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Use of a Weighted-Impulse Criterion for Estimating Injury Hazard

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Abstract

This paper describes the usage of an exponential weighting factor for appraising deceleration or force impulses registered on dummies or impacting hammers in safety testing. The proposed impulse-integration procedure, it is shown, takes into account in a more rational way, and in better conformity with published injury tolerance data, the relative importance of time and intensity of the pulse than do the "peak g" or impulse-area criteria. Use of the new Severity Index for assessment of head impact pulses is illustrated. It is shown to be of special value in comparing the relative severity of pulses which differ markedly in shape (because of structural differences in the component being struck), and it is pointed out that without a weighting factor of this nature, laboratory impact tests can yield incorrect ranking of the relative safety merit of alternative designs. Automated methods for quick calculation of the Severity Index are possible.

IN RECENT YEARS there has been an increasing need for more versatile measures or indices with which to judge the degree of injury hazard likely to be associated with impulses applied in the laboratory testing of automotive interior structures and components, or with which to draw comparisons between alternate designs proposed for reducing injury. In view of the wide range of locations and angles of impact which can be experienced by the head or other parts of the body in accidents, and recognizing the diverse mechanisms by which injury can occur within the body, it would be unrealistic to assume that we will ever have a single and rigorously quantitative rating system for the hazard inherent in a given pulse applied to a given part of the human body. Simple measures or at least "yardsticks" are nevertheless very much needed and have therefore come into use. It is believed that these can be further improved to yield better approximation of injury hazard than has been obtained in the past, and at the same time yield more repeatable and comparable results. The object of this paper is to describe one approach toward this end.

Various terms from the field of mechanics have come into use over the years to characterize the intensity of a blow or typify the manner in which the impulse must be altered to reduce its injury potential. Of these, the term "energy absorption" has been one of the most popular. This concept has, however, been difficult to apply in a specific way because it means different things to different people, and there is not generally any simple relation between energy involved and injury hazard. Variables usually present, such as the crush characteristics of the striking and struck objects, prevent good correlation of either the kinetic energy of the striking object, or the energy absorbed by the struck object, with injury hazard.

Selection of Transducer

To arrive at a logical index of injury hazard represented by laboratory or field test results, one must first face the question of what type of transducer measurement to make. Under impact, the body may be exposed to acceleration, force, pressure, stress, or strain. Depending upon the nature of the problem and type of injury, it may be most practicable to select one or another of these parameters for measurement. Here the best choice for the original tolerance measurement on the biological material is, of course, to select a transducer whose output is believed to be closely related to the mechanism of injury, and locate it as close as possible to the actual injury site. It is then preferable to use the same type of measurement in the impact testing of the design under study.

In head injury, the problem to which special attention will be given in this paper, it has been impractical except in very limited instances to obtain transducer readings (for example, pressure, force, or stress) which are directly associated with the injury, and, as a result, the overall head acceleration, a rather indirect measure, has come into wide use. While acceleration admittedly does not consistently represent the diversity of kinematics and injury mechanisms actually involved, it does provide probably the best currently available basis for judging head impact severity from an internal injury standpoint. The most reliable information which has been obtained is that for impact of the front of the head, and in particular of the forehead.

Interpretation of Pulse Wave Shape

Once a pulse depicting a blow has been obtained on the oscillograph, the next question arising is how its severity should be assessed from the standpoint of waveform or profile. Various investigators have emphasized differing aspects of the wave. The maximum value reached, or "peak g" if it is an acceleration pulse, is the most widely used rule-of-thumb measure of injury hazard inherent in the pulse because it is the simplest, even though it has been pointed out by various people (Ref. 1) that from the mechanics standpoint a single point on an applied pulse cannot accurately define the response of a physical structure to that pulse. Area under the g-time pulse has also been suggested as a simple way of at least recognizing that injury hazard generally increases with increasing time of exposure to a loading upon the body.

