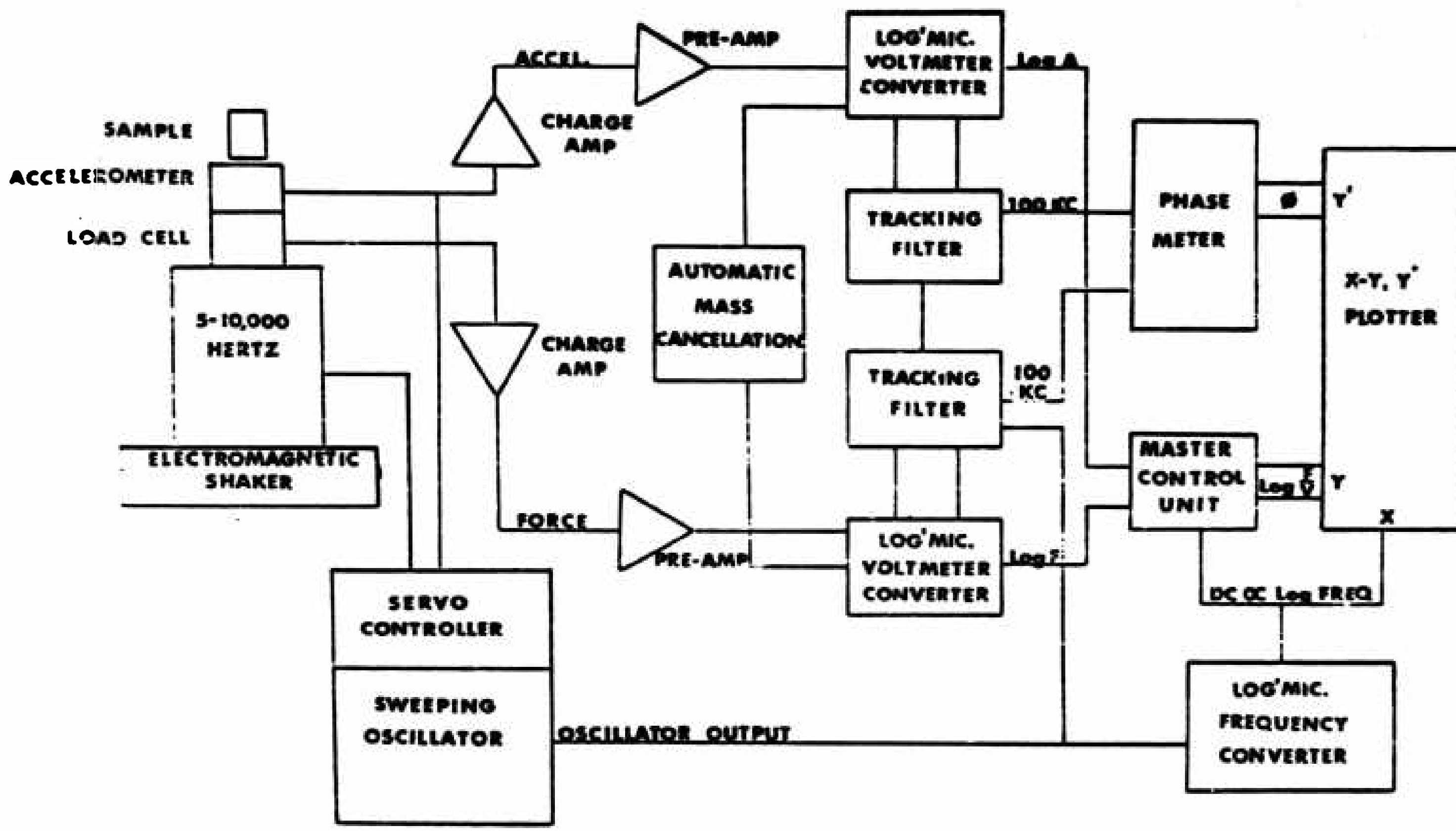

Figure 3. Automatic Impedance Measuring System

versus frequency plot. With this system, the test could be performed in less than one minute (depending on sweep rate) and produces a continuous plot of impedance. Mechanical impedance versus frequency on an x, y_1, y_2 recorder was recorded for the living anesthetized sub-human primates (Figure 4). The primates tested were then sacrificed and the test procedure repeated. The heads were removed from the bodies and the same parameters recorded again. The skin and mandible were removed and the test repeated. The brains were then removed through the foramen magnum and the mechanical impedance and acceleration on the free side of the skulls were recorded (Stalnaker and McElhaney 1970).

The same experiment was performed on an unembalmed 71-year old human male cadaver, who had been dead approximately 30 hours prior to the experiment. Constant accelerations of 1, 5 and 10 g's were applied over the frequency range 30-5000 Hz and the above mentioned mechanical parameters measured.

The two-degree-of-freedom system shown in Figure 5A has been developed to closely approximate the impedance characteristics of the heads as measured in these experiments.

If the system is represented schematically as in Figure 5B, the system elements are combined in parallel and series. Using the rule of parallel systems the impedance at point 4 is

