NBSIR 76-1107

Considerations in Developing Test Methods for Protective Headgear

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I. Introduction

Protective headgear has long been used to reduce injury in such activities as construction, mining, fire fighting, motorcycling, and sports (including football, baseball, and hockey). In addition, the use of protective headgear is becoming more widespread among some consumers who were traditionally not associated with helmet wear (e.g., bicycle riders).

Many manufacturers respond to this demand. Even for helmets designed for the same activity there are large variations in style, materials, comfort and cost. There are at least 10 manufacturers who offer more than 50 different models of football helmets alone. The prospective purchaser of this equipment, whether football coach, fire chief, industrial representative, or individual consumer, is faced with the monumental task of choosing among the helmets offered for a particular activity. To further complicate this decision, the most important variable, the degree of protection offered, is the least well understood and the most difficult to evaluate.

There exist many test methods for evaluating the protective characteristics of the headgear, and standards have been proposed to maintain minimum performance levels. Some of these are promulgated by authorized government agencies (National Highway Traffic Safety Administration, Occupational Safety and Health Administration, Law Enforcement Assistance Administration); some are promulgated by long existing voluntary standards organizations (American National Standards Institute, American Society for Testing and Materials); others are promulgated by groups established for specific purposes (National Operating Committee for Standards in Athletic Equipment). These standards are presently in various stages of completion. It is not unusual to find more than one test method for the same type of protective headgear. Sometimes helmets rejected by one method will be passed by another.

These difficulties are in part due to the limited information which is available on head injury, especially internal head injury. Another factor is that the relationship between a test method for protective headgear and the amelioration of head injury has not been satisfactorily explored. Also the importance of simulating the injury environment in the test method is not well established.

The purpose of this report is to summarize the difficulties in the evaluation of protective headgear and to suggest the existence of a relationship between test method and injury prevention. First, the state of the art concerning the mechanisms of head injury and the relationship of these injuries to the modes of energy input are briefly reviewed. Next the general guidelines in the design of protective headgear are discussed and present test methods for helmets are summarized. In order to provide a quantitative framework, a simple model which incorporates many features of present test methods is defined and executed. A rational approach to the design of meaningful test methods for protective headgear is duscussed, with reference to the results of the model.

II. Mechanisms of Injury

Injuries

A cross section schematic of the head and its contents is shown in figure 1. The brain, a viscoelastic, inhomogenous, and anisotropic material is encased in the skull which is relatively rigid except for the opening through which the spinal cord protrudes, the foramen magnum. The geometry of the skull in the frontal regions is marked by small curvatures and bony protuberances in contrast to the nearly spherical top and rear (occipital) regions. For more details, the reader is referred to references 1/ and 2/.

The types of injuries that are likely to occur when the head is impacted have been well documented 1/- 3/, and only general categories will be mentioned here. Among the ways to classify head injury is by the type of material damaged: namely, skull, blood vessels, and brain matter. Skull fracture is the structural failure of the cranial bone, usually due to the process of crack initiation and growth when tensile stresses arise during impact. Linear fractures, for which the resulting skull geometry is unchanged except for the appearance of a crack, are the most common 4/. Internal bleeding, when it accompanies head trauma, may be found either within the brain tissue (e.g. petechial hemorrhage) or between the brain and skull (e.g. subdural hematoma). Nearly all such injuries are potentially fatal and require surgical intervention 5/. The injuries to brain material are usually divided into contusions and lacerations. Contusions are found on the outer surface of the brain and are marked by a discoloration of the brain material; a laceration is a tearing of brain material often found in the regions of the bony protuberances of the skull 1/. These injuries may occur separately or may accompany one another.

When the impairment of function is immediate, the injury is usually diagnosed as a concussion. There are about as many definitions of concussion as there are authors on the subject. Thomas 6/ reported that concussion is - "characterized by immediate and transient impairment of neural function, such as alteration of consciousness, disturbance of vision, equilibrium, etc., due to mechanical forces." 7/ He restricts his attention to cases of unconsciousness and suggests that the ultimate cause is mechanical input to the brain stem axis (by stretching or shear strain). However, Ommaya 8/, quoting from Walker 9/ argues that "concussion involves not only the brain stem but other loci in the brain." He then goes on to say, quoting now from Symonds 10/:

> "Concussion should not be confined to cases in which there is immediate loss of consciousness with rapid and complete recovery, but should include the many cases in which the initial symptoms are the same but with subsequent long continued disturbances of consciousness, often followed by residual symptoms - concussion in the above sense depends upon diffuse injury to nerve cells and fibers sustained at the moment of the accident. The effects of this injury may or may not be reversible."

In this sense the alteration of consciousness known as concussion will include those cases in which the diffuse disturbance of functional damage is accompanied by structural failure which may be irreversible.

Relation between gross head motions and tissue level injuries

It is important to distinguish between the local disturbance of function or structure and the gross head motions which accompany these local disturbances. These local disturbances are always due to material deformation which may be quantified by field variables such as strain, stress, or pressure. The local deformation can be related to the gross motion of the skull. Following an impact, three distinct motions occur* (figure 2):

1. Skull bending

2. Translational component of the acceleration of the center of gravity.

3. Rotational acceleration of the head about the center of gravity.

The latter two modes of motion are inertial and may result without impact to the head (say in the case of whiplash). The skull bending mode results from the contact phenomena between a foreign object and the scalp, transmitting a local force distribution to the skull.

Each of these modes give rise to local deformations which, depending upon the magnitude, are capable of producing internal head injuries. Only the skull bending mode contributes to skull fracture. Many experimental and analytical studies have attempted to understand the mechanisms by which the energy of the gross motions is transferred to the local tissue level. In the rest of this section, some of these results will be briefly mentioned.

While all modes are operative in most impact situations, the experimental and mathematical model studies tend to consider the rotational motion separately. In many animal impact studies the translational component of acceleration has been singled out as the most important variable for measurement. Accordingly, most tolerance criteria for head injury have been referred to the translational acceleration.

The mechanism for tissue damage by the direct impact mode (skull bending and translational acceleration) has been qualitatively explained as follows: when the skull is impacted, pressure gradients arise in the fluid-like brain (as in any accelerated fluid-filled container) 11/. Positive pressures develop in the brain at the site of the blow, and negative pressures (normal tensile stresses) develop at the opposite side of the brain (contre coup site) 12/. Transient pressure waves in the brain and stress waves in the skull also tend to magnify the amplitude of the negative pressures 13/. It was postulated that the large negative pressures, which appear to exist at the contre coup site, may cause cavitation in blood or brain tissue 11/-14/. However, contracoup injuries are often observed for occipital impacts but rarely for frontal impacts 15/, a descrepancy which cavitation theories often fail to resolve 16/.

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*If the direction of the blow does not pass through the neck axis, there will also be a rotation about this axis.

In studies with water-filled cadaver skulls, it was shown that the presence of the foramen magnum enhanced the resulting pressure gradients 11/. It has been suggested that the established pressure gradients may force brain material through the foramen magnum, causing shear stresses to develop in the region of the brain stem 6/. Other displacements of brain material have also been observed to occur 11/, 17/, 18/. Locations where these relative motions are large are thought to be potential sites of injury. In particular, Gurdijian suggests that these "relative motions seem to be the most likely mechanism for frontal and temporal lobe contusions rather than negative pressures" 11/.

Figure 3 shows the Wayne State University Concussion Tolerance Curve, first published in 1962 19/. These data were based on cadaver skull fracture in the short duration range (because of the clinical association of linear skull fracture and concussion) and on human volunteer studies for impacts of long duration. It is seen that the average translational acceleration required to produce the injury depends upon the time interval over which the acceleration is applied. Haut et. al noted that such behavior is to be expected for viscoelastic material like the human brain 20/. While the short-comings of this tolerance curve have been rigoriously discussed 21/- 23/, it remains the most widely accepted compendium of concussion tolerance.

Several indices of injury potential have been derived from the Wayne State data, based on the near linearity of the data when plotted on log-log paper. Gadd developed the severity index,

$$SI = \int_0^T a^{2.5} dt$$

where a is the acceleration expressed as a multiple of the gravitational acceleration and T is the duration of impact. According to Hodgson 4/:

> "Gadd derived a critical value of 1000 (the units for SI are seconds) as being the threshold of danger to life. . . He recognized that the Severity Index may be a function of contact area. . . Later on, Gadd suggested a critical value of SI equal to 1500 . . . for distributed impact" 24/.

Arguing that the SI does not really match the Wayne State data, Versace 24/ developed the head injury criterion:

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

where t_1 and t_2 are those times between 0 and T for which the above expression is maximized. Again, "a HIC level of 1000 is assumed to be a threat to life" 4/.

Other tolerance criteria have been based upon lumped-parameter models of the head 22/, 26/, 27/. One example, the maximum strain criterion (MSC), is shown in figure 4a 26/ where parameters were chosen to match the impedance response of animal or cadaver heads. The maximum strain in the model was related to the onset of discernable structural damage in experimental animals. "The tolerable head strain for humans (found to be .006 in/in) was calculated from the mathematical model" 27/based on scaling factors that were developed. An acceleration tolerance curve was developed and compared to the Wayne State curve (figure 4b). For side impacts a different tolerance curve was indicated 28/.

Ommaya and his associates 16/, 29/-34/ have conducted a large part of the experimental research to identify the contribution to injury by the rotational component of acceleration. This mechanism of injury, originally proposed by Holbourn 35/, involves the transfer of shear stresses to the viscoelastic brain by the rotating skull.

In whiplash experiments with monkeys, rotational accelerations were measured and tolerance curves for concussion were derived 29/, 30/ (see figure 5). It was found that the tolerance levels were reduced if the resulting head rotation was produced by direct impact, and suggested that "skull distortion effects appear to reduce the level of head rotation required for the production of cerebral concussion and contusions" 16/. When animals wearing protective helmets were impacted, no concussions were observed at the same rotational levels 16/. Thus it was thought that the gross head motion of skull bending contributed to the shear stress mechanism of injury 29/. Reasoning from a scaling law, also attributed to Holbourn (in which tolerable rotation levels for different species are inversely proportional to the brain mass raised to the 2/3 power), the primate data was extrapolated to humans to obtain tolerances of rotational acceleration for long duration impacts:

7500 rad/sec^2 for concussion of 99% of population
1800 rad/sec^2 for concussion of 50% of population

In another series of experiments 31/, 32/ an attempt was made to isolate the two inertial gross motions. By rigidly securing primate heads in pots, the heads could be subjected to purely rotational or translational motions (the axis of rotation was not the center of gravity of the head; therefore, a component of translational acceleration was present, which was comparable to that of the translated animals). All rotated animals suffered concussions, while none of the translated animals were concussed. On autopsy it was found "that a greater number of lesions occurred in a more diffusely widespread symmetrical manner in the rotated group, whereas only a few asymmetrically placed focal lesions developed in the translated group" 8/. This does not suggest that such focal lesions are not dangerous (all are potentially fatal) but rather that they occur without concussion (as diagnosed immediately) in the purely translated group.

To further quantify concussion, functional damage was assessed by a somatosensory evoked response (SER) method 32/, 33/: "all animals in the rotated group exhibited neurological evidence of experimental cerebral concussion; in contrast, none of the translated group showed this effect." At the local level, similar changes in the SER of individual nerve fibers were observed under mechanical loads 34/.

Mathematical modeling of head injury

The usefulness of mathematical models lies in their ability to quantitatively relate the gross head motions to local tissue deformation. The effect of the various input modes can be isolated and studied. Of course, they are limited by how well the material properties of the skull and brain can be established. At this time the mechanical properties of the skull (see e.g. 36/, 37/) are more reliably known than those of the structurally more complicated brain (see e.g. 38/, 39/).

If, in addition, experiments can establish tissue level failure criteria (including functional failure, say by experiments similar to those of Thibault 34/ described earlier), then the models can be used to establish tolerance limits for the gross head motions. This procedure would be as follows:

1. Perform experiments on tissues that relate the severity of damage to the history of the local deformation. This relationship, expressed in functional notation* is

   $$S_T = \mathcal{G}(D)$$

   where D is the deformation tensor.

2. Use the model to derive another functional which relates severity to the history of the gross head motion X (itself a function of position and time):

   $$S_G = \mathcal{J}(X)$$

Two examples of these functionals, SI and HIC, were discussed earlier. They relate the severity potential to an integral of the translational component of the acceleration. In the MSC approach 26/ the acceleration criteria is derived from a given injury index, mean strain. However, this relationship is between two gross modes (translational acceleration and skull inbending) instead of between a local injury mechanism and a gross head motion.

The mathematical models to be discussed can be categorized according to whether or not skull rotation effects are included. Some of the studies, which do not include rotation, have modelled the brain as an inviscid fluid 41/ - 47/ in order to identify areas which are likely to sustain large pressure changes (both positive or negative). They are also useful in predicting sites of the skull at which fracture is likely. These models range from the

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*A functional is an operation which maps histories of one variable onto a one dimensional space (see, e.g. Truesdell 40/). The simplest case is where only the present value of the variable is important, in which case the functional reduces to a simple function. Therefore, for skull fracture, where failure occurs when the tensile stress exceeds a certain limit, the local severity index would reduce to the maximum value of the tensile stress.

rigid skull model of Guttinger 41/ to the sandwich skull model of Akkas 44/ and include the effects of non-sphericity 45/ and glancing blows 46/. The finite element model of Shugar 15/ incorporates many complicated structural features, including the foramen magnum. Goldsmith, using a physical model, accounted for the effect of the neck 47/. Because of wave propagation effects, these models predict large values of negative pressures at or near (depending upon the sophistication of the model) both poles (impact side and opposite site). However, in all of these studies, the duration of loading is very small (one millisecond or less), a situation which magnifies the importance of the transient fluctuations 48/. Advani and Owings 49/ have predicted that the impact duration of the skull against a rigid flat surface is at least 3 msec.

Other investigators have accounted for the low shear modulus of the brain material 41/- 48/ in order to identify sites of large shear strain even in the absence of rotation. In these studies, the pressure distribution results were similar to the inviscid studies described above. Owings and Advani 49/, 50/ with an elastic brain model, predicted large values of shear strain at both the midbrain area and the brain surface. When viscous damping is accounted for 48/, 51/, the magnitude of the shear strain decreases and the locations shift slightly. Hickling and Wenner 48/ predicted shear strain values of about .025 rad for a 3 msec pulse with an average acceleration of about 150g. This is on the same order as the shear strain for structural failure, .035 rad 38/.

In the above studies, longer loading durations were considered (up to 20 msec). As the duration increases, the magnitude of both the negative pressure and the shear strain decreases. This result would be contradictory to the Wayne State Tolerance Curve (figure 3) if only the magnitudes of these quantities were responsible for brain injury. However, the results of these investigations also show that at the local level, there is also an increase in the time duration for which negative pressures and shear strains exist. Therefore, it is likely that the injury has some history dependence at the tissue level itself.