Rate-of-change of acceleration is still another aspect of pulse wave shape which has been suggested as a critical factor in injury. In discussing this subject, it is advisable to differentiate first between the input function and the response function; if the transducer monitoring input excitation must (of necessity) be placed on the body at a point apart from the actual site of injury, and there is a mass-elastic system intervening, then both rate-of-change of acceleration and impulse area can, under certain conditions, become useful as rough indices of how the dynamic characteristics of the intervening system alter the stress intensity at the injury site, and are thus useful in this sense although not as indices of overall injury to be produced by the input function. It is the belief of the author that, in such cases, the dynamic response should be treated as a prior and separate problem, just as is done in applying shock and vibration theory to determine how the dynamic response of a mechanical structure aggravates the stress at a critical point, before one applies material strength theory to estimate the damage potential of that stress. In head impact studies, the transducer cannot usually be placed at the site of injury, nor is the true mass-elastic system usually well simulated in most cases; fortunately, however, the head appears to be designed sufficiently free of resonant effects to enable useful impact evaluations to be made by observing only the input function.

This paper discusses a method of assessing the pulse waveform in its entirety, in a manner which is in contrast to methods which consider only one point or aspect of the wave. It begins with the premise that injury is some function of both intensity of the loading and its time duration. Assuming, then, that the investigator has selected and placed his transducer, whether it measures acceleration, force, or pressure, so as to be best representative of distress at the injury site, he can then integrate the pulse obtained in such a way as to take into consideration both intensity and time, employing a mathematical weighting factor which best fits the available range of biomechanics data pertinent to injury at the point in question.

Exponential Weighting of Pulse

One of the simplest weighting factors which might be selected for trial is one which weights exponentially the intensity scale. This in effect takes into account time dependency of damage as follows. First, one can visualize a hypothetical completely brittle material which fails suddenly if the loading exceeds a certain level. At the other extreme would be a completely viscous material for which percentage increments of load intensity would be just as damaging as corresponding increments of time duration of loading, with failure defined as some excessive degree of shear strain.

Examination of the biomechanics literature indicates that animal tissues fall somewhere between these two extremes in their failure properties and, furthermore, that the use of either of these extreme criteria will lead at times to false ranking of the relative injury hazard between alternative designs.

To the knowledge of the author, the first systematic study of the role of load duration in animal impact injury was that at Wayne State University (Refs. 2, 3). This showed, for cranial pressure pulses of similar shape but differing time duration, a trend as shown by the scatter band of Fig. 1. As time of exposure to pressure increased, the tolerable intensity decreased. A similar trend is exhibited by the work of other subsequent investigators who have obtained or assembled tolerance data over a range of pulse time durations; for example, Fig. 2 shows the trends as portrayed by Eiband of NASA for various impact sled tests (Ref. 4). It should be pointed out that such curves are very difficult to develop in the face of the many variables involved, and it is doubtful that enough data will ever be obtained over a range of time durations while holding to a particular wave shape (for example, square, triangular, or trapezoidal) to arrive at a precise mathematical definition of time dependency for blows to particular parts of the body. The data do exhibit the one feature in common, that of downward sloping tolerance curve over the time duration range of vehicle occupant impact. Therefore, it can be concluded that some

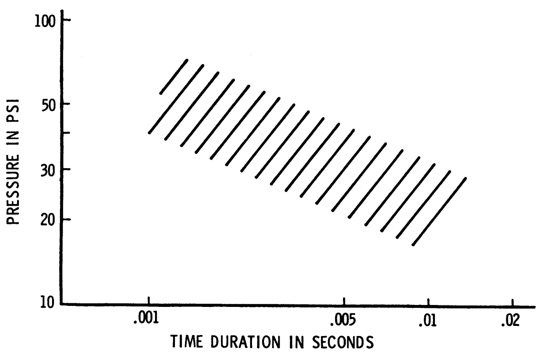

Fig. 1—Trend of intracranial pressures required for severe concussion in dogs versus time (From work of Wayne State University)