$$z_4 = z_3 + z_2 \tag{3}$$

Using the rule of series system the impedance z_2 at point 2 is

$$\frac{1}{z_2} = \frac{1}{z_1} + \frac{1}{z_k + z_c} \tag{4}$$

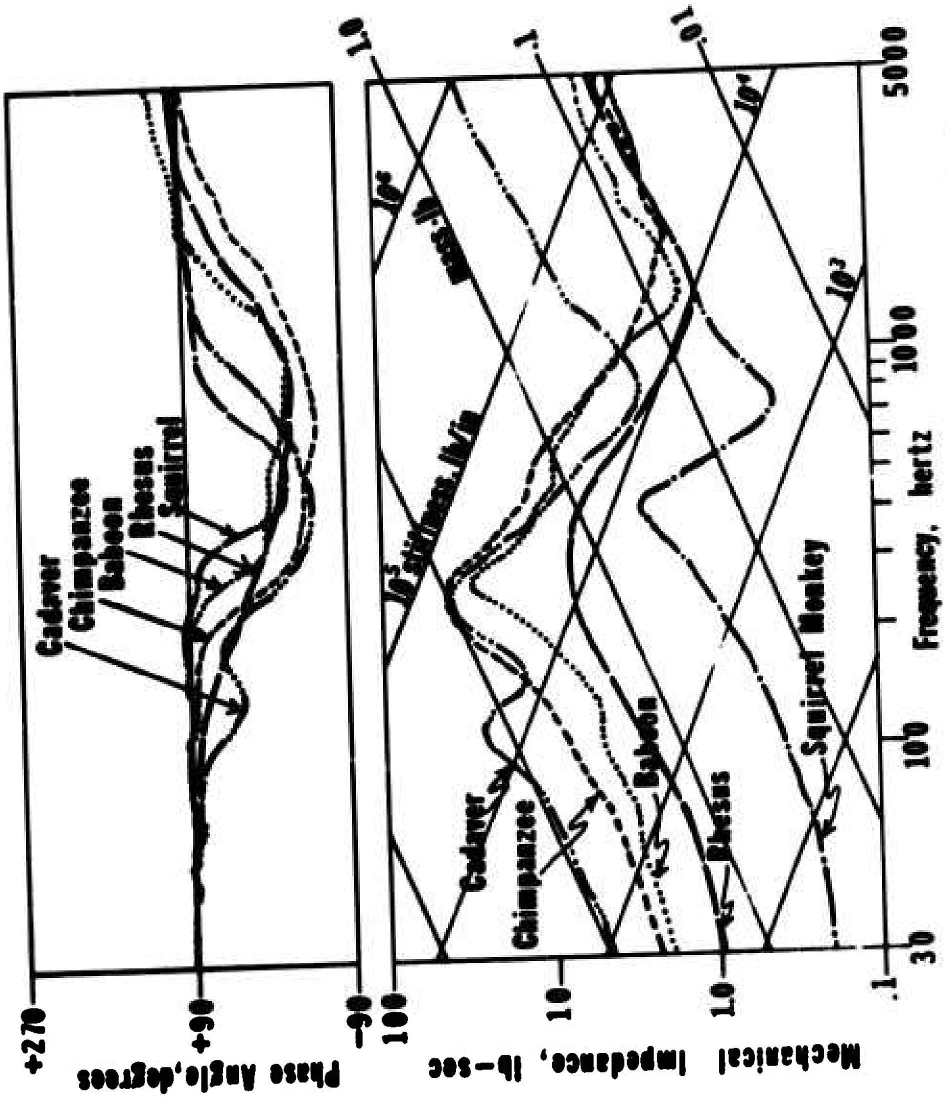

Figure 4. Mechanical Impedance of Primate Head

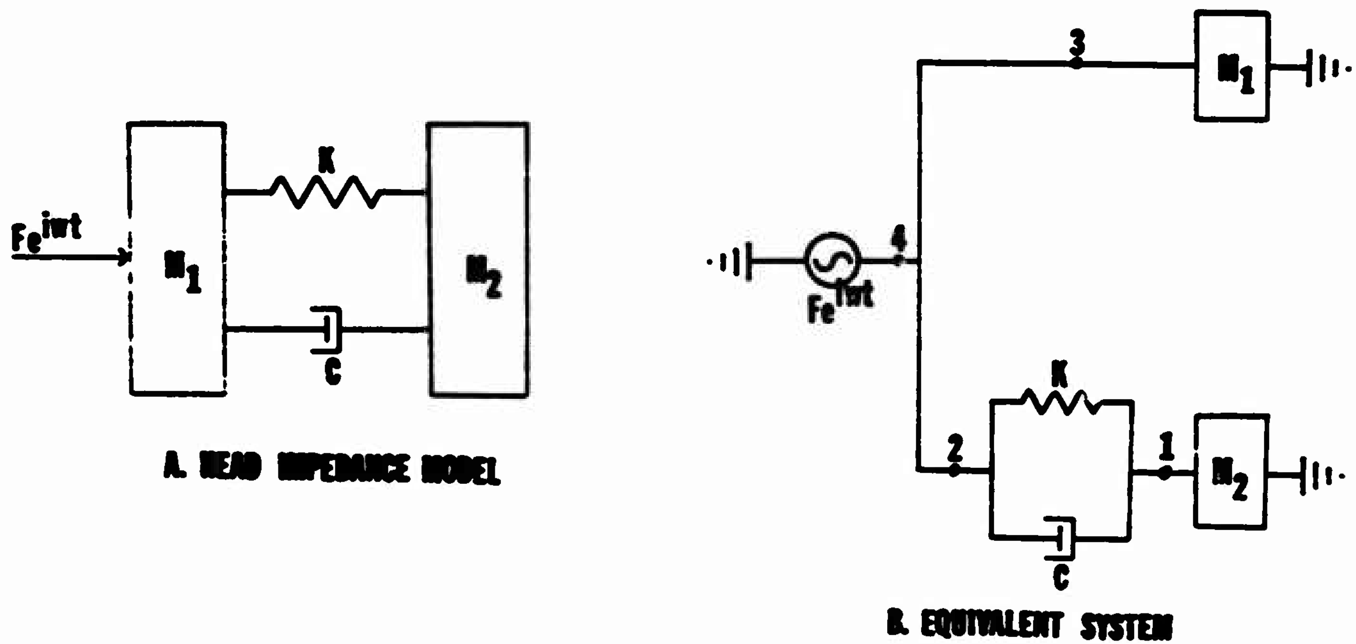

Figure 5. Two-Degree-Of-Freedom System

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so now

$$z_4 = z_3 + \frac{1}{\frac{1}{z_1} + \frac{1}{z_k + z_c}} \tag{5}$$

Substituting

$$z_4 = i\omega m_1 + \frac{1}{\frac{1}{i\omega m_2} + \frac{1}{\frac{k}{\omega i} + c}} \tag{6}$$

or

$$z_4' = i\omega(m_1 + m_2)\left[\frac{1 - \frac{\omega^2 m_1 m_2}{k(m_1 + m_2)} + \frac{i\omega c}{k}}{1 - \frac{\omega^2 m_2}{k} + \frac{i\omega c}{k}}\right] \tag{7}$$

This model has one antiresonance and one resonance. At low frequencies, the system impedance approximates the total mass of the system; at high frequencies it approximates the impedance of the drive mass element m_1. The phase angle shifts from +90° through 0° at the antiresonance frequency to -90°, and from -90° through 0° at resonant frequency back to +90°. The height of the peak and the depth of the valley are controlled by the amount of damping. The spring can be approximated for this model by stiffness line going through the inflexion point of the portion of the mechanical impedance curve between the antiresonance and resonance.