In rotation studies the skull is modelled as a rigid sphere which is subjected to prescribed rotational accelerations, either step functions 52/, 53/ or finite pulses 54/- 57/. When the brain is modelled as an elastic material, Holbourn's 2/3 power scaling law, discussed earlier, was confirmed 54/. If viscous damping effects are included, the magnitude and transient fluctuations of the shear strain are reduced 55/, and the 2/3 power scaling law applies only approximately 54/. Bycroft picked a specific brain location for concussion injury (non-dimensional radius of 0.3) and calculated the value of the maximum strain required for correlation with experimental whiplash studies. For both squirrel and rhesus monkeys, this strain value was .05 rad. Using this same location and shear strain level for human concussion, a threshold level for concussion was calculated to be 3500 rad/sec^2. Liu, Chandran, and von Rosenberg 55/ obtained similar results using a finite difference procedure. Ljung's model 56/ accounted for the effect of the falx, the near rigid portion which separates the two hemispheres of the brain. Löwehhelm 57/, using the results of this model, calculated a tolerance curve based on a local injury criterion which was strain rate dependent.

III. Headgear

The preceeding summary of injury mechanisms is useful for suggesting performance criteria for protective headgear. The experimental and analytical studies confirm that all three modes of the gross head motion are capable of producing local deformations which are potentially injurious. At this time, no single mode can be identified as the most harmful. The tolerance values for these gross motions, where they exist, are quite sketchy, but some indications are clear. The dependence of the injury potential on the history of the gross motions (effect of input duration) is a factor, especially for impacts of shorter durations. The translational acceleration indices, SI and HIC, account for this effect and should serve as guidelines until the state of the art improves. Though analogous rotational indices have not been suggested, the necessity of reducing rotational accelerations is obvious.

One approach in the design of protective headgear is to attempt to reduce the gross head motions by placing deformable material between the head and the prospective impact surface. Analogous to the simple spring-mass model of Appendix 1, this is expected to change the inertial response even if no energy is absorbed by the helmet material. In this analogy, the percentage increase in the duration of the acceleration response is the same as the percentage decrease in the average (and maximum) acceleration. According to the injury criteria for which the acceleration is weighted to the 2.5 power, this changed response is favorable. Any energy absorbing features of the helmet system would serve to further improve the response of the head. Another purpose served by these deformable materials is to spread out the applied load over a large area, thus reducing the gross head motion of skull bending*.

Several studies have listed some of the general parameters of a helmet system with qualitative descriptions of their relationship to head protection 19/, 58/-61/. More recently, simple one dimensional models of head/helmet systems were used as a guide in optimizing helmet design parameters 62/-65/. In these studies the helmets were modelled as Kelvin elements, a spring and a dashpot in parallel. In Liu's model the helmet element was coupled to a fluid filled cylinder and the injury criteria was "the averaged time spent, at contrecoup, beyond the cavitation pressure of the fluid" 63/. McElhaney et al 65/ coupled the helmet element to the two-mass head model described earlier 26/ and used the maximum strain criterion for injury.

This paper is not so much concerned with proper design aspects of protective headgear but rather with methods of evaluating whether or not a helmet offers suitable protection. In an ideal helmet performance test, one would subject a "head" to a realistic impact load and measure the response of the head to determine whether this response is within tolerable limits. The problem of determining which responses to measure has already

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*As there are few reports of skull fracture among helmet wearers, it is assumed that most helmets sufficiently reduce the skull bending mode. Therefore, the question of tolerances for skull fracture has been avoided in this report, though this particular injury is the most well understood. For a complete list of references, see Mahajan 3/.

been addressed. While it would be best to measure local deformation quantities which are related to injury at specific sites, the present state of the art only suggests guidelines for tolerance of the gross head motions. Nevertheless, this limitation has not slowed the development of helmet test methods.

The "head" to be used in a performance test is another source of concern. Ideally the best head-form would incorporate enough essential features to produce a response similar to the human head. Efforts in the development of humanoid head-forms have been reported 66/- 70/. The resilient components of these headforms were adjusted so that the response would fall within a range of static and dynamic (linear acceleration) responses of cadaver heads 66/, 67/. Other comparative tools such as impedance response 69/ and frequency analysis 70/ have also been used. However, there remain questions as to the reproducibility and durability of these headforms 71/. In most test methods, therefore, a metal headform is used.

The ability to simulate a realistic impact situation is another desirable feature of an ideal performance test. In nearly all helmet test methods, a helmeted headform is dropped onto an impact surface. The rationale for this method is based on two observations 72/: 1) in laboratory experiments, cadaver head responses were observed which were nearly independent of the body, and 2) the duration of most impacts is much less than the neck muscle reaction time. Present test methods incorporate a wide variety of impact surfaces and drop energies even in evaluating helmets for the same activity. If the purpose of the test is to insure that potentially injurious head responses are avoided, then it is necessary that the impact conditions of the test be such as to simulate (or at least be related to) a real life impact situation. In many activities, one impact mode predominates, but in some (e.g. football) there are several equally important modes.

To illustrate the variety in test methods for protective headgear, a partial list of present methods is shown in table 1. (See references 72/- 81/.) Most methods attempt to account for the time duration effect (that the head can withstand higher accelerations for short times; see Wayne State curve, figure 3) by the "dwell time" concept (where limits are set for the time duration for which the acceleration exceeds prescribed values). It is interesting to ask how these criteria compare to a biomechanical injury index like SI. As an example, consider the criteria of the FMVSS 218 standard. For simple acceleration traces (e.g. curves which approximate haversines or triangles), a curve representing a barely safe acceleration response is completely defined by specifying any two of three rejection criteria (figure 6). The corresponding SI values, also shown in figure 6, are all extremely high and depend upon which rejection criteria are specified. Of course, such a comparison may not be relevant as these methods often do not claim to represent realistic impact situations. Rather they are one means of ranking helmets in one impact configuration.

The possibility that this ranking would change if resilient headforms and/or impact surfaces were substituted for their metal counterparts is usually not addressed. In one study, for which the same helmets were tested by two different methods, there was little or no correlation in helmet performance as measured by the different methods 82/. In addition to these differences in test method (headform, impact surface, impact energy, and rejection criteria), other effects may also influence the proper ranking of helmets. E.g. would the helmets which perform best in a typical drop test method also offer the best protection against head rotations? Does the size of the impacting object make a difference?

IV. Model

Description

In order to address the questions raised in the previous section and to provide a quantitative framework for comparing one test method with another, a simple one dimensional, lumped parameter, model is utilized. The headform, helmet and impact surface are all identifiable elements and a broad range of test conditions can be studied. The scope of this work differs from previous modelling attempts 62/, 67/, discussed earlier, in that our purpose is not to identify optimum values of helmet parameters for preventing a particular injury. The simplistic models for the helmet and head and the limitations in the available tolerance data would severely reduce the applicability of such an exercise. Rather we are interested in comparing headform responses (acceleration vs. time) when test conditions are changed. The model is shown in figure 7, and each of the elements are explained in turn.

The headform is modelled as either a rigid body, to represent metal headforms, or as the two-mass resilient system suggested by Stalnaker et al 83/ (figure 4). The force F_H between the masses is given by (see figure 7):

$$F_H = K\left[L - (X_1 - X_2)\right] + C(\dot{X}_2 - \dot{X}_1) \qquad (1)$$

where L is the initial separation of the two masses. The parameters of this model were evaluated to match the driving point impedance characteristics of cadaver heads (groups 1 and 2 of table 2) 26/. In these studies the mass M_1 was subjected to prescribed accelerations, and tolerance curves were developed to correspond to threshold strain levels. However, in impact simulations, the mass M_1 makes contact with some foreign object (possibly the inside of a helmet) and accelerations are recorded for the mass M_2 (either by placing an accelerometer at the center of gravity, for humanoid headforms, or opposite the impact site, in cadaver experiments).

In order to determine whether these parameter values give reasonable impact accelerations, the acceleration of mass M_2 in the model was compared to the response of cadaver heads in an experimental impact situation. McElhaney et al 27/ reported the resulting accelerations of cadaver heads which were impacted by a flat rigid object of known mass (10 kg) and velocity (11 m/s). Humanoid headforms have been developed to match the acceleration responses in these cadaver experiments 67/. When the head model was subjected to the same input conditions (equations of motion and initial conditions to follow), the maximum acceleration, A_{max}, was much greater than in the cadaver experiments, and the time, t_{max}, at which the maximum occurred, was much sooner. These quantities, a_{max} and t_{max}, could be matched when the parameters in group 3 (table 2) were used in the model*. Slattenschek and Tauffkirchin 23/ using a one degree of freedom model also chose a similarly low value for the spring constant. Also static force-displacement curves of laterally loaded cadaver heads (and a representative humanoid headform) suggested that the value of an effective spring constant would lie in the range 0.7 - 3.5 MN/m 67/. Therefore, in most of this work, the humanoid headform will be represented by the parameter values in group 3 of table 2.

Both rigid and resilient impact surfaces were simulated. In the latter case the deformation of the surface was assumed to be quasi-static**. Then the relation between the force, F_s, and the displacement can be derived from the contact relations between elastic bodies 85/,

$$F_s = B(X_3 - d)^{3/2} \qquad (2)$$

(see figure 8) where d is the initial thickness of the helmet liner material and

$$B = \frac{4}{3}\left(\frac{\bar{R}}{(K_1 + K_2)}\right)^{1/2}$$

$$\bar{R} = \frac{R_1 R_2}{R_1 + R_2}, \quad K_1 = \frac{1 - \nu_1^2}{E_1}, \quad K_2 = \frac{1 - \nu_2^2}{E_2} .$$

E, v and R are the elastic modulus, poisson's ratio, and radius of curvature respectively for each body (subscript 1 for impact surface, subscript 2 for helmet). For an impact surface made of polyurethane, E_1 = 28 - 63 MN/m^2 (4000 - 9000 psi) 86/. This is much smaller than the modulus for typical helmet shell materials like polycarbonate, $E_2 = 6.9 \times 10^3$ MN/m^2 (10^6 psi) 86/. For a flat pad, $R_1 = \infty$, and helmet radius of order 10 cm, a range of values for the parameter B can be calculated:

$$B \sim 10 - 25 \text{ MN/m}^{3/2}$$

Melvin and Roberts 87/ report the results of experiments with typical helmet liner materials which were loaded at high strain rates. The visco-elastic nature of these materials is evident. To simulate these deformations we will consider the simplest viscoelastic lumped parameter elements: the

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*The masses, M_1 and M_2, were arbitrarily chosen to be the averages of the side and longitudinal cases. K and C were chosen so that a_{max} and t_{max} would fall within the range of cadaver responses.

**According to Goldsmith 84/, such approximations are valid if the duration of contact between the impacting objects is long compared to the time that it takes for elastic waves to traverse the objects. Wave speeds in typical impact surface materials like polyurethane are sufficiently large to allow this approximation.

Maxwell element, a spring and dashpot in series; and the Kelvin element, a spring and dashpot in parallel. The equations for the force in these elements (derived in Appendix 2) are (refer to figure 8):

$$\text{Maxwell:} \quad \dot{F} = EA\left(\frac{\dot{X}_1 - \dot{X}_2}{d}\right) - \frac{E}{\eta} F \tag{3a}$$

$$\text{Kelvin:} \quad F = EA\left(1 - \frac{X_3 - X_1}{d}\right) + \frac{\eta A}{d}\left(\dot{X}_1 - \dot{X}_3\right) \tag{3b}$$

where E and η are the elastic modulus and viscosity for the helmet liner material, A is the area over which the load is distributed (see Appendix 2), d is the initial thickness of the helmet liner material, and a dot above a symbol indicates differentiation with respect to time. It was found that the Maxwell model is more capable of reproducing the load-deflection curves that were reported by Melvin and Roberts 87/ (Appendix 2) and therefore this model will be used more extensively in this work. This is in contrast with other previous one dimensional studies 62/, 65/ in which helmet materials were modelled by Kelvin elements. The range of values for the parameters E and η were chosen to represent a reasonable range for helmet materials based on the work of Melvin and Roberts 87/ (see Appendix 2). Therefore, in this model, the characteristics of a single helmet are described by a pair of values, (E, η).

The governing system of equations is completed by force balance relations for each of the masses (figure 7):

$$M_1\ddot{X}_1 = F_H - F \tag{4}$$

$$M_2\ddot{X}_2 = F_H$$

For a large rigid impact surface ($X_3 = d$, $\dot{X}_3 = 0$) the system (1), (3a), (4) and (5) represents four equations for four unknowns X_1, X_2, F, F_H. The initial conditions are $X_1(0) = 0$, $\dot{X}_1(0) = V$, $X_2(0) = -l$, $\dot{X}_2(0) = V$, $F(0) = 0$, where the velocity V is related to the drop height h by $V = \sqrt{2gh}$.

If a resilient impact surface is to be included, $F = F_S$ (neglecting helmet mass) and equation (2) becomes the fifth equation for the now unknown position X_3. For the case of a small rigid mass M_3 impacting the head another equation must be added

$$M_3\ddot{X}_3 = F$$

and the initial conditions on velocity must be changed to

$$\dot{X}_1(0) = \dot{X}_2(0) = 0, \; \dot{X}_3(0) = -V$$

Note that the mass of the helmet is neglected in this system. Preliminary experiments indicate that the response of the headform is nearly independent of the helmet mass. It is likely that this is the reason that present test methods do not include the mass of the helmet when calculating the input energy. The rationale for neglecting the helmet mass is as follows: only a small fraction of the helmet mass is contained within the area of contact; the potential energy associated with the rest of the helmet mass is not available for compressing the liner material and instead is dissipated by flexural wave motions of the helmet material outside of the contact region. This latter behavior was observed in high speed motion pictures 88/.

These equations were solved numerically on a Univac 1108 computer using a modified Runge-Kutta method as outlined by Gill 89/. The accuracy was verified by comparision to simple cases for which the exact solution was easily determined and by computing the same results when time step was reduced by an order of magnitude. The output variable of most interest is the acceleration of the headform (the mass M_2 when the resilient headform is used in the model) expressed as a multiple of the gravitational acceleration. Once the acceleration history was obtained, the injury indicies, SI and HIC (see section on mechanisms) were computed numerically by fitting the acceleration data with cubic splines 90/. The accuracy of this latter procedure was verified by comparing the results to that obtained with simple acceleration curves for which SI and HIC could be calculated analytically. The computation runs were terminated when the acceleration dropped below zero.

In the proposed model, a test method is considered to be a combination of headform, impact surface, drop height, and a parameter for evaluating performance. The simulations for all of these elements are intentionally simple. It is assumed that enough detail is included to indicate trends when comparing one test method to another. The results are expected to be useful in guiding future experiments and are not to be considered conclusive by themselves.