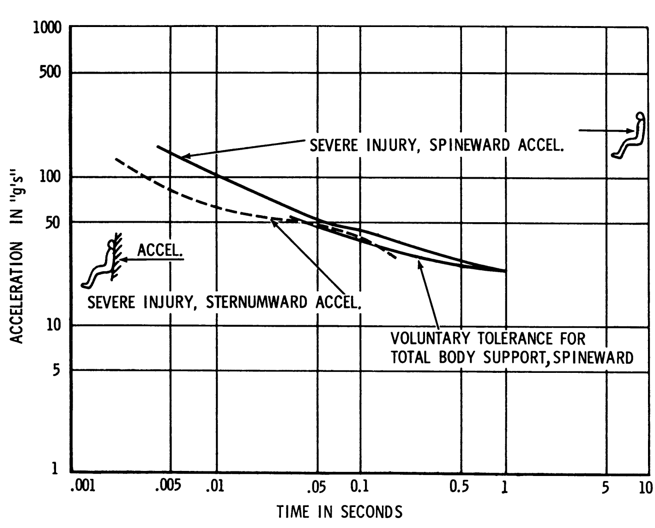

Fig. 2—Trends of tolerance to square and trapezoidal acceleration from Ref. 4

limiting g, force or pressure should not serve as the most appropriate criterion for the threshold of injury.

A second feature of the threshold curves is that, over this time duration range, their slope is considerably less than 45 deg when plotted on a log-log scale, and therefore a pulse-area criterion is also not the best approximation. It follows that some intermediate threshold curve is most appropriate. Further, in view of the scatter in the data, it is suggested that a straight line approximation on a log-log plot is sufficient at this time for head injury likely to result from front-to-rear head acceleration over a range of between approximately 1-50 msec. This brackets the pulse time duration range encountered by vehicle interior head impact.

Mathematically, the inverse of the slope of such a straight line threshold corresponds numerically with a simple exponential weighting factor, from which it follows that injury threshold can be defined as a single number,

$$I = \int a^n dt \tag{1}$$

where:

a = Acceleration, force, or pressure of response function producing threshold of injury of given degree
n = Weighting factor greater than 1
t = Time, sec

Integration of this expression yields a severity index which is applicable to a particular class of injury and whose numerical value varies depending upon whether it is developed in terms of acceleration, force, or other indication of loading intensity.

The exponential weighting may be thought of as recognizing that the lower portions of the pulse contribute very little to the injury, but that the more intensive portions contribute to a disproportionately great degree.

The number obtained can be used in two ways: either for comparing different tests for relative severity of impact, or for estimating whether an impact exceeds a safe maximum value.

Application of Severity Index

To use the Index to estimate injury hazard of a given type, two judgments must be made from the available biomechanics data as follows:

1. The appropriate weighting exponent must be selected. If one is interested in only relative ranking of designs, then this one constant is sufficient. For internal injury to the head from frontal blows, we have been using an exponent of 2.5, based primarily on the slope of the Wayne animal impact data representing dangerous concussion. The slope selected by Eiband (Ref. 4) for spineward (front to rear) acceleration of the seated human also has approximately the same value.

2. The maximum pulse intensity which can be sustained without danger to life must also be selected if absolute, rather than relative, estimates are to be made. In our work we have been using a numerical value of 1000 for the

threshold of serious internal head injury in frontal impact recorded in terms of g's. In other words, if one impacts a structure with a dummy, cadaver, or standard 15 lb headform, and finds upon integration of the g-time trace that the above Severity Index exceeds 1000, he assumes that danger to life is indicated for that particular test.

A numerical value of 1000 is in reasonable conformity with the data thus far published. For example, if square waves representing various combinations of g and time are taken from the Eiband tolerance curve (for example, 100 g for 0.010 sec), these integrate to approximately 1000. Inasmuch as reading the NASA curve in this manner does not take into consideration the additional damage from the exposure of the test subject to the onset and offset ramps of the acceleration profile, an Index of 1000 may be considered as conservative on the basis of this set of data.