This theoretical equation was programmed for the 1130 IBM Digital Computer. By varying the magnitude of the mass, spring, and dash pot of this model a best fit of the mechanical impedance between the model and the *Macaca mulatta* test data can be approximated (Figure 6). (Stalnaker and McElhaney 1970)

The mechanical impedance of the cadaver head was similar enough to that of the living monkey head so the same model could be used. The model constants that provide a best fit with the cadaver head data are $m_1 g$ = .4 lb, $m_2 g$ = 9.0 lb, c = 2.4 lb/in, k = 2.6 x 10^4 lb-sec/in., with an antiresonance at 180 hertz and a resonance at 820 hertz (Stalnaker and McElhaney 1970).

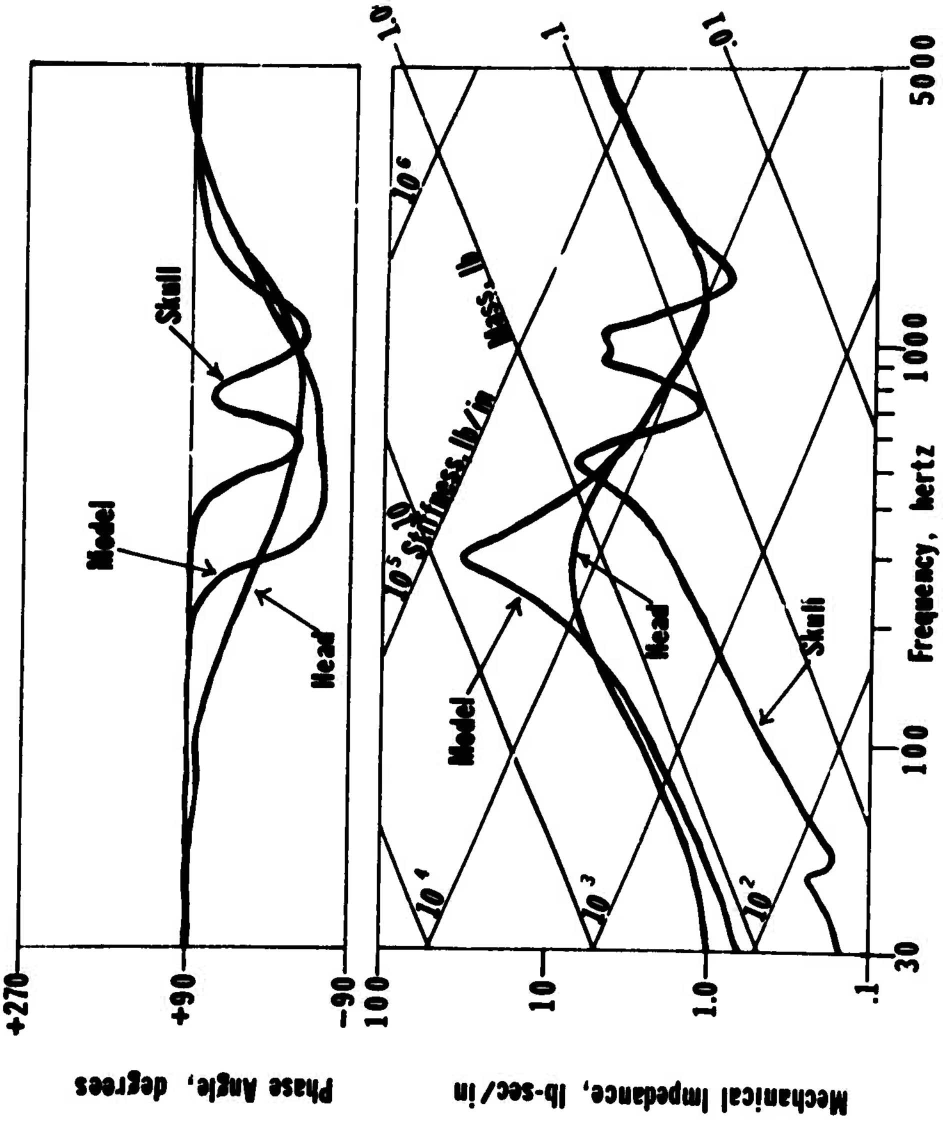

Figure 6. Comparison of Model with Living Rhesus Monkey Head and Skull

In interpreting the model constants, the following considerations apply. The calvarium is divided into approximately four major sections: the frontal bone, left and right parietal bone, and the occipital bone, which in the case of a monkey is almost entirely under the brain. The attachment to the shaker was made through one of the parietal sections. These sections of the skull are connected by sutures which provide isolation from one section to another. This implies that m_1 in the model may be one or more of these sections. The <u>Macaca mulatta</u> parietal sections were found to weigh approximately 0.065 pounds, thus it is believed that m_1 in the model corresponds approximately to the parietal sections of the skull. Comparing the spring constant obtained in the monkey series impedance tests (Stalnaker and McElhaney 1970) (1.8×10^4 lb/in) with the spring constant obtained through the impedance modeling (10^4 lb/in) leads to the conclusion that the spring element in the model corresponds for the most part to the skull stiffness. Comparing the damping constant for the whole head of the Rhesus monkey (1.2 lb-sec/in) with that of the skull alone (0.6 lb-sec/in) indicating that half the damping is due to the skin, muscle and brain (Figure 7).