Results

Variations in each parameter were referred to the following reference system:

Headform:

Rigid: M_1 = 4.54 kg (10 lb)

Resilient: M_1 = .23 kg (0.5 lb)

M_2 = 4.31 kg (9.5 lb)

K = 1.75 MN/m

$C = 7.7 \times 10^2$ N-sec/m

Helmet:

Maxwell model: E = 20.7 MN/m^2

$\eta = 2.07 \times 10^5$ poise

d = 2.54 cm

Impact surface:

Resilient: $B = 10\ MN/m^{3/2}$

Drop height: $h = 1.83$ m (6 ft.)

Each parameter was varied through a representative range while the others were held fixed at the above values. In addition, the effect of a smaller impacting mass M_3 was also computed. These results are shown in figures 8 through 16. In order to examine the effect of different test methods for a range of helmet parameters, some limited performance measures, a_{max}, SI, and HIC are listed in tables 3 to 6.

The effect of variations in the helmet dashpot parameter, η , is shown in figure 8 for several headform/impact surface combinations. It is seen that the effect of changing η depends upon the test method and becomes less important as more resiliency is added to the system. Similarly the effect of changing test method is less important for the smaller values of η. Generally, it will be seen that the effect of changes in any one parameter decreases in importance as resiliency is added to other parts of the system. These indications are independent of the parameters chosen for measuring performance. However, it is seen that HIC is more sensitive than a_{max} to changes in η (although not shown in the figures, SI is about 10% higher than HIC and behaves similarly).

Similar results are shown in figure 9 for variations in the helmet spring parameter, E. The same trends apply as described above for η. Note that changes in this parameter seem to have a smaller effect on the acceleration response; i.e. percentage changes in E are less noticible than a similar change in η. Also, for changes in E, HIC is less sensitive than a_{max}.

In figures 10, 11 and 12, the effects of resilient headform parameters, K, C, M_1 and M_2, are shown for both rigid and resilient impact surfaces. As expected, the effect of variations in each parameter is more noticible for the rigid impact surface. Of these headform parameters, the acceleration response is most sensitive to changes in the spring constant, K. For values of K above 3.5 MN/m the response is not very different from that of the rigid headform.

The effect of the resilient impact surface parameter, B, is shown in figure 13. Again variations in B are more noticeable for the rigid headform, and the effect of headform changes is more noticeable for large values of B.

Figure 14 shows the effect of drop height, h, for several headform/impact surface combinations. Again the aforementioned trends apply. Also of note is the very regular dependence of both a_{max} and HIC on h. It is easily seen that, regardless of test method,

$$a_{max}(h) \propto h^{1/2}; \quad HIC(h) \propto h^{5/4}$$

That is, the maximum acceleration (or maximum force) is proportional to the impact velocity, all other conditions being equal. Such simple relationships were not noticed for the other parameters.

In figure 15, the effect of changing the mass of a rigid headform is shown. Note that both indicies, a_{max} and HIC decrease with increasing headform mass. Considering the simple spring-mass analogy of Appendix 1, this result should not be surprising. However, the compression of the helmet liner material increases with headform mass, and there may be a greater tendency for the liner material to bottom. Bottoming aside, these results may indicate a problem for those test methods which only prescribe a value for the total impact energy, as both higher headform masses and lower drop heights tend to lower the acceleration response.

As described in the previous section, the governing equations were modified to include the case of a finite-sized missile striking a helmeted head (or headform). Figure 16 shows the effect of varying the mass of this missile while its velocity remains fixed. The response for $M_3 = \infty$ is the same as when the head impacts an infinite rigid mass. Note that as M_3 decreases, the durations of contact are reduced as well as the maximum accelerations.

In table 3, the effect of headform and impact surface are shown for a range of helmet parameters. Here it is seen that the general trends referred to earlier (figures 8 and 9) are consistent for the entire range of helmet parameters considered. If the maximum acceleration is used as a performance measure, there are several examples where the relative performance (of one "helmet" compared to another) reverses order when the test method changes. For example consider the helmets represented by the following pairs of parameters: $E = 13.8$ MN/m^2, $\eta = 3.45 \times 10^5$ poise and $E = 20.7$ MN/m^2, $\eta = 2.76 \times 10^5$ poise. The former performs better when tested with a rigid headform and rigid impact surface; as the resiliency of either test method parameter increases, there is a point at which the latter helmet performs better. These reversals occur because the quantitative effect of changes in test method is not independent of the helmet parameters. The occurance of such reversals always involves a change in both of the helmet parameters (E and η). For the examples where these reversals occur, the computed accelerations are not very different; however, it must be emphasized that only a simple helmet model is being utilized and that an increased likelihood of such reversals should be expected if real helmets were tested. Experimental evidence of such reversals in the order of helmet performance have also been reported 82/. Note also that the occurence of these reversals appears to be far less likely if the HIC or SI is used as a measure of performance rather than the maximum acceleration. In fact there is only one example of a HIC reversal in table 3. Probably, this is because these indicies tend to spread out the performance data for the most sensitive helmet parameter, η .

From table 3, it appears that a rigid headform/resilient surface combination can be found such that the performance measure closely resembles that for a resilient headform for the entire range of helmet parameters. That is, within the confines of this model, the resiliency of the impact surface can be used to compensate for the resiliency of the headform. As seen from figures 10 and 13, the effect of changing the impact surface parameter, B, is similar to the effect of changing the most important headform variable, K. Thus, the response for the rigid headform/B = 10 $MN/m^{3/2}$ system is close to that of the resilient headform/rigid surface, and the response of the rigid headform/B = 5 $MN/m^{3/2}$ system is close to that for the resilient headform/B = 10 $MN/m^{3/2}$. These indications are independent of the performance measure, and there are no examples of reversals in these cases.

Similar results are presented in table 4 for a range of spring and dashpot parameters that are applicable to a Kelvin model (see Appendix 2). This model gives unrealistically large initial accelerations (because the dashpot is in parallel) for the rigid headform/rigid impact surface and this case is not presented. The remaining cases show the same trends that were evident for the Maxwell model: 1) the occurence of reversals in the order of helmet performance when the maximum acceleration is used as a performance measure, 2) the absence of reversals when the HIC is used, and 3) the indicated suitability of compensating for the resiliency of the headform by the resiliency of the impact surfaces. It is also of interest to note the close agreement between the two models when the values of the parameters are chosen to represent the same helmet liner material (Maxwell: $E = 20.7\ MN/m^2$, $\eta = 2.07 \times 10^5$ poise; Kelvin: $E = 2.07\ MN/m^2$, $\eta = 1.38 \times 10^5$ poise; see appendix).

The effect of impact by a smaller object is shown in table 5. In this table, the velocities are adjusted so that the impact energy in each case is the same (in figure 16, the velocities were constant), and the same range of Maxwell helmet parameters is considered. From the resilient headform results, it is seen that the response of the "head" is very sensitive to these impact conditions. If the HIC is used as a performance measure, the impact of the head with a large object is much more severe than if the head is struck by a small object even though the energies are the same. Again, many examples of reversals are evident if the maximum acceleration is used as a performance measure.

To further illustrate the results of figure 15, the responses for a range of helmet parameters is shown for two cases of constant energy impacts in table 6. For both cases a rigid headform/rigid surface system is used. Large differences in the HIC value are seen for the two cases, and again reversals in the order of helmet performance are indicated when the maximum acceleration is used as a performance measure.

V. Discussion

As the preceeding results are derived from a simple one-dimensional model, they must be regarded with caution until confirmed by suitable experiments. For now the results should be regarded as reasonable indications of expected trends when test method parameters are changed.

The real life impact situation should be the starting point for test method development. At the very least the relative velocity at which the head meets an impacting object and the properties of that object must be described for one or more modes of impact which pertain to a given activity. A test method is defined by a choice of headform, drop height, and impact surface (input parameters) and a measure of performance (output parameter). There are several approaches:

1. Choose input parameters to simulate the real life impact situation. The tolerable value of the output parameter is then based upon the present state of the art on head injury. The model shows that deviations from reality in the headform and impact surface are capable of producing reversals (the improper ranking of helmets).

2. Compensate for changes in one input parameter by intentionally changing another. For example, the model suggests that the resiliency of the headform can be taken into account by an increased resiliency in the impact surface. In such cases, the tolerable value of the output parameter would remain fixed at the appropriate biomechanical value.

3. Compensate for changes in an input parameter by adjusting the output parameter. For example, if a metal headform is used, the critical value of the performance measure could be relaxed.

The results of the model suggest likely trends when moving from the real life situation to the test method situation.

The model also suggests that the ranking may be affected by the parameter chosen to measure helmet performance. The present state of the art regarding injury mechanisms strongly suggests that the likelihood of injury from linear accelerations correlates better with the duration dependent measures like HIC and SI than with the maximum acceleration. The results of the parameter studies suggests that the use of these former parameters is also less likely to introduce errors in the ranking of helmets that arise from changed test conditions.

With regard to evaluating present test methods the model can be used to compare present rejection criteria to the biomechanical parameters. As an example, consider the rejection criteria of the FMVSS 218 motorcycle standard (table 2). From table 3, the "helmets" that would fail for the conditions of the test method (rigid headform/rigid impact surface), are characterized by the parameters: $E = 27.6\ MN/m^2$; $\eta = 3.45 \times 10^5$ poise; $E = 20.7\ MN/m^2$, $\eta = 3.45 \times 10^5$ poise and $E = 13.8\ MN/m^2$, $\eta = 3.45 \times 10^5$ poise. If the results for the resilient headform/rigid impact surface are taken as representative of the real life situation, it is seen that these same three "helmets" are the only ones for which the SI exceeds 1500 (the suggested critical value for distributed impacts). Therefore, the practice of rejecting helmets when the acceleration exceeds a given value for a given time duration may be an acceptable alternative to computing the integral measures, provided that the criteria can be related to the biomechanical measures in the real life situation.

It remains to discuss some of the limitations of this work. First, the model was only concerned with the linear acceleration response. If the headform were allowed to rotate in some realistic manner, the likelihood of a changed helmet ranking is a distinct possibility and will be the subject of a future paper. Secondly, all of the conditions considered here were based on an impact mode in which the head is relatively free to rebound after the impact. Therefore, the results would not be applicable for a blow to the top of the head (though present test methods do not make any modifications for this condition). Finally another limitation is that the helmet model has no provision for bottoming (the increased rigidity of the helmet liner material with increasing compression) which is sometimes observed. It is likely that the effect of bottoming would further contribute to the difficulties in properly ranking helmets.

While this report may raise more questions than it answers, it is hoped that a quantitative framework is provided which can be used as a guide in evaluating how well a test method identifies the best helmets for a given activity.

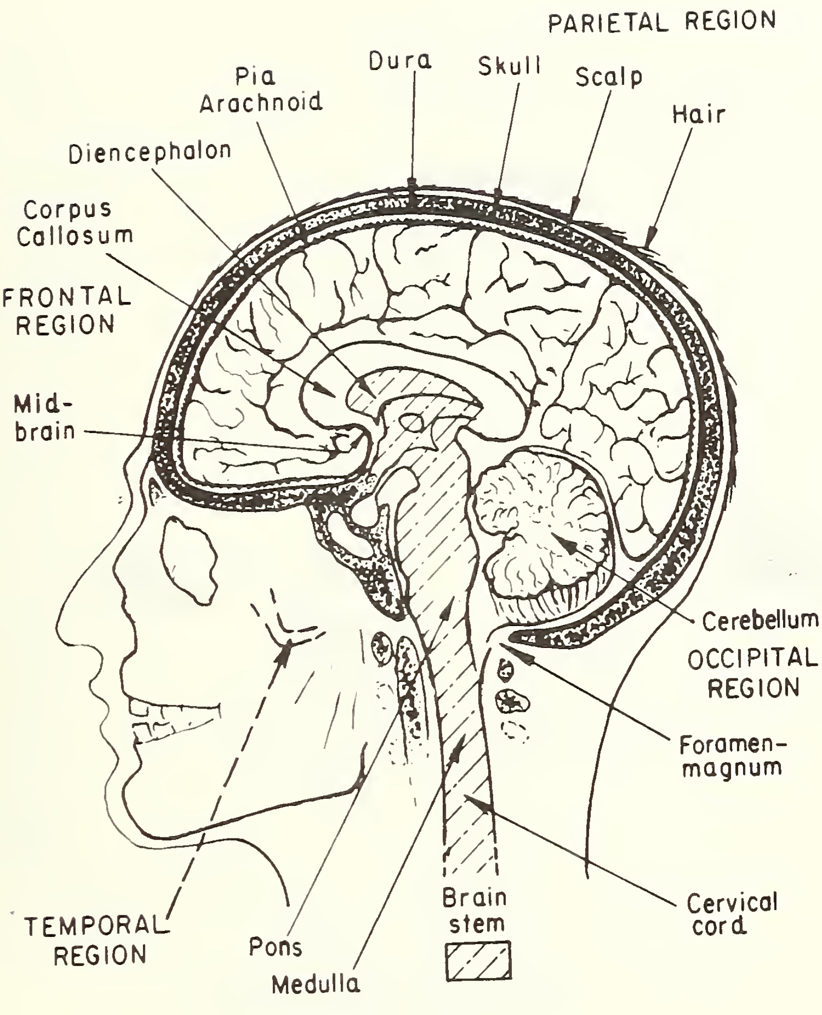


Figure 1. Schematic drawing of skull and contents (from Goldsmith 2/)

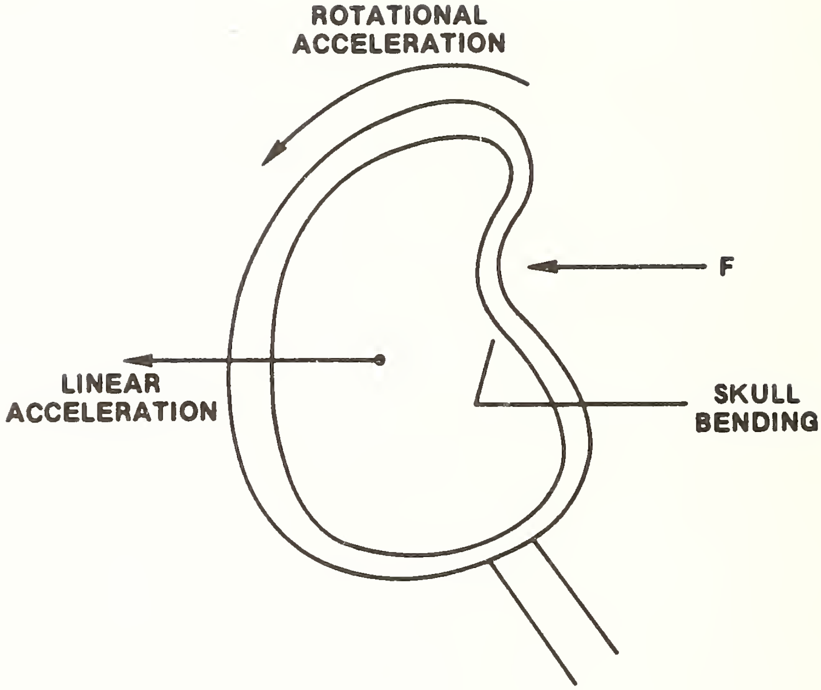


Figure 2. Gross motions of skull following impact.

