A GENERALIZED ACCELERATION MODEL FOR BRAIN INJURY THRESHOLD (GAMBIT)

James A. Newman
Biokinetics and Associates Limited
Ottawa, Canada

ABSTRACT

A criterion for brain injury threshold is proposed which, for the first time, endeavours to take into consideration the combined effects of both translational and rotational kinematics. The validity of the model is assessed by way of a review of all known head injury data bases in which translational and rotational accelerations have been monitored. Available data appears to support no more than a simple linear proportioning of the two types of motion though a squared weighting, as originally proposed (and which intuitively is more plausible) also seems valid.

INTRODUCTION

The assessment of safety systems, and in particular, those associated with the use, and accidental collision of automobiles, requires the use of a criterion by which the relative likelihood of brain injury, or protection therefrom, can be evaluated.

The generalized model for brain injury threshold (GAMBIT), first introduced in 1985 [1], borrows from classical engineering treatment of the design of systems in which combined axial and shear stresses are both simultaneously generated because of the particular location and direction of the applied load. The premise for such an approach (in inanimate engineering systems) is that whatever the combination of normal and shear stresses, the material failure can be forecast on the basis of an assumed "equivalent" maximum principal or shear stress or strain. GAMBIT treats induced translational acceleration and rotational acceleration as if they could be regarded as stresses and considers that "failure" due to such combined loading is, in the case of the brain, brain injury.

The original GAMBIT equation is of the form:

$$G(t) = \left[\left(\frac{a(t)}{a_c} \right)^n + \left(\frac{\alpha(t)}{\alpha_c} \right)^m \right]^{1/s}$$

where $a(t)$ and $\alpha(t)$ are the instantaneous values of translational and rotational acceleration respectively and n, m, and s are empirical constants selected to fit the available data. $n = m = s = 1$ is a simple linear weighting of the translational and rotational components (G1). $n = m = s = 2$ provides an elliptical function in the two kinds of motion (G2).

The assignment of a failure criterion for brain injury, can be handled in a variety of ways. One could, for example, set some limiting value of AIS from 1 to 6. Alternatively, one might specify some particular injury, such as a mild concussion, as the appropriate limit. To a certain degree, this

matter is largely philosophic. In safety system evaluation, judgement as
to what system is acceptable and which is not, requires the definition of
an "acceptable" brain injury. In the case of brain injury, the GAMBIT
assigns to both the translational and the rotational acceleration a
limiting or critical value beyond which (in the case of "pure" inertial
loading) an "unacceptable" injury would occur. a_c and α_c are the limiting
"critical" values. These same limits are presumed to remain valid when the
brain is loaded by the combined effects of both forms of motion.

Regrettably, very little "pure motion" limiting data for brain damage has
been generated. In most experiments conducted to investigate "tolerance"
limits, both forms of motion have usually been present. Critical limits
will thus have to await more suitable experiments or must rely, as we shall
for the present, on extrapolation.

The one major difference between the classical engineering problem and the
one under consideration here is that brain injury is associated with
inertial loading which is time variant. Because of this, it has been
considered appropriate to continuously monitor the "equivalent load"
throughout the duration of the event. If at some time during the event the
value of G exceeds 1, a "failure" is noted. G = 1 thus is a boundary in
the a– plane beyond which the system fails and within which the system
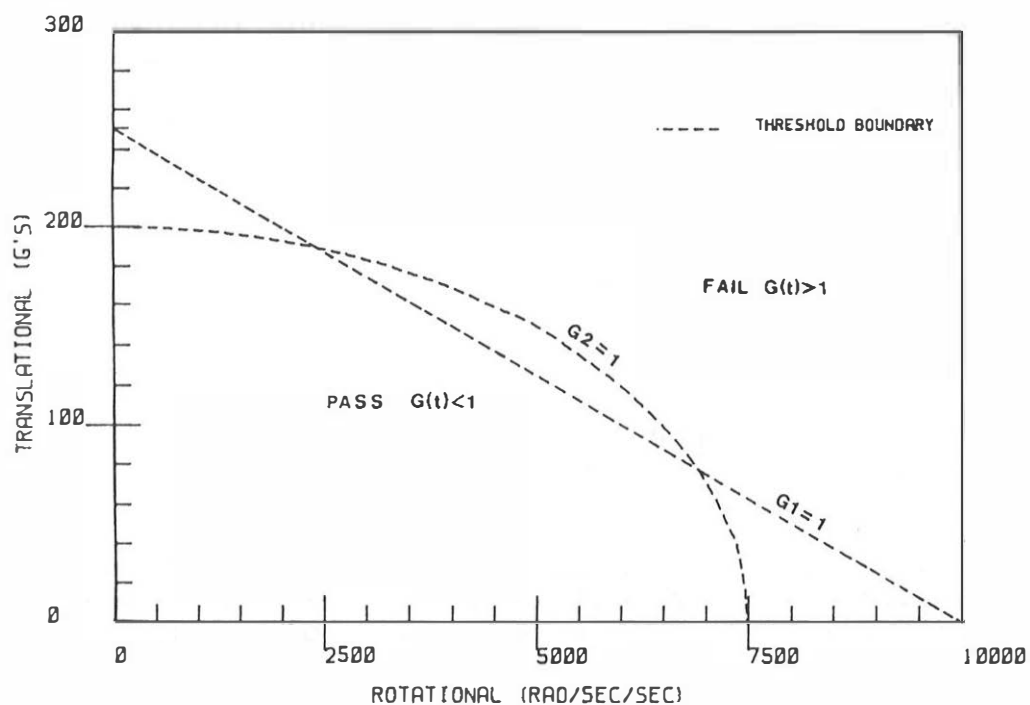passes. The concept is illustrated schematically in Figure 1.