With this linear two-degree-of-freedom model as a mathematical analogy of the head, many dynamic inputs to the head can be studied. The model response can be expressed in terms of the following linear differential equations (Figure 8):

$$m_1\ddot{x}_1 = c(\dot{x}_2 - \dot{x}_1) + k(x_2 - x_1) \quad (8)$$

$$m_2\ddot{x}_2 = -c(\dot{x}_2 - \dot{x}_1) - k(x_2 - x_1) \quad (9)$$

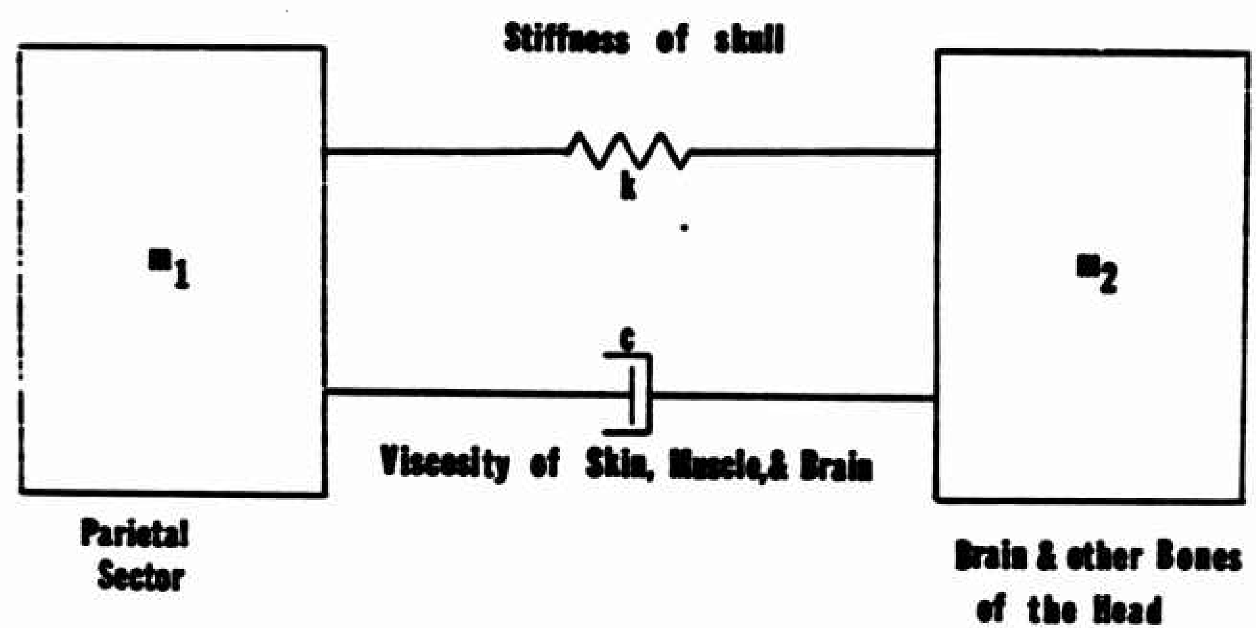

Figure 7. Model of the Head

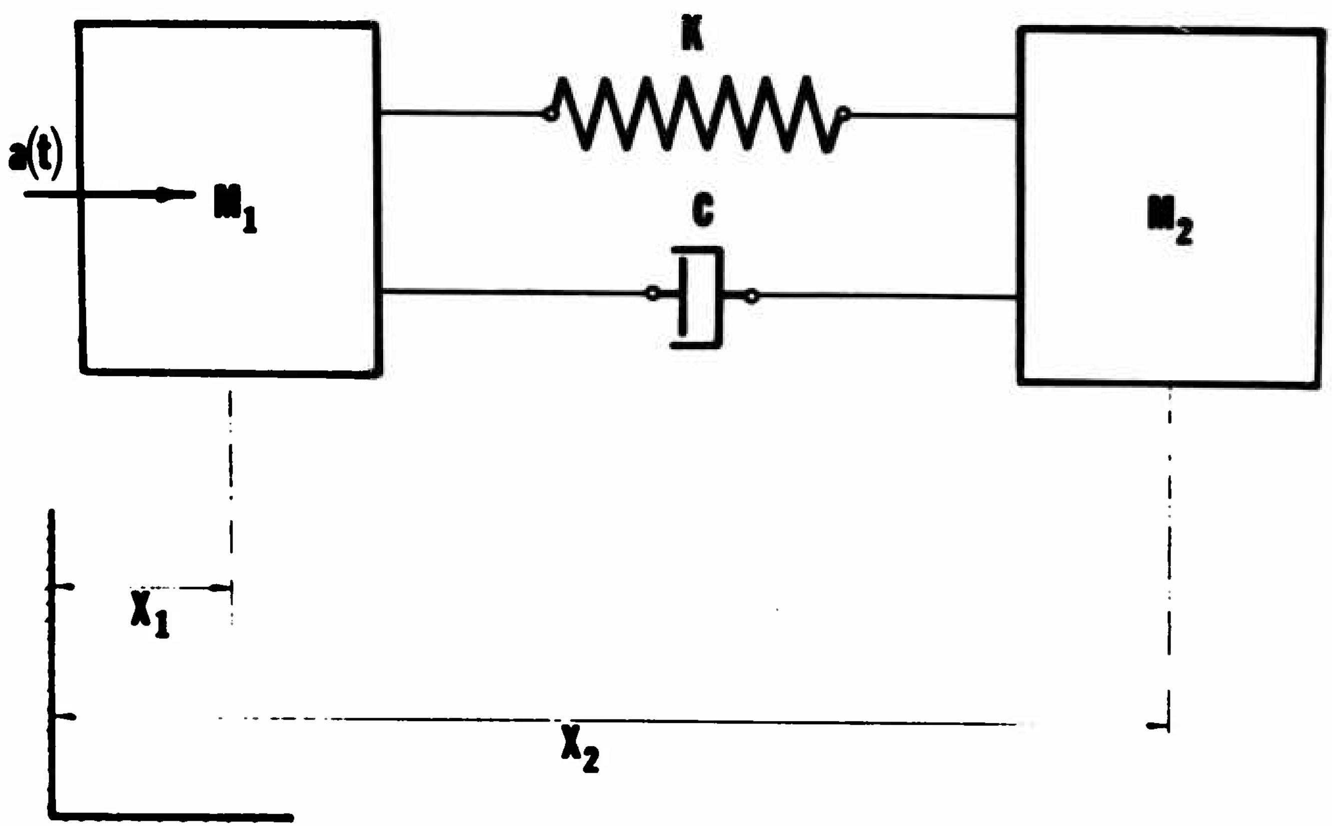

Figure 8. Reference System for Head Model

where

$$X = x_2 - x_1 \tag{10}$$

and

$$x_2 = X + x_1 \tag{10a}$$

thus

$$m_2\ddot{X} + c\dot{X} + kX = -m_2\ddot{x}_1 \tag{11}$$

Letting

$$\ddot{x}_1 = a(t) \text{ any input acceleration}$$

then

$$\ddot{X} + \frac{c}{m_2}\dot{X} + \frac{k}{m_2}X = -a(t) \tag{11a}$$

Substituting equation (10) into equations (8) and (9) and then substituting equation (9) from equation (8) we finally arrive at

$$\ddot{X} + (1 + \frac{m_2}{m_1})\frac{c}{m_2}\dot{X} + (1 + \frac{m_2}{m_1})\frac{k}{m_2}X = 0 \tag{12}$$

The required equations of motion of the model are therefore equation (11a) for a forced vibration input and equation (12) for a free vibration. With these two equations and the model constants developed above, the dynamic response of the head model can be studied for a variety of input impulses.