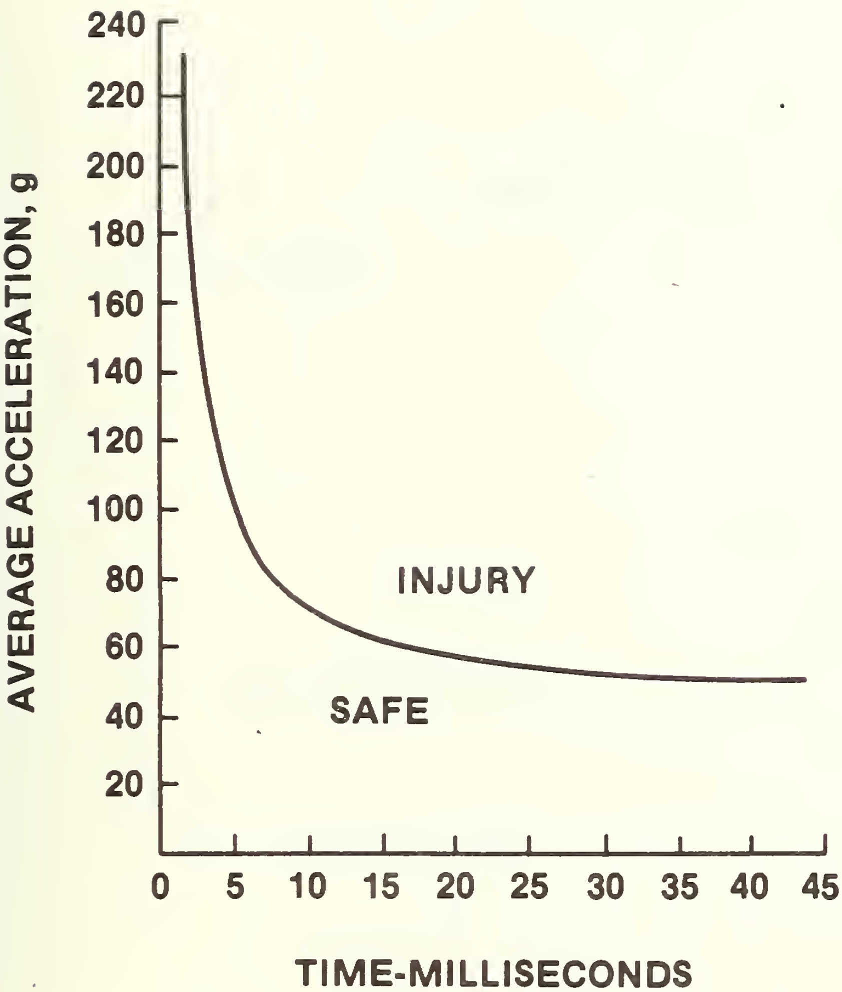


Figure 3. Wayne State Concussion Tolerance Curve.

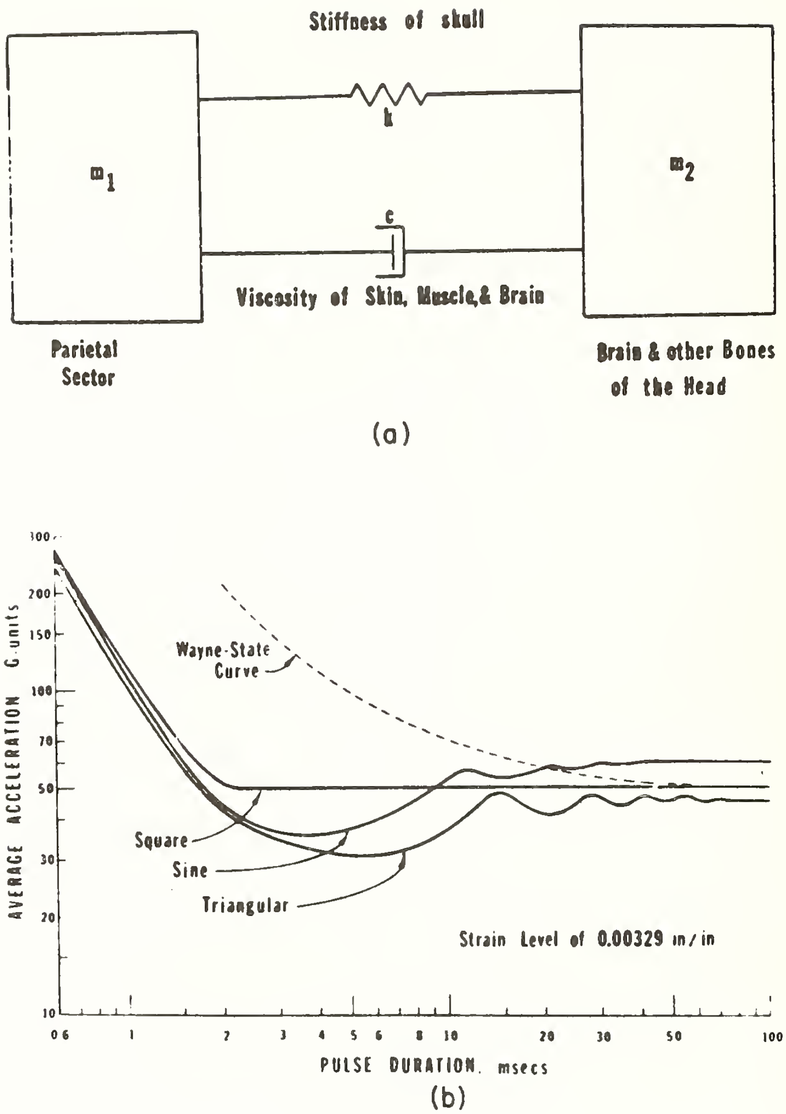


Figure 4. Head model (a) and tolerance curves (b) for maximum strain criterion (from Stalnaker, et. al. 26/).

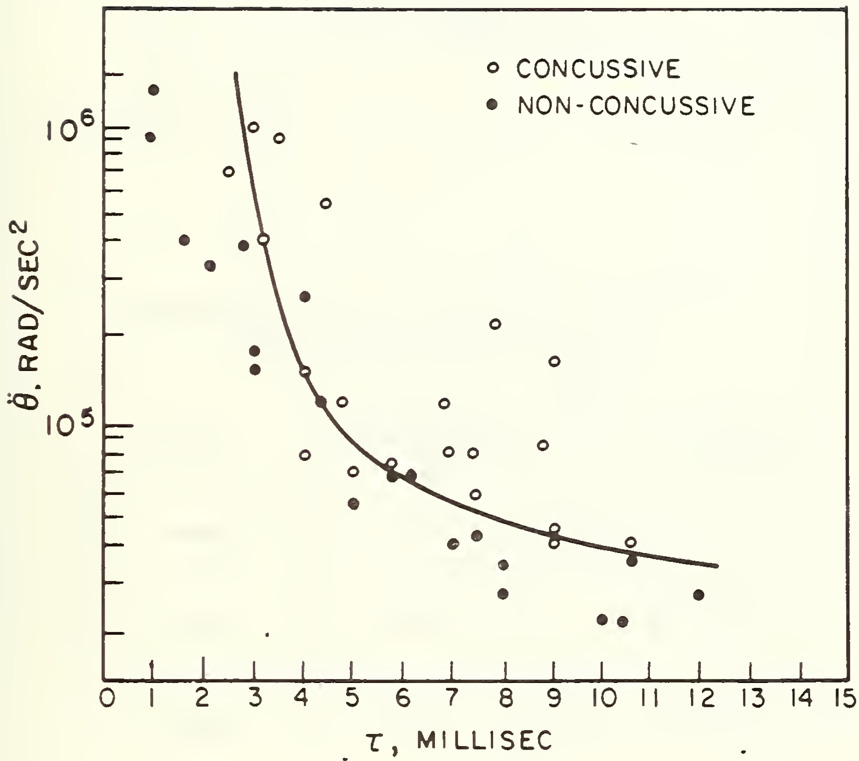
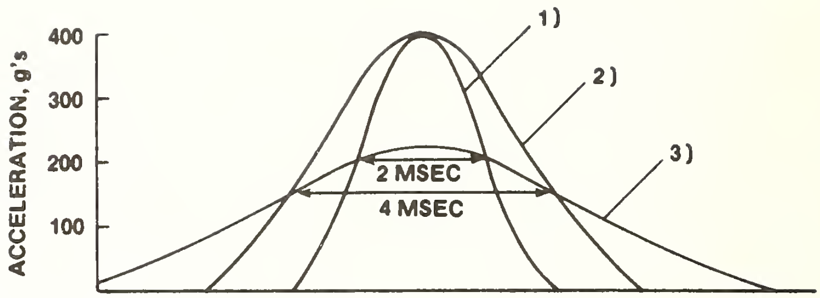


Figure 5. Tolerance curve for rotational accelerations in monkeys (from Ommaya, et. al. 30/).



SI VALUES FOR BARELY SAFE ACCELERATION RESPONSES (AS DETERMINED BY FMVSS 218, 74/)

REJECTION CRITERIA \ CURVE SHAPE	HAVERSINE (SHOWN ABOVE)	TRIANGLE
1) a = 400 g and a>200g for t = 2 msec	4350	3660
2) a = 400g and a>150g for t = 4msec	7380	5870
3) a>200g for t=2 msec and a>150g for t = 4 msec	2700	2800

Figure 6. Acceptable acceleration responses (as prescribed by FMVSS 218 Standard 74/), and corresponding SI values.

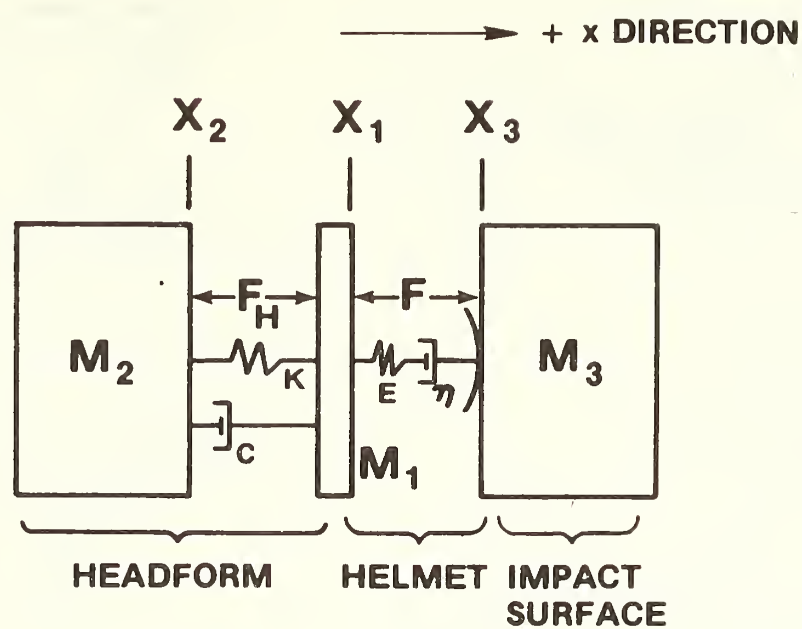


Figure 7. Lumped-parameter mathematical model used to study effect of test method parameters.

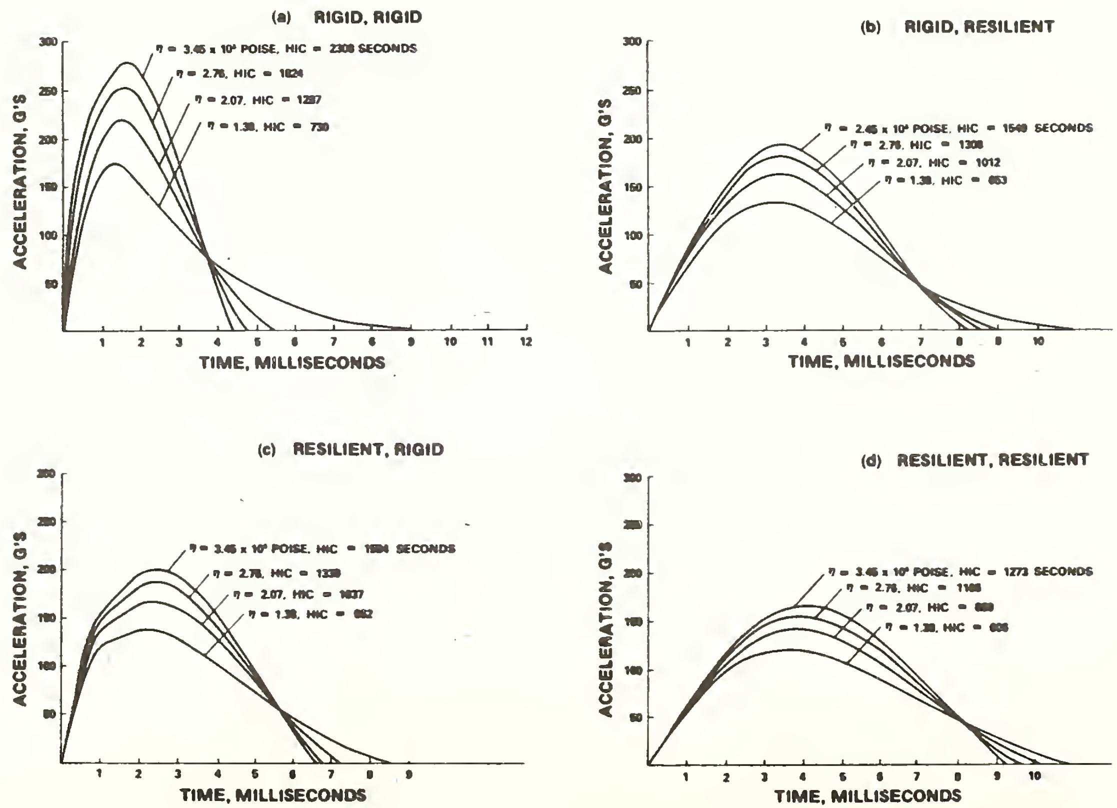


Figure 8. Effect of helmet dashpot parameter, η: a) rigid headform/rigid impact surface, b) rigid headform/resilient impact surface, c) resilient headform/rigid impact surface, d) resilient headform/resilient impact surface.