This value is also in reasonable agreement with the Wayne head tolerance curve frequently cited (Ref. 5). On first inspection, this will not appear to be the case since, for example, this curve passes through a point (Fig. 3) whose coordinates are 100 g for only 0.005 sec, whereas a square wave of 100 g for 0.010 sec would calculate to an injury number of 1000. In discussion with Prof. Patrick of Wayne, however, it was verified that the g-time traces used in developing the intensity scale for the human tolerance curve were not square, and effective values of g were therefore chosen which were less than the peak values. Thus a weighting factor was employed, in effect, on the correct assumption that other investigators using the curve in the future would seldom encounter square waves, and would also need to make a judgment of effective pulse height. Integration of a number of the original Wayne oscillograms shown to the author by Prof. Patrick indicated good agreement with the Wayne data.

One of the principal advantages of the Severity Index discussed herein, over a visual weighting which otherwise must be employed, is that it eliminates differences in judgment which are bound to arise even between experienced

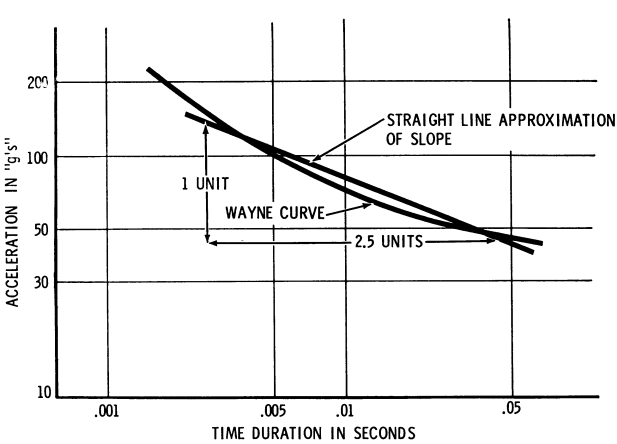

Fig. 3—Log-log plot of Wayne tolerance curve and straight line approximation of its slope over time duration range of automotive interior head impact

workers, and thus permits repeatable and comparable test results to be obtained in different laboratories and over an indefinite span of time by different personnel.

A third check against biomechanics data was more recently possible through the cooperation of Mr. Swearingen of FAA, who kindly furnished information regarding the field accident cases, some resulting in fatality, which he simulated in the laboratory. Here an injury number of 1000 fell at approximately the median point between the number of occupants surviving and number who did not survive. It is quite possible that some of the latter received additional injury over and above that from frontal head impact alone, and that, on the basis of these tests, an Index of 1000 would indicate a survival rate of well over 50%.

It should be pointed out that the value of 1000 for threshold of danger to life for internal head injury in frontal blows is not a fixed quantity; it can be adjusted upward or downward in the future as more survival studies are carried out and if the consensus of the data justifies such an adjustment.

Assessment of Differing Pulse Waveforms

Probably the greatest advantage of an integration procedure is that it can systematically handle or compare widely differing waveforms. It is known that very high "g" can be tolerated by the head for a few milliseconds, and that only a fraction of this pulse intensity can be withstood in the range of 40-50 msec. When, as is usually the case, the pulses are irregular in profile, one cannot accurately compare them visually for relative injury hazard; for example, traces a and b in Fig. 4. Another dilemma occurs if an extremely sharp "spike" is present as in trace c. If the waveform is in the range of 1-2 msec duration, it would be very questionable to select the peak of this spike as one's criterion; yet, it would not be logical to ignore the spike altogether. An integration procedure offers a repeatable and, at the same time, more rational means to handle a situation of this kind. Again, if a high frequency ringing occurs (as in trace d) which normally represents a spurious vibration