Figure 1: Hypothetical GAMBIT Boundaries
G1 – Linear, G2 – Eliptical

MODEL VALIDATION

Data Suitability

Experimental evidence that might be used to validate any form of injury criterion has been characteristically confined to that gathered from cadavers, animals, volunteers and, to a limited degree, accident victims. The majority of published attempts to derive tolerance limits from such studies has considered injury mechanisms and their relation to translational or linear phenomena only.

The merits and limitations of each of the various surrogates have been reviewed extensively in the literature. In the present situation, these limitations are even more restrictive as GAMBIT requires information not only on injury and linear kinematics, but also on, or at least means to establish, rotational motion.

Whatever data there is, it is restricted in its usefulness by virtue of at least the following considerations:

1. Cadaver experiments can, at best, provide insight into brain injuries of AIS 3 or more only. Physical disruption of brain tissue may be observed in cadaver autopsies by the extravasation of fluid dyes injected into the arterial system before impact. However, so-called diffuse axonal injuries (DAI) associated with brain cell damage, which might appear as concussion or generalized diffuse brain injury are not visually evident.

2. Animal experiments do permit the observation of the effects of an impact resulting in minor injuries. Animals will exhibit concussion and/or temporary brain disfunction. However, of course, except through dubious methods of scaling, such data cannot provide numerical limits directly applicable to humans. Trends (if they exist) however, can be discerned and can lend support to (or discredit) a particular model.

3. Experiments with volunteers are always limited to non-injurious situations but, as such, can possibly provide lower bounds on tolerance limits.

4. Accident victims can be subject to the entire range of brain injury but, except for very special cases, are associated with too many unknowns to be of much value in a validation exercise.

Data Processing Difficulties

Until recently, only passing consideration was given to the effects that digital filtering of raw data has on correlations that might exist between various parameters. It is now becoming very apparent that processing of data through these means constitutes a somewhat subjective and sensitive form of data manipulation. To a great extent, the degree to which any correlation may be true, is highly dependent on the choice of filter type, cut-off frequency, etc.

Rotational accelerations must be computed from measured translational acceleration measured at known sites on a rigid body.

Nusholtz (2) and Burkhard (3) have both examined the effects of filtering on the computation of angular accelerations from linear data. The effects can be pronounced as illustrated in Figure 2. Here, translational acceleration has been cross-plotted against rotational acceleration for one of the cadaver tests described by Nusholtz et al (2).
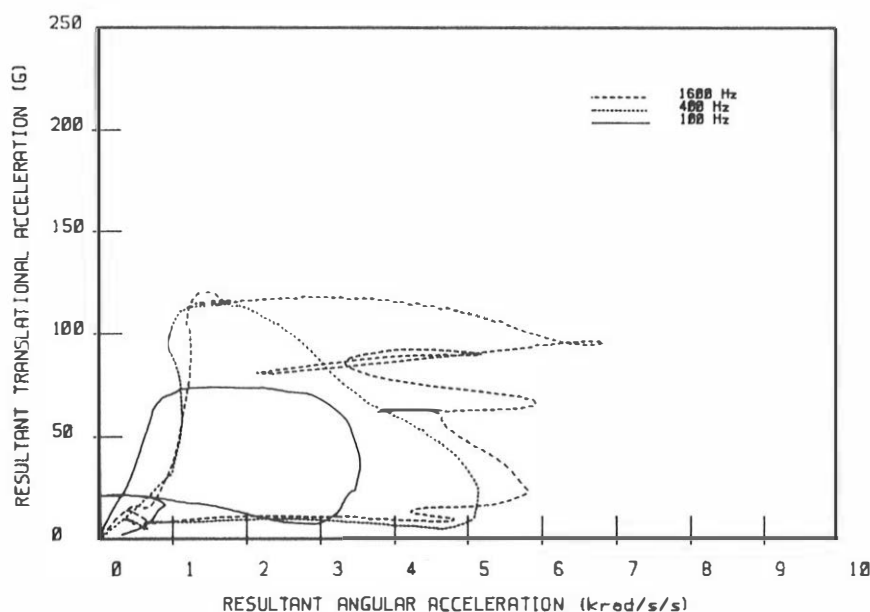
Figure 2: Effects of Data Filtering on GAMBIT Locus
(Data from Reference 2)