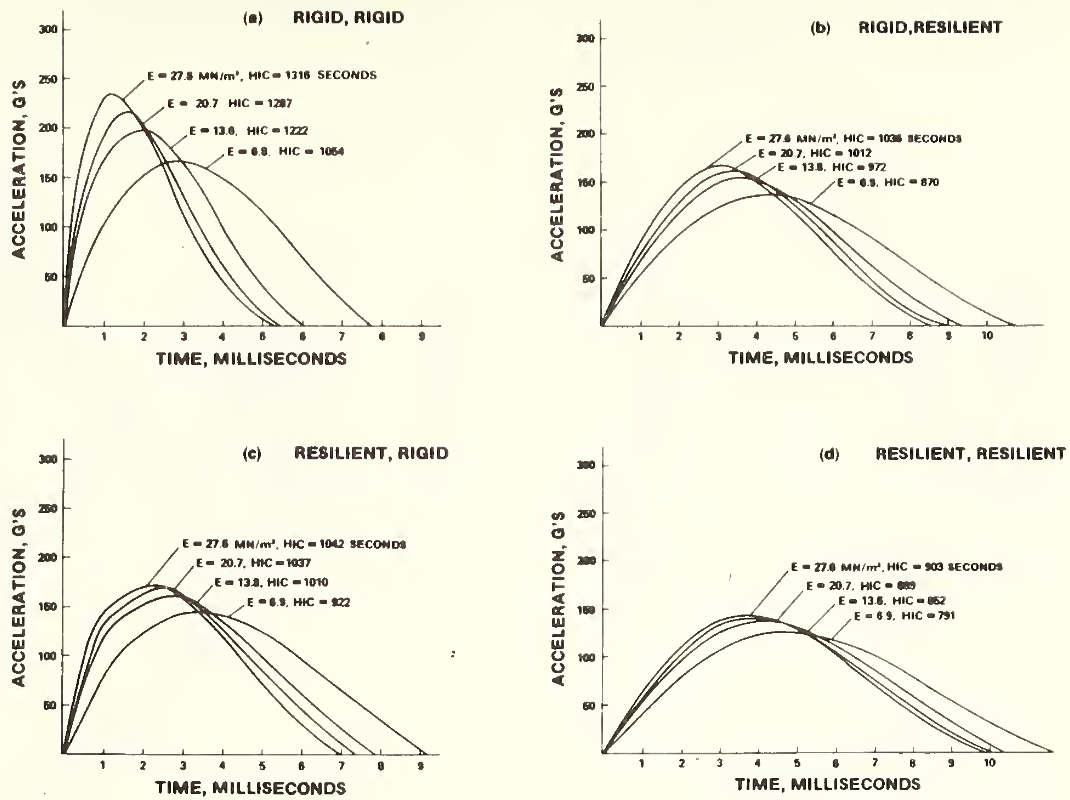


Figure 9. Effect of helmet spring parameter, E: a) rigid headform/rigid impact surface, b) rigid headform/resilient impact surface, c) resilient headform/rigid impact surface, d) resilient headform/resilient impact surface.

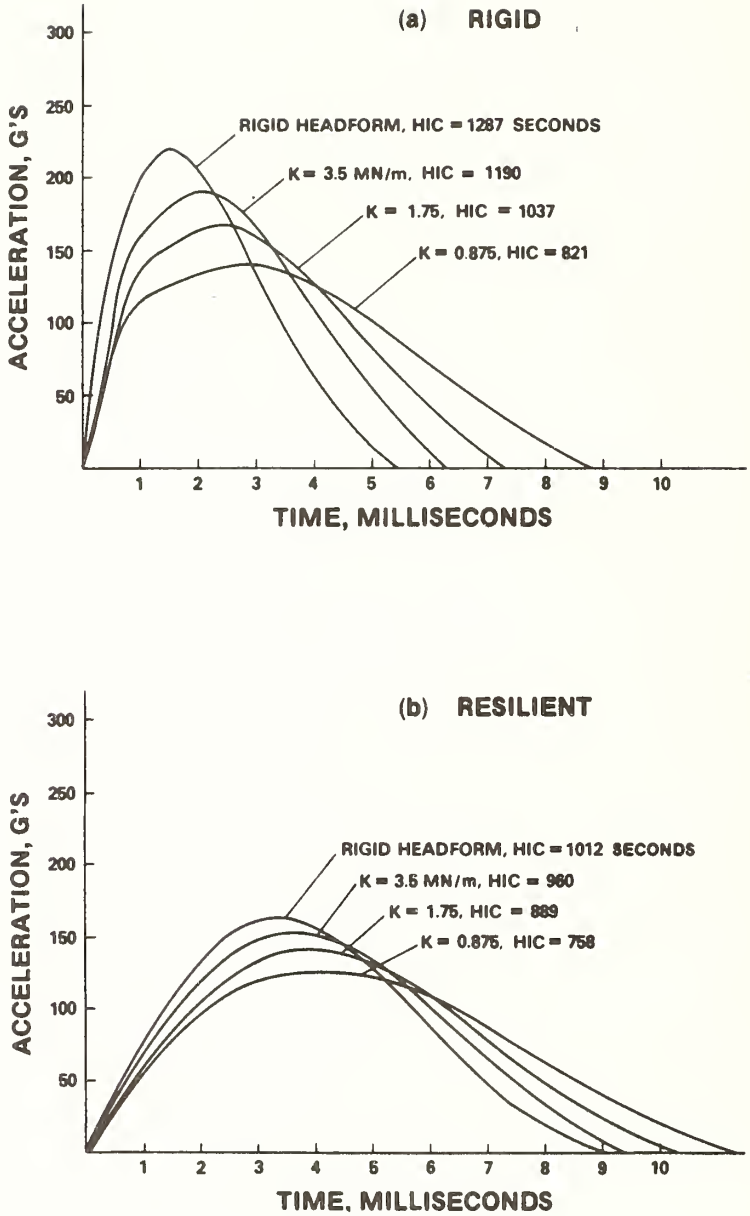


Figure 10. Effect of resilient headform spring parameter, K:
a) rigid impact surface,
b) resilient impact surface.

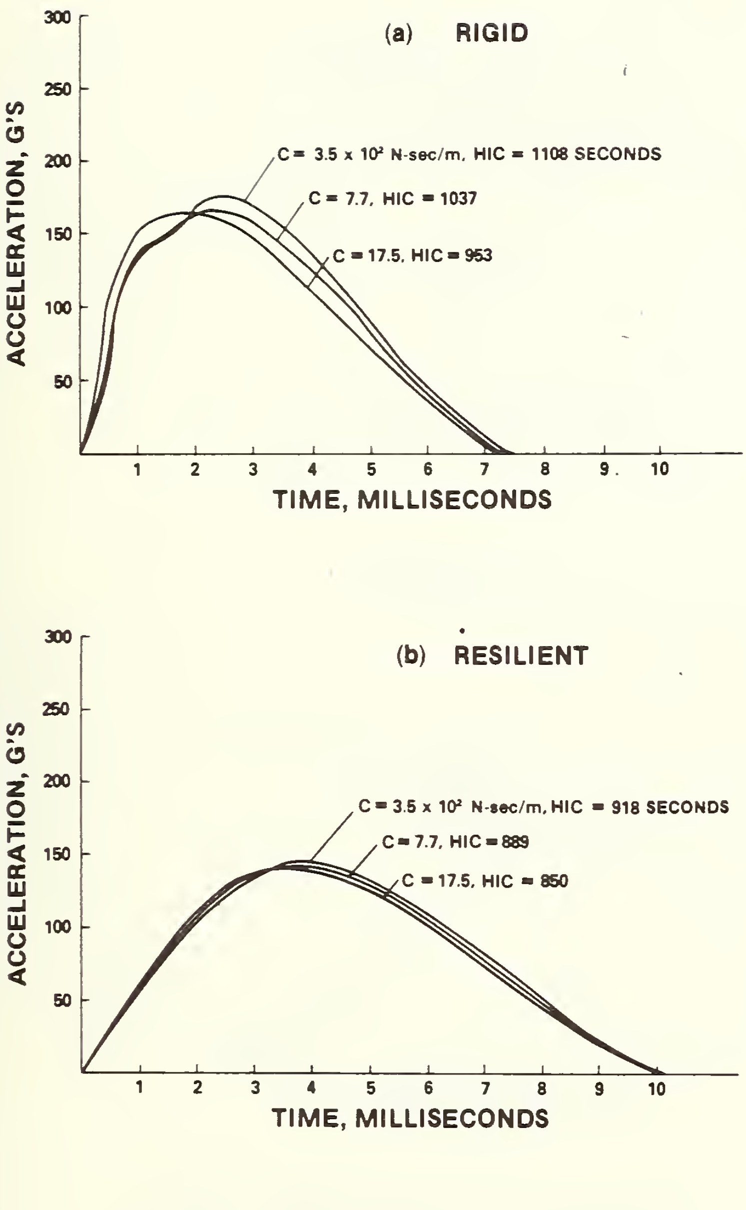


Figure 11. Effect of resilient headform dashpot parameter, C:
a) rigid impact surface,
b) resilient impact surface.

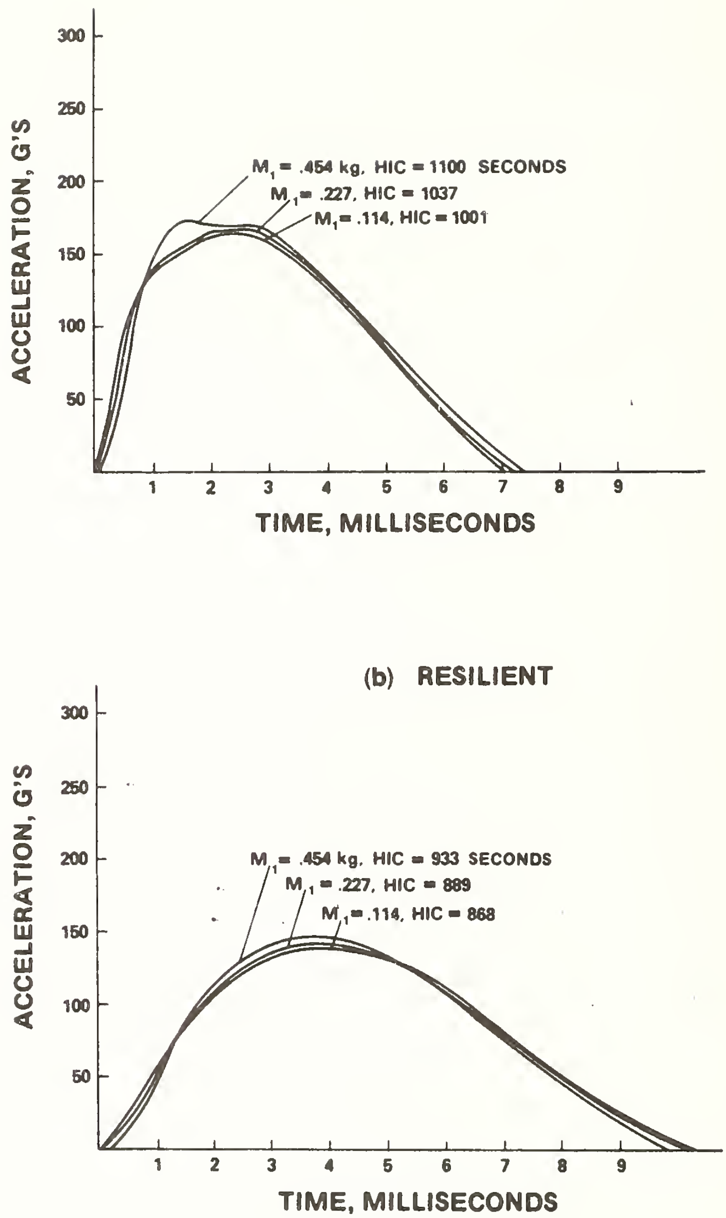
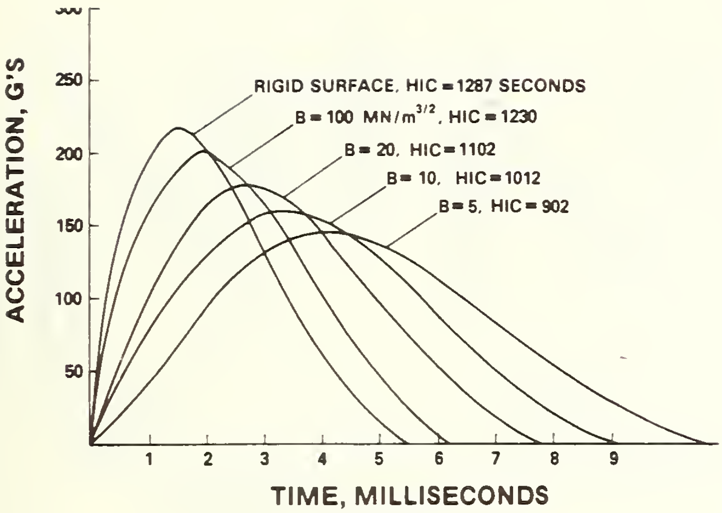


Figure 12. Effect of resilient headform mass parameters, M_1, M_2 (where $M_1 + M_2 = 4.54$ kg): a) rigid impact surface, b) resilient impact surface.



(b) RESILIENT

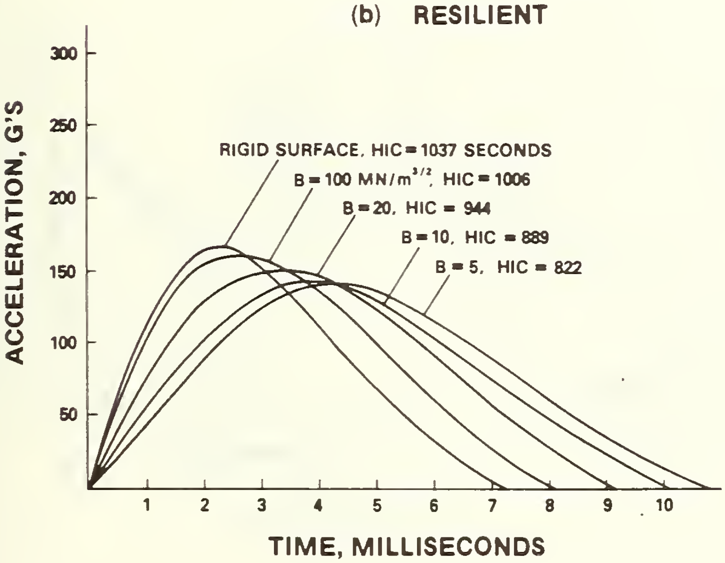


Figure 13. Effect of resilient impact surface parameter, B: a) rigid headform, b) resilient headform.