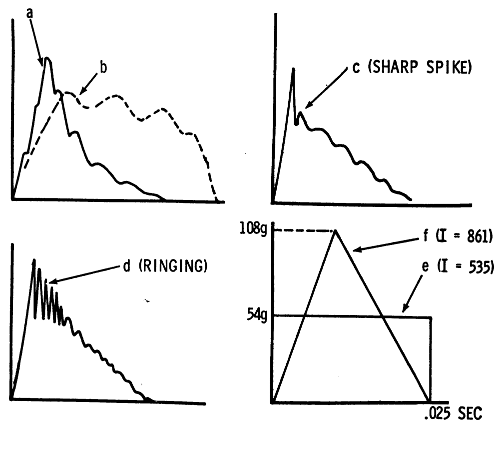

Fig. 4—Typical waveforms to be appraised

in the testing system rather than a real damaging factor, integration will yield a repeatable assessment closer to the true real damage potential represented by the pulse than if one were to select the peak values of the oscillation as the envelope of the curve.

It is of interest to compare a square versus triangular pulse according to various methods of assessment. The square pulse (trace e) of 54 g for 0.025 sec has an injury number of 535 using the 2.5 power weighting factor. Trace f is a triangular pulse of the same time duration, which under a pulse-area criterion would have the same damage potential, but under a peak criterion would be twice as damaging. Using 2.5 power integration, the triangular pulse of equal area has an injury hazard 1.61 times as great, or 861.

Computation of Severity Index

This is a relatively simple problem. The integration may be done graphically by dividing the pulse into a sufficient number of time increments to define its shape, which usually requires a number in the range of 20-30. At each increment of time, the ordinate is raised to the 2.5 power (using a graph or table if one desires) and multiplied by the time duration of the increment. The increments are then added to obtain the injury number.

Modern curve readers and machine computation equipment are advantageous for this purpose and, in the author's organization, programs have been set up for this purpose.

Pulse Durations Beyond 0.050 Sec

It is known that tolerance curves for similar pulse waveforms tend to asymptote toward a relatively fixed g level at long-time duration. It is a matter of some controversy just what value of the latter should be regarded as safe for long-time durations, and this is not presently known to any great degree of accuracy even for specific classes of injury. In view of this, and because the exponential function fits quite well over the time durations experienced in vehicle head impact, the above Severity Index is suggested for this type of impact. We have recognized, on the other hand, that at some time in the future, and for certain purposes, it might be well to have available a more complex index designed to extend into the long-time duration range and based upon a larger body of tolerance data than is now available. This would be capable of being fitted to a tolerance curve of arbitrary shape which might be constructed, as were the Eiband and Wayne curves, by noting borderline injury for a range of similar pulses of widely differing time duration.

J. P. Danforth has suggested such an index, in which the exponent "n" is not a constant but is a prescribed function of acceleration level on the experimentally developed tolerance curve; in other words, one does not need to employ a straight line approximation of the slope of the tolerance curve, but can assume, for example, that it is steeper in the realm of extremely short time durations and approaches a horizontal asymptote at long duration. A polynomial or a series of two or three simple analytical functions is then employed to express n as a function of g-level over the entire time duration of interest, and this can serve as a basis for machine computation of the Severity Index as before. It is expected that this will be the subject of a future paper.

Further Possible Applications of Severity Index Concept

Thus far, our use of this concept has been limited for the most part to the estimation of hazard of internal head injury in frontal impact. It is felt, however, that there are other classes of impact injury, some to bodily zones other than the head, for which an index that recognizes time dependency should prove beneficial, as exemplified by the following.

FACIAL INJURY—In contrast with head injury which arises essentially from a disturbance to the head as a whole, there is the important class of frontal injuries which often involves fracture of the facial bone structure or depressed fracture of the forehead. These usually result from severe loading concentrated over a small area. While trauma indicating headforms serve in a useful way as a relative measure of this hazard, an extension of the use of the Severity Index described herein shows promise as an alternative measure.