Human tolerance curves have been developed from impedance data for lateral driving point impedance data discussed above and from frontal impedance data compiled by V. R. Hodgson (1968). In addition, the primate lateral impact tolerance curves were obtained from the lateral impedance data. In this way, six data sets were formed (Table I).

	L (inches)	X_{max} (inches)	ε in/in	W_1 (lbs.)	W_2 (lbs.)	C $(\frac{lb\text{-}sec}{in.})$	K (lb/in)	Anti-Resonance Hz	Resonance Hz
1. R.L. Stalnaker Squirrel Monkey Lateral	1.293	0.1131	0.088	0.05	0.20	0.25	4,000.	443	987
2. R.L. Stalnaker Rhesus Monkey Lateral	2.18	0.2143	0.098	0.06	1.20	1.00	10,000.	283	1305
3. R.L. Stalnaker Baboon Lateral	2.758	--	--	0.08	3.46	1.60	30,000.	289	1926
4. R.L. Stalnaker Chimpanzee Lateral	3.504	--	--	0.08	4.75	2.40	35,000.	265	2070
5. R.L. Stalnaker Human Lateral	4.718	0.0155	0.0033	0.40	9.00	2.40	26,000.	167	812
6. V.R. Hodgson Human Longitudinal	5.78	0.0190	0.0033	0.60	10.00	2.00	50,000.	207	923
7. Vienna Inst. of Tech. Human Longitudinal	--	--	--	--	10.00	33.00	10,400.	102	--

Table I. Head Model Parameter.

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The criterion for injury was assumed to be the average strain in the brain. This then is the X in the model normalized by the linear dimensions of the brain at the point of impact and in the direction of the impact.

The maximum deflection X for human heads in the longitudinal direction was determined from A. M. Eiband's (1959) work on abrupt transverse decelerations (Figure 9). A rectangular pulse of 50 G's for 45 msec. was used as a survival acceleration pulse. This same strain used for the longitudinal direction was assumed as the limit for lateral impact, the underlying assumption here is that the brain is equally vulnerable to strain in all directions.

The maximum X for sub-human primates was calculated from lateral head impact data (Figure 10). The sub-human primates were impacted on the side of the head with a rigid constant velocity impactor. The extent of the injuries were determined in 72-hour post-impact autopsy. The tolerable impact level for each primate was determined. The acceleration-time pulse for the tolerable impact was then used as the head model data input.

RESULTS

The results of this study are represented in the form of average acceleration versus pulse duration curves.

The average strain generated by a (50 G, 45 msec) rectangular pulse in the longitudinal direction in the human head was found to be 0.00329 in/in. For the same given strain level a 15 g spread was found above 10 msec. depending on the pulse shape. A minimum was found to occur with a tolerable value of 36 g's at 3.7 msec. in the sine pulse input, and a value of 30 g's at 5.6 msec. in the triangular pulse input. No minimum occurred for the rectangular pulse curve (Figure 11).