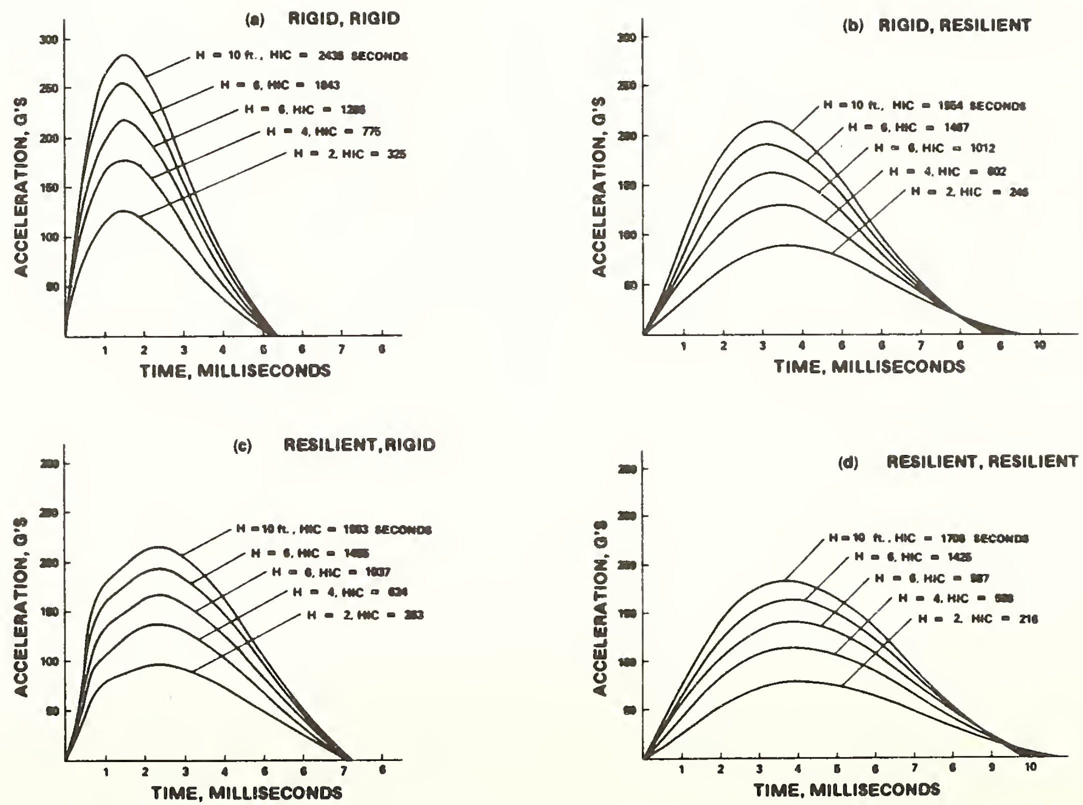
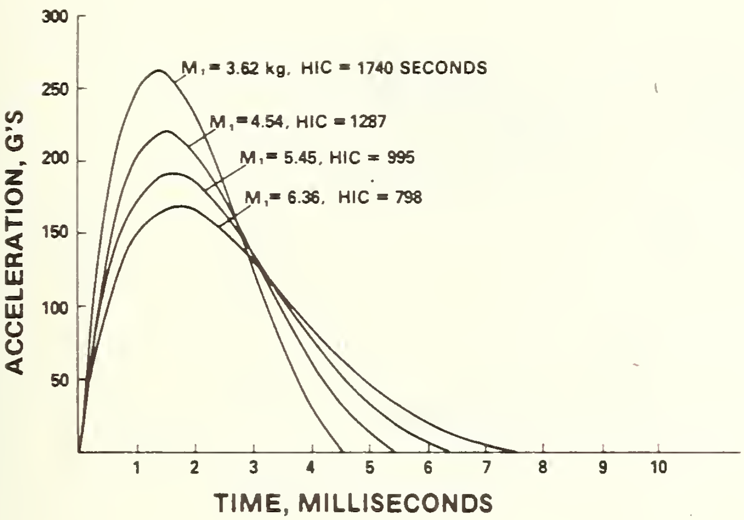


Figure 14. Effect of drop height, h: a) rigid headform/rigid impact surface, b) rigid headform/resilient impact surface, c) resilient headform/rigid impact surface, d) resilient headform/resilient impact surface.



(b) RESILIENT

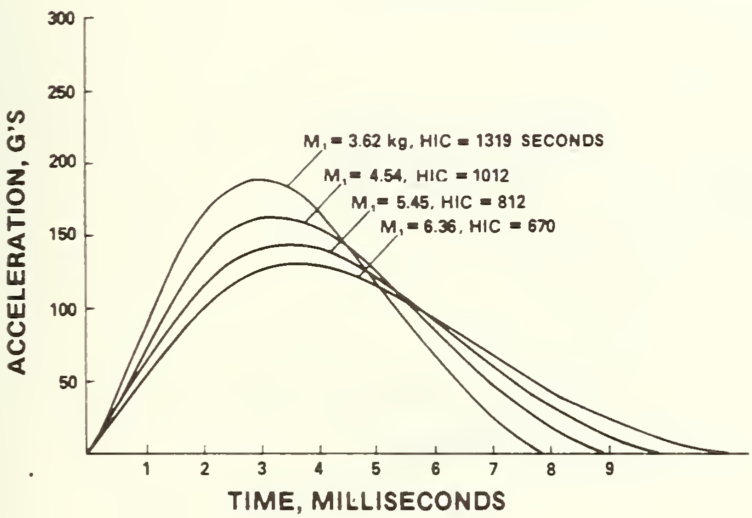
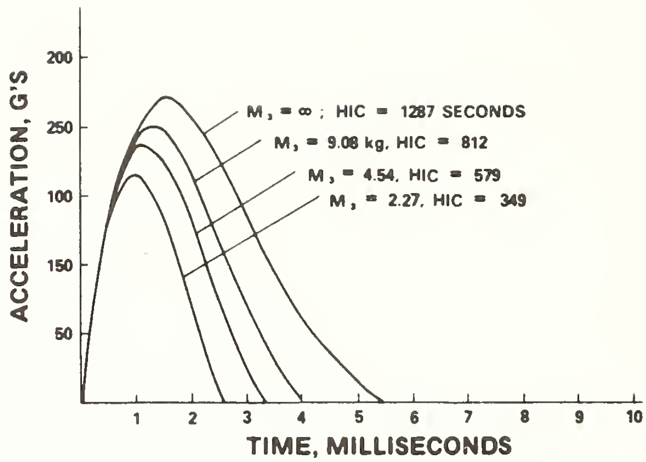


Figure 15. Effect of mass of rigid headform, M: a) rigid impact surface, b) resilient impact surface.



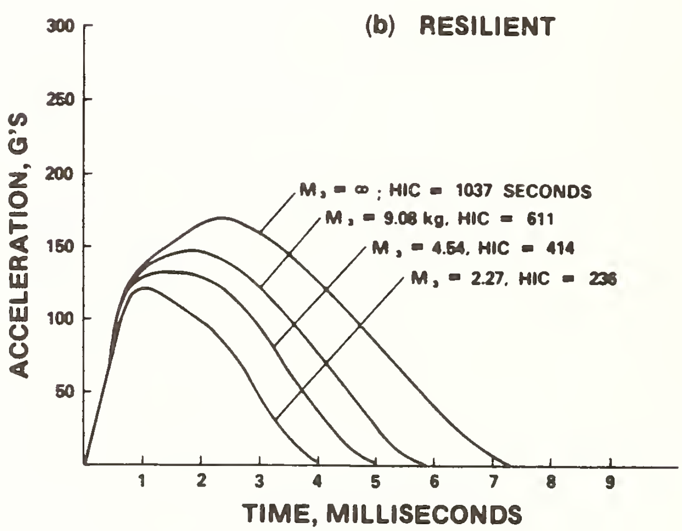


Figure 16. Effect of mass of a rigid missile, M_3, striking the helmeted headform: a) rigid headform, b) resilient headform.

Table 1. Some Features of Selected Test Methods for Protective Headgear

Activity	Test Method	Headform	Impact Surface	Energy (joules)	Rejection Criteria
Motorcycle	Snell 73/	Metal	Steel Flat Hemi	140 110	$a > 300$
	FMVSS 218 74/	Metal	Steel Flat Hemi	90 68	$a > 400$ g $a > 200$ g, $t > 2$ msec $a > 150$ g, $t > 4$ msec
	ANSI Z90.1 75/	Metal	Steel Flat Hemi	90 68	$a > 400$ g $a > 200$ g, $t > 3$ msec $a > 150$ g, $t > 6$ msec
Football	JF 73 76/	Metal	Steel Flat Hemi and MEP	90 68	$a > 400$ g $a > 200$ g, $t > 3$ msec $a > 150$ g, $t > 6$ msec
	ASTM 77/	Metal	Modular Elastomer Programmer (MEP) (60 - 70 durometer)	Arbitrary	None
	NOCSAE 72/,78/	Humanoid	MEP (30 durometer)	75	$SI > 1500$ sec
Occupational	LEAA 79/	Metal	Steel Hemi	108	$a > 400$ g $a > 200$ g, $t > 3$ msec $a > 150$ g, $t > 5$ msec
	NFPCA 80/	Metal	Steel Flat	90 top, back 35 sides, front	$a > 400$ g $a > 200$ g, $t > 3$ msec $a > 150$ g, $t > 5$ msec

Table 2. Values of Parameters for Head Model

	Group	M_1 (kg)	M_2 (kg)	C (N-sec/m)	K (MN/m)
Human Ref. <u>26</u>/	1) Side impact	.178	4.00	4.2×10^2	4.55
	2) Front impact	.267	4.45	3.5×10^2	8.75
Model	3)	.223	4.22	7.7×10^2	1.75

Table 3. Comparison of performance measures for a range of Maxwell model helmet parameters. E and η . Entries are maximum acceleration (a_{max}), severity index (SI), and head injury criterion (HIC).

Helmet parameters	Headform		Rigid				Resilient	
	Impact Surface		Rigid	Resilient			Rigid	Resilient
				$B = 20\ MN/m^{3/2}$	10	5		B = 10
$E = 6.9\ MN/m^2$	$\eta = 1.38$ ($X10^5$ poise)	a_{max} =	138 g	124	118	110	123	109
		SI =	756 s	698	668	625	691	624
		HIC =	657 s	625	597	559	626	562
	2.07		165	147	138	127	144	126
			1170	1034	969	883	1014	878
			1054	930	870	792	922	791
	2.76		183	161	151	138	158	137
			1505	1293	1197	1072	1259	1062
			1360	1165	1076	963	1145	958
	3.45		196	171	160	145	167	144
			1773	1498	1371	1214	1448	1199
			1604	1347	1234	1061	1318	1082
$E = 13.8\ MN/m^2$	1.38		160	139	130	118	134	118
			819	753	718	667	730	662
			722	668	638	594	657	595
	2.07		200	168	155	139	161	138
			1363	1174	1086	973	1110	957
			1222	1051	972	871	1010	852
	2.76		228	188	172	153	179	151
			1858	1523	1381	1208	1417	1180
			1672	1365	1238	1082	1292	1064
	3.45		249	202	184	163	191	159
			2285	1807	1615	1389	1662	1349
			2062	1624	1449	1245	1517	1218
$E = 20.7\ MN/m^2$	1.38		173	145	135	122	139	121
			840	774	737	683	739	676
			730	684	653	607	662	606
	2.07		221	179	163	145	168	142
			1446	1234	1135	1010	1139	987
			1287	1102	1012	902	1037	889
	2.76		256	202	182	160	188	156
			2035	1629	1463	1265	1468	1226
			1824	1458	1308	1132	1339	1106
	3.45		282	218	196	170	201	166
			2569	1959	1728	1464	1736	1421
			2308	1757	1549	1311	1584	1273
$E = 27.6\ MN/m^2$	1.38		181	149	138	124	141	123
			851	785	746	691	742	683
			727	691	661	613	667	612
	2.07		235	185	168	148	172	145
			1489	1267	1162	1030	1150	1003
			1318	1128	1036	919	1042	903
	2.76		275	210	188	164	192	159
			2140	1690	1509	1296	1490	1251
			1908	1512	1349	1159	1360	1128
	3.45		306	228	202	175	207	170
			2752	2049	1793	1507	1769	1443
			2471	1835	1605	1348	1619	1302

Table 4. Comparison of performance measures for a range of Kelvin model helmet parameters E and η. Entries are maximum acceleration (a_{max}), severity index (SI), and head injury criterion (HIC)

HELMET PARAMETERS	Headform	Rigid		Resilient	
	Impact Surface	Resilient $B=10\ MN/m^{3/2}$	Resilient $B=5$	Rigid	Resilient $B=10$
$E = 1.38\ MN/m^2$	$\eta = 0.69$ ($\times 10^5$ poise)	a_{max} = 111 g	106	111	105
		SI = 686 s	676	676	674
		HIC = 610 s	603	605	607
	1.04	137	125	136	123
		830	799	802	789
		726	705	706	706
	1.38	158	141	157	138
		1024	955	972	934
		900	845	-	837
$E = 2.07\ MN/m^2$	0.69	122	117	123	116
		886	865	872	860
		799	780	792	782
	1.04	145	132	143	130
		989	948	956	935
		874	844	852	841
	1.38	163	147	162	143
		1154	1075	1095	1050
		1018	954	-	942
$E = 2.76\ MN/m^2$	0.69	134	128	135	127
		1079	1041	1063	1030
		978	940	974	938
	1.04	151	139	150	137
		1147	1091	1108	1073
		1022	976	996	969
	1.38	169	152	167	147
		1285	1192	1219	1162
		1136	1062	1092	1046
$E = 3.45\ MN/m^2$	0.69	146	138	147	136
		1263	1202	1245	1185
		1146	1086	1142	1046
	1.04	158	146	158	143
		1300	1226	1255	1202
		1165	1100	1135	1089
	1.38	174	157	172	152
		1413	1305	1341	1268
		1257	1165	1205	1145

Table 5. Effect of mass of impacting object for a range of Maxwell model helmet parameters E and η. Entries are maximum acceleration (a_{max}), severity index (SI), and head injury criterion (HIC). The case for $M_3 = \infty$ is the same as for headform striking massive surface.

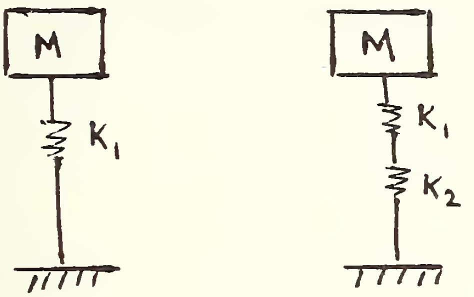
HELMET PARAMETERS	Headform		Rigid			Resilient		
	Mass, M_3, kg		∞	4.45	2.22	∞	4.45	2.22
	Impact Velocity, M/S		6.0	6.0	8.5	6.0	6.0	8.5
$E = 6.9\ MN/m^2$	$\eta = 1.38$ ($X10^5$ poise)	a_{max} =	138 g	114	143	123	100	125
		SI =	756 s	328	464	691	290	405
		HIC =	657 s	295	419	626	263	368
	2.07		165	132	162	144	113	139
			1170	469	633	1014	393	526
			1054	423	571	922	358	478
	2.76		183	143	173	158	122	147
			1505	571	750	1259	464	606
			1360	516	678	1145	423	552
	3.45		196	151	181	167	127	153
			1773	648	835	1448	516	662
			1604	586	755	1318	471	605
$E = 13.8\ MN/m^2$	1.38		160	138	175	134	110	143
			819	378	557	730	315	446
			722	337	500	657	285	400
	2.07		200	165	205	161	127	161
			1363	585	821	1110	441	599
			1222	527	739	1010	402	544
	2.76		228	183	225	179	139	172
			1858	753	1023	1417	534	705
			1672	679	924	1292	487	641
	3.45		249	196	238	191	145	179
			2285	887	1178	1662	603	781
			2062	800	1065	1517	551	712
$E = 20.7\ MN/m^2$	1.38		173	151	194	139	114	151
			840	399	600	739	321	551
			730	350	535	662	291	410
	2.07		221	186	234	168	132	171
			1446	645	928	1139	456	620
			1287	579	829	1037	414	562
	2.76		256	209	259	188	144	183
			2035	852	1193	1468	556	735
			1824	772	1078	1339	506	668
	3.45		282	226	277	201	152	191
			2569	1037	1406	1736	632	819
			2308	935	1271	1584	578	746
$E = 27.6\ MN/m^2$	1.38		181	160	208	141	115	154
			851	409	624	742	323	459
			727	361	553	667	289	412
	2.07		235	200	254	172	134	176
			1489	682	997	1150	461	626
			1318	603	895	1042	420	562
	2.76		275	228	285	192	147	189
			2140	929	1313	1490	565	745
			1908	836	1180	1360	517	670
	3.45		306	248	307	207	156	198
			2752	1143	1575	1769	543	832
			2471	1031	1415	1619	591	751

Table 6. Effect of rigid headform mass for constant energy impacts for a range of Maxwell model helmet parameters, E and η. Entries are maximum acceleration (a_{max}), severity index (SI), and head injury criterion (HIC).