As a basis for the development of an index for facial injury, one can refer to the work of Swearingen (Ref. 6) and that of Hodgson, Lange, and Talwalker (Ref. 7). Both of these independent studies examined the question of time dependency of loading and disclosed that, in spite of the fact that relatively brittle bone material was involved, the tolerance to impulses of longer duration was less than that to short duration. Plotting the data of Fig. 20 in Ref. 6, and that of Fig. 11 in Ref. 7, on a log-log scale again discloses a trend of time dependency surprisingly close to that for internal head injury. The work of Swearingen included all major areas of the face including the forehead, while the latter plot (Ref. 7) summarized a larger number of blows of various kinds to the zygoma. In constructing the log-log tolerance plot from the latter reference, only the lower points, that is, the lowest striker forces which produced zygoma fracture at a given pulse duration, were employed. Both curves exhibit a slope as great as that represented by a 2.5 power weighting factor.

These data indicate, then, that there is considerable justification for employing at this time a 2.5 power weighting factor for indicating injury hazard inherent in impacts applied to the face. As for a tolerable threshold for damaging, (but usually survivable) injury to the face, one could then employ a Severity Index graded according to the effective contact area. In impact against certain parts of the interior of the body where it is impossible to insure survivability from internal head injury at the higher velocities, such an index could be used as a comparative check of the probability of damage to the facial bones and tissues in the lower speed accidents. The threshold curve for this type of injury would be as given in rudimentary form in Fig. 5. The injury number of 500 for facial bone fracture in impact over an effective area of 3 sq in. is a preliminary and conservative value based on Refs. 6 and 7, which show a tolerance disparity not as yet resolved.

CHEST IMPACT—As more experimental tolerance data are obtained in the future, it should be possible to recognize time dependency here. Since there is a possibility of transient resonance of the chest, it is necessary to look closely for this in conducting the tests on biological material upon which the Severity Index is to be based. This is done by comparison of the experimentally measured input function with the response function, the first being the force-time trace imposed upon the front of the chest and the second being the chest deflection or other measure of internal injury likely to occur. Chest impact tests with which the author has been associated (Ref. 8) have not as yet delineated an appreciable dynamic response; if further tests show that this is a small factor, then the input force may be regarded as a reasonable parameter for integration to obtain the Severity Index. (The peak value of input force is the current index for governmental steering qualification.) If further studies indicate, on the other hand, that an important dynamic factor is present, it will be necessary to employ a response parameter (for example, chest deflection) to best reflect the likelihood of internal injury. The procedure for developing a Severity Index is then to apply impacts having similar waveform but differing time durations, and solve for the weighting exponent which best approximates the time dependency.

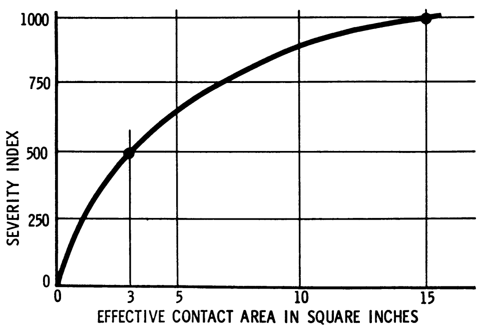

Fig. 5—General form of curve expressing lower tolerance of facial zone of head to impact over differing areas (damaging, but not fatal, injury)

Conclusions

A Severity Index has been developed for assessing impact test results which has proven useful in the following respects:

1. It is able to take time dependency of the injury into account in a manner which can be adjusted to that exhibited by the biological material involved, and permits comparisons between pulses of differing time durations experienced by occupants of automotive vehicles in accidents.

2. It permits comparison of the relative hazard between pulses of differing waveform or profile.

3. It provides a means for different investigators in different laboratories to make numerically consistent interpretations of the hazard represented by a recorded pulse.

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