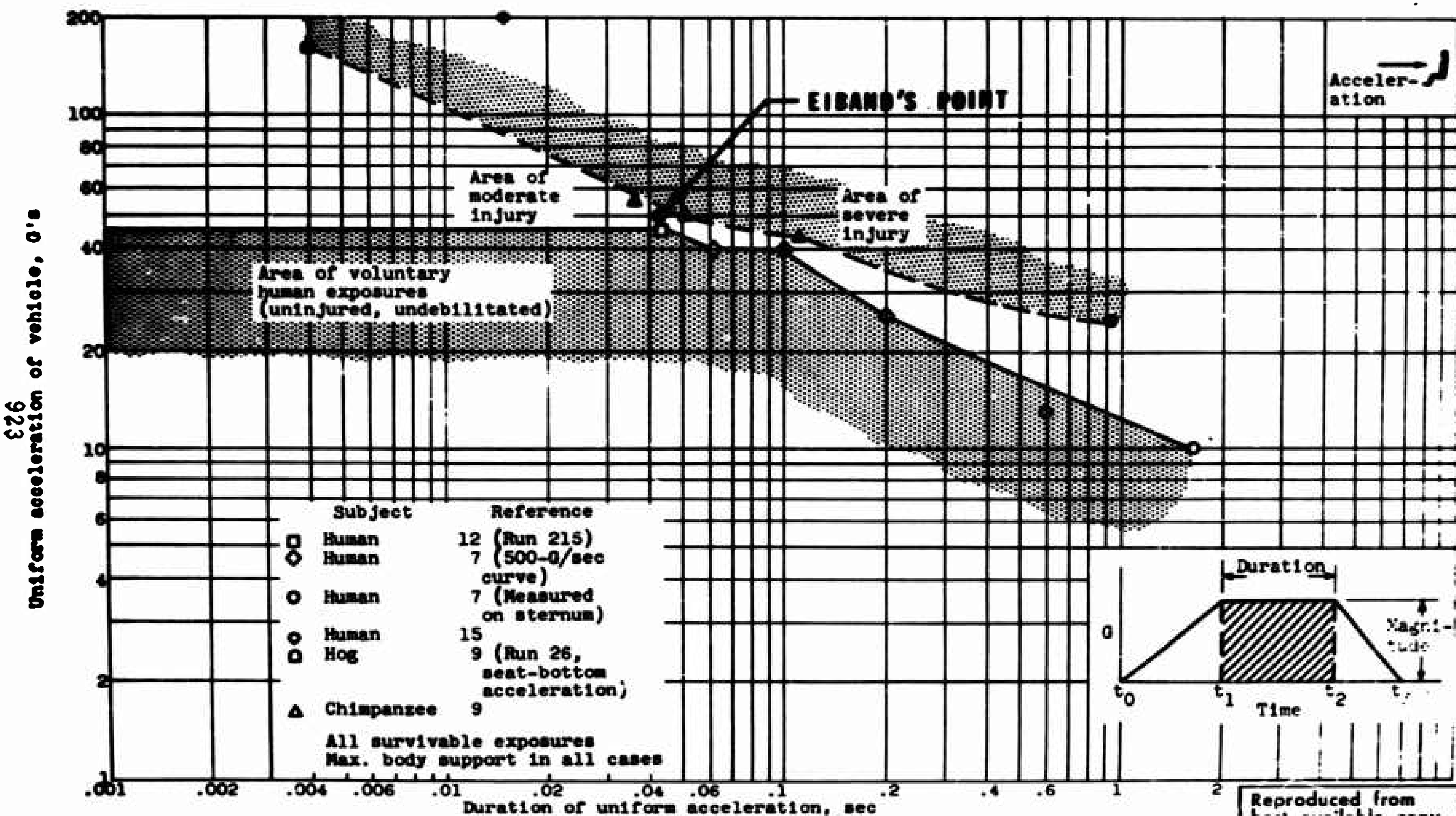

Figure 9. Eiband's Abrupt Transverse Decelerations Endured for Animals and Man

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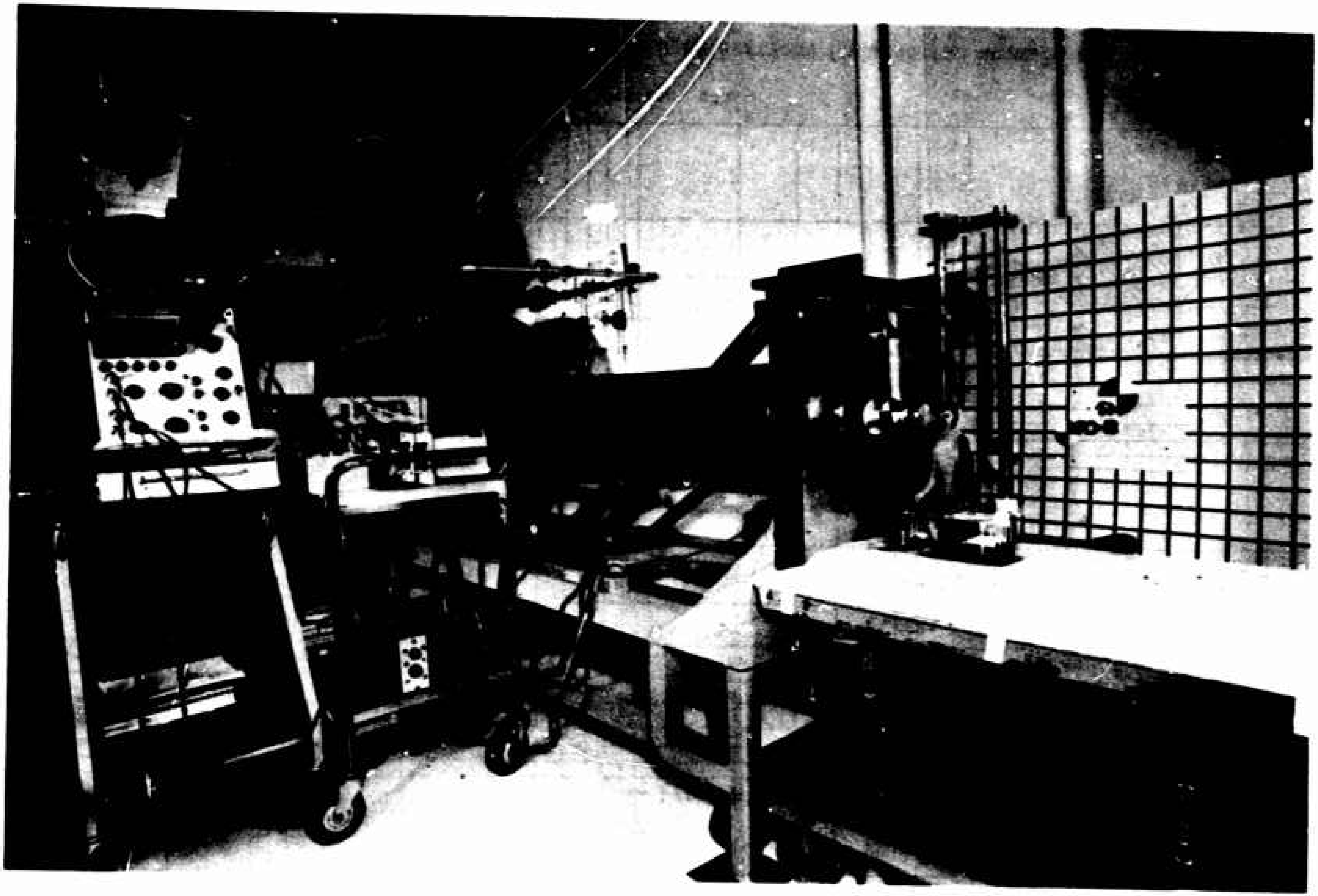