		Headform Mass (kg)	5	6
		Drop Height (m)	1.83	1.53
HELMET PARAMETERS	$E = 6.9\ MN/m^2$	$\eta = 1.38$ ($\times 10^5$ poise)	a_{max} = 128 g SI = 665 s HIC = 594 s	102 413 367
		2.07	175 1040 936	124 659 592
		2.76	172 1349 1218	139 867 782
		3.45	185 1598 1444	150 1039 939
	$E = 13.8\ MN/m^2$	1.38	148 715 629	117 438 383
		2.07	187 1200 1074	148 747 666
		2.76	213 1649 1483	171 1041 935
		3.45	233 2041 1840	188 1303 1174
	$E = 20.7\ MN/m^2$	1.38	159 732 633	125 447 383
		2.07	205 1268 1126	162 781 692
		2.76	238 1796 1606	190 1122 1001
		3.45	264 2281 2052	211 1442 1296
	$E = 27.6\ MN/m^2$	1.38	167 740 630	130 451 379
		2.07	218 1301 1149	171 799 701
		2.76	256 1881 1674	203 1167 1036
		3.45	285 2434 2186	228 1526 1369

Appendix 1

Consider the very simplest resilient "head" model with one mass and one spring. The effect of a "helmet" is modelled by adding a second spring in series.



$$K^* = \frac{K_1 K_2}{K_1 + K_2} < K_1$$

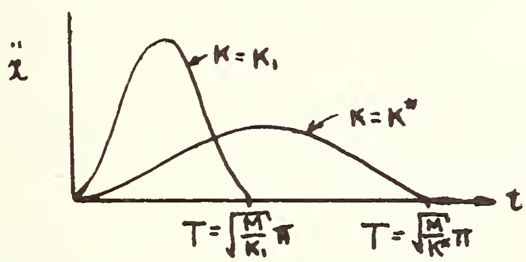
The governing equation and initial conditions for this system are:

$$M\ddot{X} = -KX, \quad X(0) = 0, \quad \dot{X}(0) = V$$

The acceleration as a function of time is given by

$$\ddot{X} = -V\sqrt{\frac{K}{M}} \sin\left(\sqrt{\frac{K}{M}}\, t\right)$$

A sketch of the acceleration response is shown below

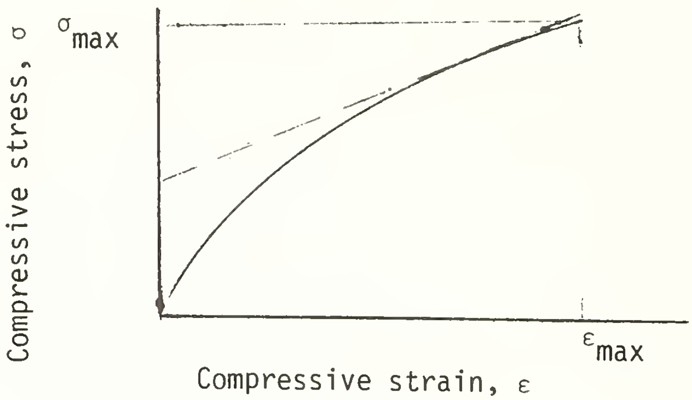


where T is the time at which the acceleration first vanishes. A severity index can be computed for this portion of the acceleration curve:

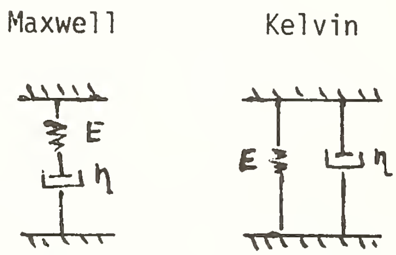
$$SI = \int_0^T \left(\frac{\ddot{x}}{g}\right)^{2.5} dt = \left(\frac{K}{M}\right)^{.75} \left(\frac{V}{g}\right)^{2.5} \int_0^{\pi} (\sin u)^{2.5}\, du$$

Therefore it is seen that the severity index increases with K. It is also of note that both the maximum acceleration and the severity index decrease with increasing mass of the "head."

In order to evaluate the proposed models for helmet liner materials, and to determine a realistic range of parameter values, the experimental results of Melvin and Roberts 87/ were utilized. In this study, cylindrical specimens of representative helmet liner materials were loaded at constant velocities and the compressive stress was monitored. The cross-sectional dimensions of the specimens remained constant up to maximum strain. A typical stress-strain curve is shown below (as the solid line in the sketch):



The purpose of this appendix is to fit two simple viscoelastic models, the Maxwell element and the Kelvin element, to this data and determine the cooresponding values of the parameters of the models.



The relationships between stress and strain for these models are 91/:

$$\text{Maxwell:}\quad \dot{\sigma} + \frac{E}{\eta}\sigma = E\dot{\varepsilon}$$

$$\text{Kelvin:}\quad \sigma = E\varepsilon + \eta\dot{\varepsilon}$$

For loading at constant strain rate, $\varepsilon = K$, the Maxwell equation is easily solved, and the relationships between stress and strain become:

$$\text{Maxwell:}\quad \sigma = \eta K \left[1 - \exp\left(-\frac{E}{\eta K}\varepsilon\right)\right]$$

$$\text{Kelvin:}\quad \sigma = E\varepsilon + \eta K$$

As Melvin and Roberts 87/ noted, their experiments were conducted at constant velocity rather than constant strain rate, a difference which is unimportant for small strains. As our objective is to determine a reasonable range of parameter values rather than an accurate curve fitting, we use the constant strain rate assumption up to maximum strain as it simplifies the analysis.

It can be seen from the above equations that the Maxwell model is much more capable of matching the stress-strain curves. The Kelvin model is characterized by an instantaneous jump in the stress and by the linear stress-strain behavior. As shown by the dashed line in the above sketch, the Kelvin model is only approximately appropriate if the early non-linear portion of the curve is restricted to sufficiently small strains.

The following quantities were calculated and reported in the Melvin and Roberts study 87/: 1) the stress, σ_{max}, at maximum strain, ε_{max}, and 2) the specific energy absorbed up to maximum strain, defined as

$$Q = \frac{\int_0^{x_{max}} F dx}{\text{Initial Volume}}$$

where F and x are force and displacement. If A and d are the initial cross sectional area and length of the specimen, Q can also be expressed as:

$$Q = \int_0^{\varepsilon_{max}} \sigma d\varepsilon$$

For the two viscoelastic models, the expression for σ_{max} and Q are

$$\text{Maxwell:} \begin{cases} \sigma_{max} = \eta K \left[1 - \exp(-\frac{E}{\eta K}\varepsilon_{max})\right] \\ Q = \eta K \left\{ \varepsilon_{max} - \frac{\eta K}{E}\left[1 - \exp(-\frac{E}{\eta K}\varepsilon_{max})\right]\right\} \end{cases}$$

$$\text{Kelvin:} \begin{cases} \sigma_{max} = E\varepsilon_{max} + \eta K \\ Q = E \frac{\varepsilon_{max}}{2} + \eta K \varepsilon_{max} \end{cases}$$

Then if σ_{max}, Q, K and ε_{max} are known for each model, there are two equations for the unknown values of the material parameters, E and η.

The data which corresponds to the greatest speed in the Melvin and Roberts study, 217 in/sec, is used, as this is nearest to the range of impact velocities considered in this report. The calculated values for polystyrene, a typical liner material for motorcycle helmets, is shown below.

				Maxwell		Kelvin	
Specific gravity	ε_{max}	σ_{max} MN/m^2	Q MN/m^2	E MN/m^2	η poise X 10^{-5}	E MN/m^2	η poise X 10^{-5}
0.0175	.46	.34	.15	4.4	.45	.28	.21
0.0540	.45	1.21	.50	11.0	1.10	.97	.69

As both E and η, for both models, appear to vary linearly with density*, the values for more typical densities (specific gravities of about 0.1 for motorcycle helmets) can be extrapolated:

Maxwell: $E = 20.7\ MN/m^2$, $\eta = 2.07 \times 10^5$ poise

Kelvin: $E = 2.07\ MN/m^2$, $\eta = 1.38 \times 10^5$ poise

The range of values to be considered in this report are:

Maxwell: $E \sim 5 - 30\ MN/m^2$
$\eta \sim 1 - 4 \times 10^5$ poise

Kelvin: $E \sim 1 - 4\ MN/m^2$
$\eta \sim 0.5 - 1.5 \times 10^5$ poise

To derive equations (3a) and (3b) of the text, the strain ε is related to the displacements X_1 and X_3 (see figure 8) by

$$\varepsilon = \frac{d - (X_3 - X_1)}{d}$$

and the force is given by

$$\sigma = \frac{F}{A}$$

assuming that the load is distributed over a constant area**.

*Such linear relationships are not unexpected for polystyrene foams if the volume fraction of polystyrene is small 91/ as it is for the densities considered above.

**In a continuum model, the contact area is a function of the force. In this one dimensional model the area is assumed constant and equal to the somewhat arbitrary value of .016 m^2, chosen to give realistic headform responses.

References

1. Schneider, L.C., Head and Neck Injuries in Football, William and Wilkins, Baltimore (1973)

2. Goldsmith, W., Biomechanics of Head Injury, in Biomechanics, its Foundations and Objectives, Y.C. Fung, N. Perrone, M. Anliker, editors, Prentice Hall, Ind. (1972)

3. Mahajan, B.M. Standards for Athletic Helmets - State of the Art and Recommendations, National Bureau of Standards Report: NBSIR 73-276 (1974)

4. Hodgson, V.R. and Thomas, L.M., Head Injury Tolerance, in Aircraft Crashworthiness, K. Saczalski, ed. (1975)

5. Ommaya, A.K., Head Injuries in Adults, Current Therapy, H.F. Conn (Ed.), W.B. Saunders, Philadelphia (1972)

6. Thomas, L.M., Mechanisms of Head Injury, in Impact Injury and Crash Protection, E.S. Gurdjian, editor, Charles C. Thomas, publisher, Springfield, Illinois (1970)

7. Caveness, W.F. and Walker, A.E. (Eds.): Head Injury Glossary (Appendix), in Head Injury Conference Proceedings, Philadelphia, Lippincott, pp. 571-576 (1966)

8. Ommaya, A.K. and Gennarelli, T.A., Cerebral Concussion and Traumatic Unconsciousness: Correlation of Experimental and Clinical Observations on Blunt Head Injuries, Brain, 97, pp. 633-654 (1974)

9. Walker, A.E., Mechanisms of Cerebral Trauma and the Impairment of Consciousness, in Neurological Surgery, Vol. II, J. Youmans (Ed.), Sanders, Philadelphia, 936-949 (1973)

10. Symonds, C.P. Concussion and its Sequlae, Lancet 1, 1-5 (1962)

11. Gurdjian, E.S., Recent Advances in the Study of the Mechanism of Impact Injury of the Head - A Summary, Clinical Neurosurgery, Vol. 19 pp. 1-42 (1972)

12. Hodgson, V.R., Physical Factors Related to Experimental Concussion, in Impact Injury and Crash Protection, E.S. Gurdjian, ed. Charles C. Thomas, publisher, Springfield, Illinois (1970)

13. Kenner, V.H. and Goldsmith, W., Impact on a Simple Physical Model of the Head, J. Biomechanics, 6, pp. 1-11 (1973)

14. Gross, A.G., A New Theory on the Dynamics of Brain Concussion and Brain Injury, J. Neurosurgery, 15 (5): 548-561, (1950)

15. Shugar, T.A., Simulating and Modeling the Human Head's Response to Impact, in Aircraft Crashworthiness, K. Saczalski, ed., selected proceedings of symposium at U. Conn. (October 1975)

16. Ommaya, A.K. and Hirsh, A.E., Protection of the Brain from Injury during Impact: Experimental Studies in the Biomechanics of Head Injury, AGARD Conference on Linear Acceleration, NATO (1971)

17. Shatsky, S.A., Alter, W.A., Evans, D.E., Armbrustmacher, V., and Clark, G., Traumatic Distortions of the Primate Head and Chest: Correlation of Biomechanical Radiological and Pathological Data, Proceedings of the 18th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 351-381 (1974)

18. Shatsky, S.A., Evans, D.E., Miller, F.E., and Martins, A.N., High Speed Angiography of Experimental Head Injury, Armed Forces Radiobiology Research Institute, Scientific Report, SR75-6 (March 1975)

19. Gurdjian, E.S., Lissner, H.R., and Patrick, L.M., "Protection of the Head and Neck in Sports," JAMA, Vol. 182, pp. 509-512 (November 3, 1962)

20. Haut, R.C., Gadd, C.W., and Madiera, R.G., Non-linear Viscoelastic Model for Head Impact Injury Hazard, Proceedings of the 16th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 149-159 (1972)

21. Newman, J.A., On the Use of the Head Injury Criterion (HIC) in Protective Headgear Evaluation, Proceedings of the 19th Stapp Car Crash Conference, Society of Automotive Engineers, New York (1975)

22. Fan, W.R.S., "Internal Head Injury Assessment," Proceedings of 15th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 645-665 (1971)

23. Slattenschek, A., and Tauffkirchen, W., "Critical 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, Society of Automotive Engineers, New York, Paper 700426, p. 30 (1970)

24. Gadd, C.W., "Report to SAE Performance Criteria Subcommittee, "Vehicle Research Department, CM Research Laboratories, GM Technical Center, Warren, Mich. 48090 (March 8, 1972)

25. Versace, J., "A Review of the Severity Index," Proceedings of 15th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 771-796 (1971)