Figure 10. Squirrel Monkey Head Impact Set-Up

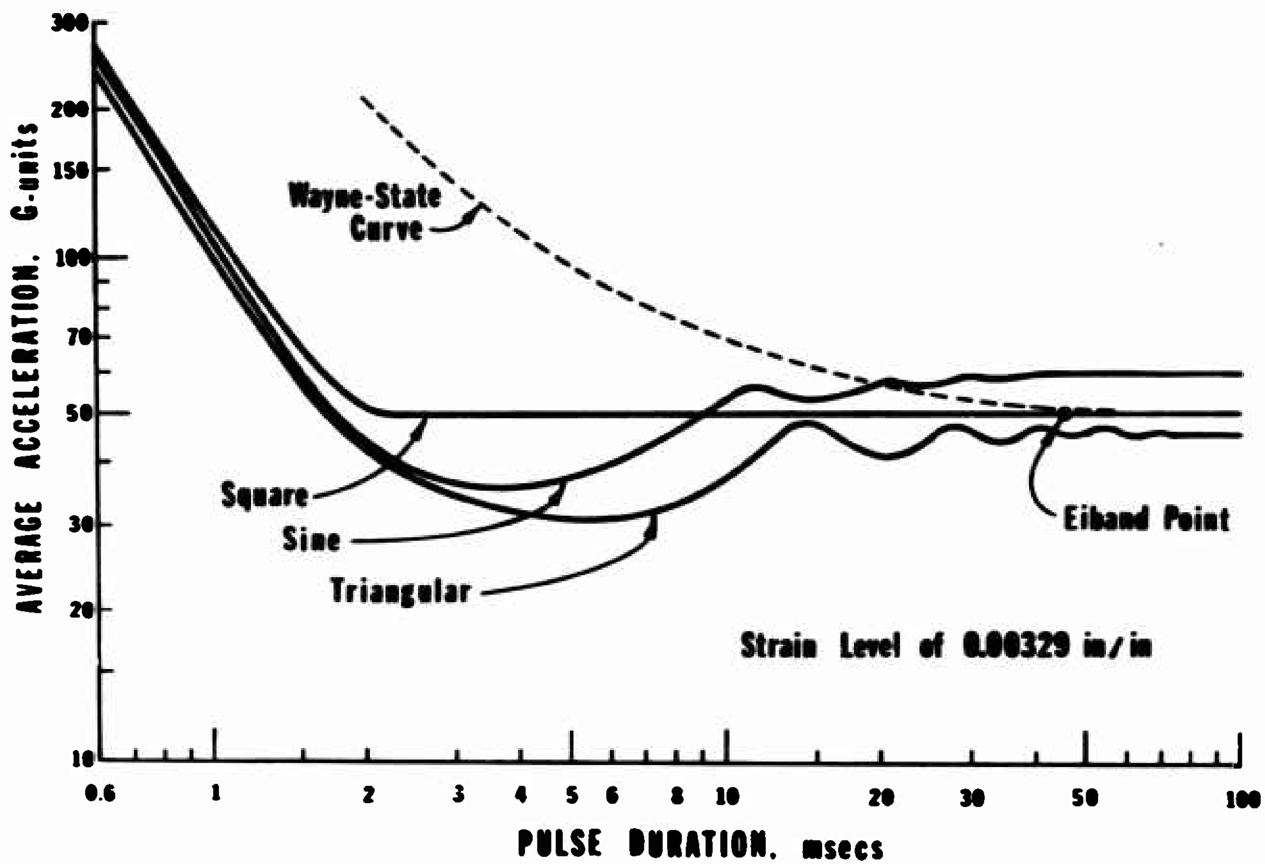

Figure 11. Maximum Strain Criterion Curves for Human Longitudinal Head Impact

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The same strain level was used to set the tolerance level for the lateral impacts. The shape of the lateral impact curves were found to be similar to the longitudinal impact curve with a shift of the time and acceleration axis. The spread was only 4 g's above 15 msec., that is, the sine flattens off at 26 g's, the rectangular at 25 g's and the triangular at 22 g's. A dip of 17 g's at 5 msec. occurred for the sine curve and a dip of 15 g's at 7.1 msec. for the triangular, with no dip in the rectangular curve (Figure 12).

The sub-human primate tolerance curves were developed for the squirrel monkey, and the Rhesus monkey from impact tolerance data. Maximum strain levels of 0.089 in/in. and 0.098 in/in. respectively were obtained (Figure 13).

CONCLUSIONS

The conclusions of this study indicates that there is a significant difference between the Wayne State Tolerance Curve and the Maximum Strain Criterion (MSC) presented here.

The shape of the input pulse significantly effects the tolerable acceleration for durations greater than 1 msec.

Since Δv is dependent on the pulse duration and shape, a sine pulse is more "tolerable", for large pulse duration, while a rectangular pulse is "tolerable" for short pulse duration.

For pulse durations less than 1 msec the shape and location of the MSC curves are independent only on the model constants.

The human lateral MSC curves were found to be 50% lower than the human longitudinal MSC curve.

The squirrel monkey lateral MSC curve has the highest tolerance level with the Rhesus monkey next. The tolerable strain level for both animals were found to be quite similar. Thus atleast for these two sub-human primates the MSC tolerance concept holds.

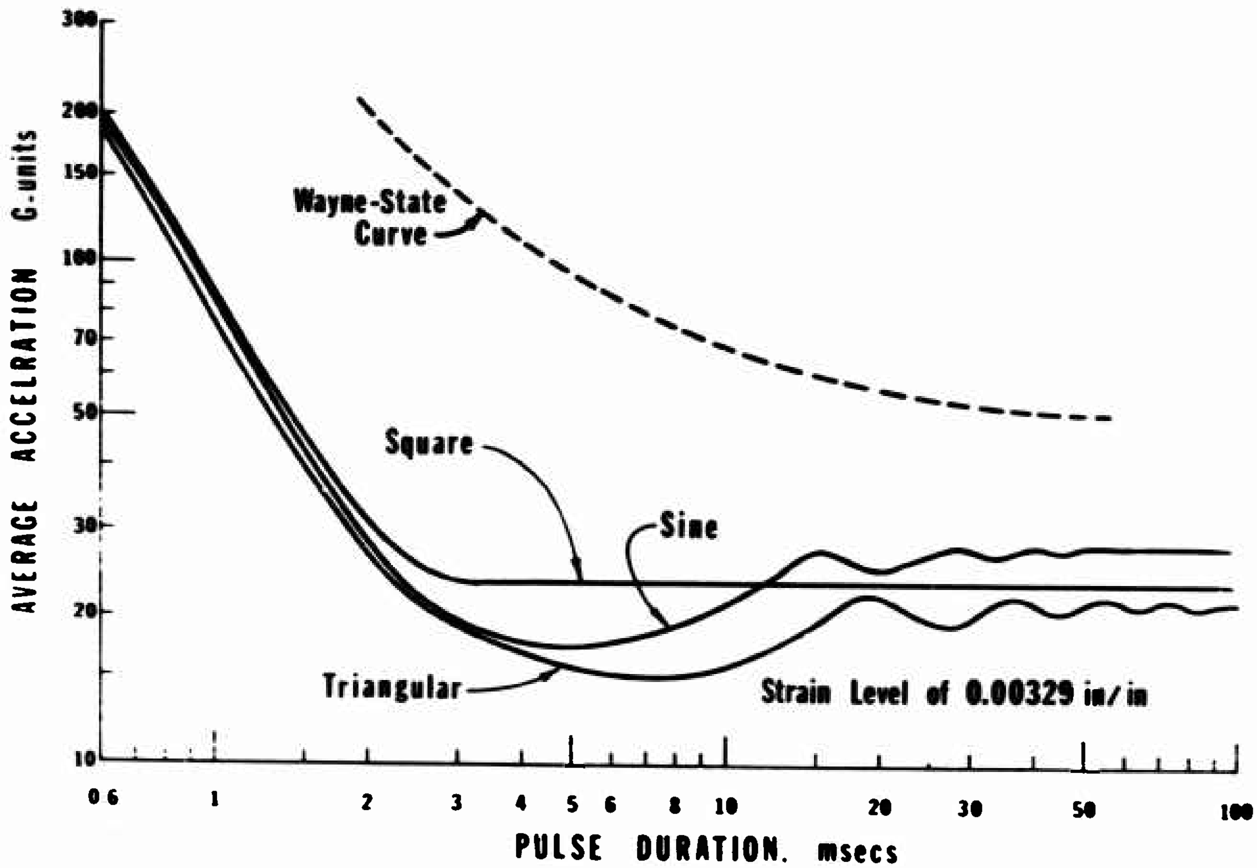

Figure 12. Maximum Strain Criterion Curves for Human Lateral Head Impact

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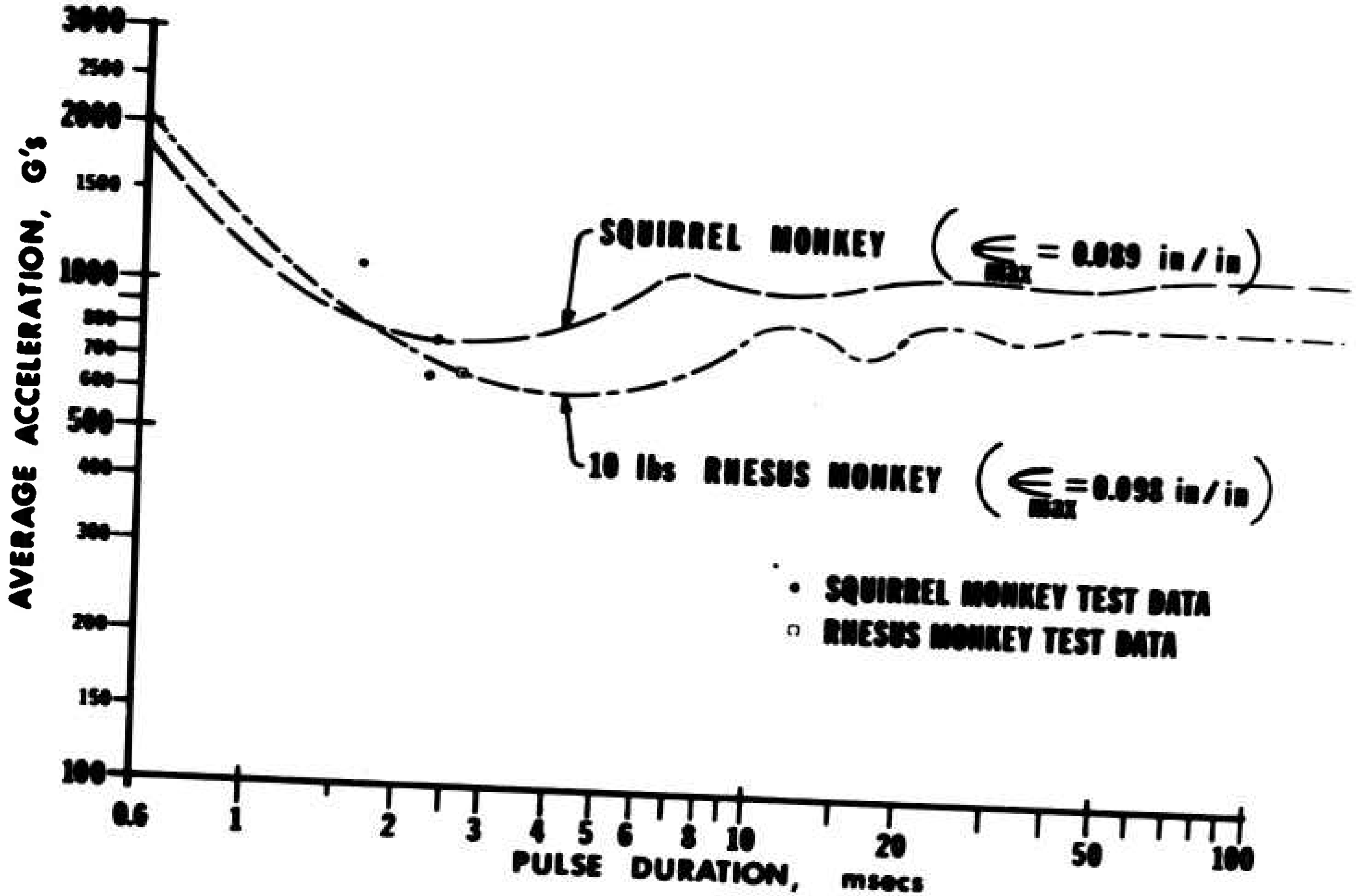

Figure 13. Maximum Strain Criterion Curves for Sub-Human Primates Lateral Head Impact

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NONMENCLATURE

a = acceleration, ft/sec^2

c = damping constant, lb-sec/in.

F = force, lb

i = $\sqrt{-1}$

k = spring constnat, lb/in.

m = mass, slugs

t = time, sec.

v = velocity, ips

x = displacement, in.

z = mechanical impedance, lb-sec/in.

$\dot{\varepsilon}$ = strain rate, 1/sec

σ = stress, lb/sq in.

ϕ = phase angle, deg.

ω = frequency, rad/sec

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