26. Stalnaker, R.L., McElhaney, J.H., and Roberts, V.L., MSC Tolerance Curve for Human Head Impacts, ASME paper 71-WA/BHF - 10 (1971)

27. McElhaney, J.H., Stalnaker, R.L. and Roberts, V.L., Biomechanical Aspects of Head Injury, in Human Impact Response, Proc. of G.M. Symposium, Plenum Publishing Corp. (1972)

28. Stalnaker, R.L., Roberts, V.L. and McElhaney, J.H., "Side Impact Tolerance to Blunt Trauma," Proceedings of 17th Stapp Car Crash Conference, Society of Autmotive Engineers, New York, pp. 377-408 (1973)

29. Ommaya, I.K., and Hirsh, A.E., Tolerances for Cerebral Concussion from Head Impact and Whiplash in Primates, J. Biomechanics, 4, pp. 13-31 (1971)

30. Ommaya, A.K., Yarnell, P., Hirsch, A.E., and Harris, E.H., "Scaling of Experimental Data on Cerebral Concussion in Sub-Human Primates to Concussion Threshold for Man," Proceedings of 11th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 47-52 (1967)

31. Gennarelli, T.A., Ommaya, A.K., and Thibault, L.E., "Comparison of Translational and Rotational Head Motions in Experimental Cerebral Concussions," Proceedings of 15th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 797-803 (1971)

32. Gennarelli, T.A., Thibault, L.E., and Ommaya, A.K., Pathophysiologic Responses to Rotational and Translational Accelerations of the Head, Proceedings of 16th Stapp Car Crash Conference, Society of Automotive Engineers, New York (1972)

33. Gennarelli, T.A., and Thibault, L.E., Functional Response of the Central Nervous System to Controlled Inertial Loading, Proceedings of the 27 Annual Conf. on Eng. in Med. and Bio., p. 175 (1974)

34. Thibault, L.E., Gennarelli, T.A., Tipton, H.W., and Carepenter, D.O., The Physiological Response of Isolated Nerve Tissue to Dynamic Mechanical Loads, Proceedings of the 27th Annual Conf. on Eng. in Med. and Bio., p. 176 (1974)

35. Holbourne, A.H.S., "Mechanics of Head Injury," Lancet, Vol. 245, pp. 438-441 (1943)

36. Hubbard, P.H., Flexure of Layered Cranial Bone, J. Biomechanics 4, 251-263 (1971)

37. Gurdjian, E.S., Hodgson, V.R., and Thomas, L.M., Studies on Mechanical Impedance of the Human Skull: Preliminary Report, J. Biomechanics, 3, 239-248 (1970)

38. L. Shuck and S. Advani, "Rheological Response of Human Brain Tissue in Shear," J. Basic Engineering, Transactions ASME (December 1972)

39. Galford, J.E. and McElhaney, J.H., "A Viscoelastic Study of Scalp, Brain and Dura," J. Biomechanics, 3, 211-222 (1970)

40. Truesdell, C., The Elements of Continum Mechanics, Springer-Verlag, New York (1966)

41. Guttinger, W., Der Stosseffekt aut eine Flussigkeits-kugel als Grandlage einer physikalischen Theorie der Entstehung von Gehirnverletzungen, Z. Naturf, A.5, 622-628 (1950)

42. A. Engin, "The Axisymmetric Dynamic Response of a Fluid-Filled Spherical Shell to a Local Radial Pulse-A Model for Head Injury," J. Biomechanics, Vol. 2, pp. 324-341 (1969)

43. Benedict, J.W., Harris, E.H. and Rosenberg, D.U., An Analytical Investigation of the Cavitation Hypothesis of Brain Damage, J. Basic Engng. 597, 603 (1970)

44. Akkas, N., Dynamic Analysis of a Fluid-Filled Spherical Sandwich Shell - A Model of the Human Head, J. Biomechanics, 8, pp. 275-284 (1975)

45. Merchant, H.C. and Crispino, A.J., A Dynamic Analysis of an Elastic Model of the Human Head, J. Biomechanics 7, 295-301 (1974)

46. Chan, H.S. and Liu, Y.K., The Asymmetric Response of a Fluid-Filled Spherical Shell A Mathematical Simulation of a Glancing Blow to the Head, J. Biomechanics 7, 43-59 (1974)

47. Landkof, B. and Goldsmith, W., and Sackman, J.L., Impact on a Head-Neck Structure, J. Biomechanics, 9, pp. 141-151 (1976)

48. Hickling, R. and Wenner, M.L., Mathematical Model of a Head Subjected to an Axisymmetric Impact, J. Biomechanics, 6, pp. 115-132 (1973)

49. Advani, S.H. and Owings, R.P., Evaluation of Head Injury Criteria, Automotive Engineering Conf., P-11, Society of Automotive Engineers, New York (1974)

50. Advani, S.H. and Owings, R.P., Structural Modeling of the Human Head, Journal of the Engineering Mechanics Division, Proceedings of the American Society of Civil Engineers, Vol. 101, No. EM3 June (1975)

51. Chan, H.S., Mathematical Model for Closed Head Impact, Proceedings of the 18th Stapp Car Crash Conference, Society of Autmotive Engineers, New York, pp. 557-578 (1974)

52. Lee, Y.C. and Advani, S.H., Transient Response of a Solid Viscoelastic Sphere to Torsional Loading A Head Injury Model, J. Math. Biosci., 6, 473-486 (1970)

53. Firoozbakhsh, K.K. and DeSilva, A Model of Brain Shear Under Impulsive Torsional Loads, J. Biomechanics, 8, 65-73 (1975)

54. Bycroft, G.N., Mathematical Model of a Head Subjected to an Angular Acceleration, J. Biomechanics 6, 487-495 (1973)

55. Liu, Y.K., Chandran, K.B. and von Rosenberg, D.V., Angular Acceleration of Viscoelastic (Kelvin) Material in a Rigid Spherical Shell - a Rotational Head Injury Model, J. Biomechanics, 8, pp. 285-292 (1975)

56. Ljung, C., A Model for Brain Deformation due to Rotation of the Skull, J. Biomechanics, 8, pp. 263-274 (1975)

57. Löwenhielm, P., Mathematical Simulation of Gliding Contusions, J. Biomechanics, 8, pp. 351-356 (1975)

58. Gurdjian, E.S., Prevention and Mitigation of Injuries, in Clinical Neurosurgery, Vol. 19, Congress of Neurological Surgeons pp. 43-57 (1972)

59. Gurdjian, E.S., Hodgson, V.R., Hardy, W.G., Patrick, L.M. and Lissner, H.R., Evaluation of the Protective Characteristics of Helmets in Sports, J. of Trauma, 4, pp. 309-324 (1964)

60. Rayne, J.M. and Maslen, K.R., Factors in the Design of Protective Helmets, Aerospace Medicine, pp. 631-637 (June 1969)

61. Patrick, L.M., "Head Impact Protection," Head Injury Conference Proceedings, Lippincott, Philadelphia, pp. 41-48 (1966)

62. Liu, Y.K. and Chandrau, K.B., Package Cushioning for the Human Head, ASME paper No. 74-WA/Bio. -10 (1974)

63. Liu, Y.K., Nikravesh, P.E., and Beck, C.H., Optimal Protection in Direct Head Impact, Proceedings of the 28th Annual Conf. on Engineering in Medicine and Biology (1975)

64. McElhaney, J.H., Head Protection, Design and Testing, Proceedings of the 28th Annual Conference on Engineering in Medicine and Biology, p. 168 (1975)

65. McElhaney, J.H., Roberts, V.L. and Stalnaker, R.L., A Brain Injury Model for Crash Helmet Design, Proc. of the 18th IES Meeting (1972)

66. V.R., Hodgson, M.W. Mason, and L.M. Thomas, "Head Model for Impact," Paper 720969, Proceedings of 16th Stapp Car Crash Conference, P-45, New York, Society of Automotive Engineers, Inc. (1972)

67. McElhaney, J.H., Mate, P.I., and Roberts, V.L., A New Crash Test Device - "Repeatable Pete," Society of Automotive Engineers Paper #739037 (1973)

68. Hubbard, R.P. and McLeod, D.G., Definition and Development of a Crash Test Dummy Head, Proceedings of 18th Stapp Car Crash Conference, Society of Automotive Engineers, New York, pp. 599-628 (1974)

69. Stalnaker, R.L., McElhaney, J.H., and Roberts, V.L., "Mechanical Impedance Response of Prototype and Production Anthropomorphic Dummy Heads," Paper No. 73-WA/Bio-13 presented at ASME Winter Annual Meeting, Detroit (1973)

70. Master, B.G., and Saczalski, K.J., Anthropomorphic Headform Development and Simplified Technique for Impact Evaluation of Protective Headgear, Paper No. 72-WA/BHF-7 presented at ASME Winter Annual Meeting, New York (1972)

71. Andrews, L.M., Football Helmet Tests: Round Robin Using NOCSAE Recommended Test Method, National Bureau of Standards Letter Report to the Consumer Product Safety Commission (March 3, 1976)

72. Hodgson, V.R., National Operating Committee on Standards for Athletic Equipment Football Helmet Certification Program, Medicine and Science in Sports, 7 (3) pp. 225-232 (1975)

73. Standard for Protective Headgear, Snell Memorial Foundation, G.G. Snively (Dir.), 761 Laurel Dr., Sacramento, CA (1975)

74. Motorcycle Helmets, Federal Motor Vehicle Safety Standard, FMVSS No. 218, National Highway Traffic Safety Administration, Federal Register 38 F.R. 22390 (Aug. 20, 1973)

75. Specifications for Protective Headgear for Vehicular Users, American National Standard ANSI Z90.1-1971, American National Standards Institute, Inc., New York (1971); and Supplement ANSI Z90.1a (1973)

76. James, A.F. and Fenner, H.A., Recommended Test Method and Performance Standard for Protective Headgear for Football Players, JF73 Standard (1973)

77. Standard Method of Test for Shock Attenuation Characteristics of Protective Headgear for Football, ASTM F429-75, American Society for Testing and Materials, Philadelphia (1975)

78. Standard Method of Impact Test and Performance Requirements for Football Helmets, Prepared by the National Operating Committee on Standards for Athletic Equipment, Inc. (Sept. 1973)

79. NILECJ Standard for Riot Helmets, U.S. Department of Justice, Law Enforcement Assistance Administration, National Institute for Law Enforcement and Criminal Justice (October 1974)

80. Calvano, N.J., Performance Criteria for Structural Firefighters' Helmets (preliminary draft), National Bureau of Standards, prepared for National Fire Prevention and Control Administration (July 1976)

81. Safety Requirements for Industrial Head Protection, ANSI Z89.1-1969, American National Standards Institute, Inc., New York (1969)

82. Andrews, L.M., Football Helmet Tests: Comparison of Data Using Both NOCSAE and ASTM Recommended Test Methods, National Bureau of Standards letter report to the Consumer Product Safety Commission (March 3, 1976)

83. Stalnaker, R.L., Fogle, J.L., and McElhaney, J.H., Driving Point Impedance Characteristics of the Head, J. Biomechanics, 4, 127-139 (1971)

84. Goldsmith, W. Impact, Edward Arnold, London (1960)

85. Timoshenko, S. and Goodier, J.N., Theory of Elasticity (2nd Edition) McGraw Hill, New York (1951)

86. Agranoff, J. (ed.), Modern Plastics Encyclopedia, Vol. 52, No. 10a, McGraw Hill, New York (1975)

87. Melvin, J.W. and Roberts, V.L., Compression of Cellular Plastics at High Strain Rates, J. Cellular Plastics, Vol. 7, No. 2 (1971)

88. Henderson, G., Correlation Anomalies Between Helmet Drop-Test Systems, paper presented at Technical Meeting, Safety Helmet Council of America, Los Angeles, California (November 1, 1974)

89. Gill, S., A Process for the Step-by-Step Integration of Differential Equations in an Automatic Digital Computing Machine, Proc. Cambridge Phil. Soc. 47, p. 96 (1951)

90. Nilson, E.N., Cubic Splines on Uniform Meshes, Comm. A.C.M., 13, (4) (1970)

91. Nielsen, L.E., Mechanical Properties of Polymers, Rheinhold, New York (1962)

Acknowledgements

The author wishes to thank Nicholas J. Calvano for his interest and advice and Louise Shannon for her patient typing of the manuscript.

NBS-114A (REV. 7-73)

U.S. DEPT. OF COMM. BIBLIOGRAPHIC DATA SHEET	1. PUBLICATION OR REPORT NO. NBSIR76-1107	2. Gov't Accession No.	3. Recipient's Accession No.

4. TITLE AND SUBTITLE Considerations in Developing Test Methods For Protective Headgear	5. Publication Date August 1976 6. Performing Organization Code 446.03
7. AUTHOR(S) Robert E. Berger	8. Performing Organ. Report No.
9. PERFORMING ORGANIZATION NAME AND ADDRESS NATIONAL BUREAU OF STANDARDS DEPARTMENT OF COMMERCE WASHINGTON, D.C. 20234	10. Project/Task/Work Unit No. 4460143 11. Contract/Grant No.
12. Sponsoring Organization Name and Complete Address *(Street, City, State, ZIP)* National Bureau of Standards Washington, D.C. 20234	13. Type of Report & Period Covered Interim 14. Sponsoring Agency Code

15. SUPPLEMENTARY NOTES

16. ABSTRACT *(A 200-word or less factual summary of most significant information. If document includes a significant bibliography or literature survey, mention it here.)*

The purpose of this report is to outline the relationship between test methods for protective headgear and injury prevention. The state-of-the-art concerning the mechanisms of head injury and the relationship of these injuries to the modes of energy input are briefly reviewed. The general guidelines in the design of protective headgear are discussed, and the difficulties with present test methods for helmets are summarized.

In order to provide a quantitative framework, a simple model which incorporates many features of present test methods is defined and executed. The model predicts the effect of changes in test method parameters (headform, impact surface, drop height) for a range of helmet parameters. Among the indications suggested by the model are: 1) the occurrence of reversals in the order of helmet performance when the maximum acceleration is used as a performance measure, 2) the absence of reversals when biomechanical measures are used, and 3) the indicated suitability of compensating for the resiliency of the headform by the resiliency of the impact surface.

17. KEY WORDS *(six to twelve entries; alphabetical order; capitalize only the first letter of the first key word unless a proper name; separated by semicolons)*
Protective headgear; head injury; helmets; test methods; math model; safety

18. AVAILABILITY	19. SECURITY CLASS (THIS REPORT)	21. NO. OF PAGES
XX Unlimited ☐ For Official Distribution. Do Not Release to NTIS ☐ Order From Sup. of Doc., U.S. Government Printing Office Washington, D.C. 20402, SD Cat. No. C13 X Order From National Technical Information Service (NTIS) Springfield, Virginia 22151	UNCLASSIFIED 20. SECURITY CLASS (THIS PAGE) UNCLASSIFIED	53 22. Price $4.50

USCOMM-DC 29042-P74