The data cannot be used directly for GAMBIT validation for these data are with reference to the origin of the instrument cluster not the anatomical center of the head. The same effects would however be expected at the head centre of gravity. Notwithstanding these effects, both authors seem to agree that, for cadaver and Hybrid III ATD heads, fourth order Butterworth filters with a cut-off frequency in the 400-500 Hz range on the linear accelerometer data provide a good approximation to the rigid body idealization of inertial injury criteria, (i.e., second and third order vibrational modes are effectively extracted from the data).

In addition, however, there are several inherent problems with linear axial accelerometery (3). First, one must have precise knowledge of the centre of gravity (mass) of the object, as the motion of this point dictates the nature of the computed rotational components. Secondly, accelerometer cross-talk, which may be as high as 5% with contemporary instruments, leads to serious errors in calculation of orthogonal direction accelerations. Thirdly, very precise calibration and alignment of the accelerometers is required. Slight errors in either of these can produce very large computational errors.
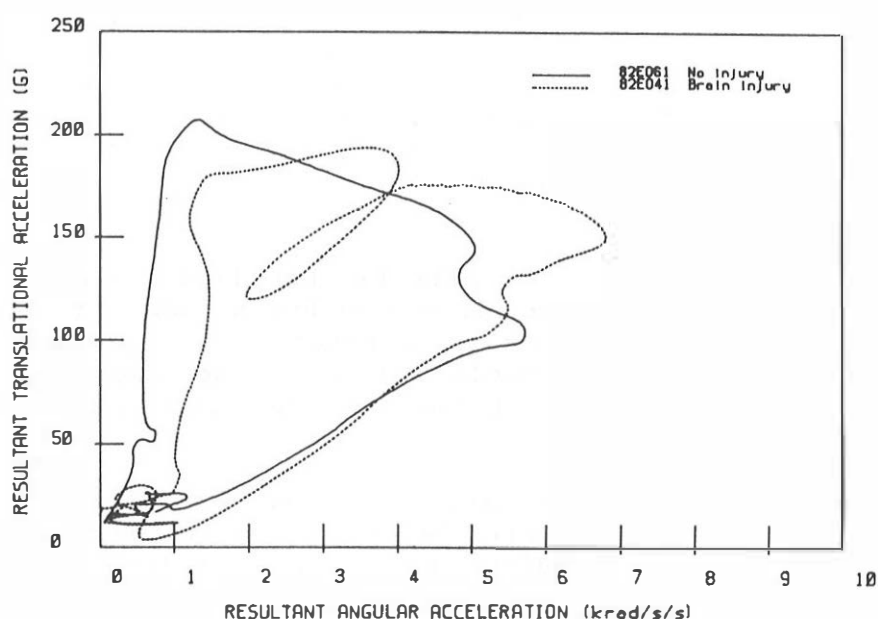
CADAVER HEAD IMPACT TESTS

In 1984, Nusholtz et al. (4), reported on nineteen tests conducted on nine different cadaver specimens. The reader is referred to the original publication for a full description of the test protocol and methodology.

Most significantly, a nine accelerometer cluster was employed and rotational accelerations were computed. Unfortunately (from the present point of view), the tests were not intended to provide kinematic-injury correlations. In most cases, the specimens were impacted more than once before autopsy was performed. One cannot, as a consequence, determine if an injury, if it is observed, is associated with the first or a subsequent impact or if indeed it represents the cumulative effect of multiple impacts.

Amongst the data, there exists two tests (82E041 and 82E061) that are of interest to the present study. Both tests involved identical initial conditions of padded forehead impact to a seated repressurized cadaver. In both cases the specimens were impacted twice at the same level. The data were processed with identical software and filtered at 800 Hz with a sixth order Butterworth filter (sufficient to exclude most of the spurious skull vibration response).

The 82E041 cadaver sustained subarachnoid hematoma at the right frontal lobe and subarachnoid hemorrhage in the parietal area. Specimen 82E061 sustained no injury. The GAMBIT locii for these two cases is shown in Figure 3.

Figure 3: GAMBIT Locii For Two Identical Cadaver Head Impact Tests (Data from Reference 4)

It surely can be argued that the differences are not great and that injury to one specimen and not the other is a reflection of some inherent differences in cadaver brain tolerances. Nevertheless, the subject sustaining the injury does have a locus which at high linear accelerations (140-190G's) has rotational accelerations which exceed that of the non-injured subject. One might consider further that injury to one specimen and not the other is an indication that the responses are "borderline" cases. The elliptical boundary first suggested in (1) lies remarkably close to such a border. Of interest to some readers will be the observation that in 82E041 the HIC was 1063 while for test 82E061 it was 1073.
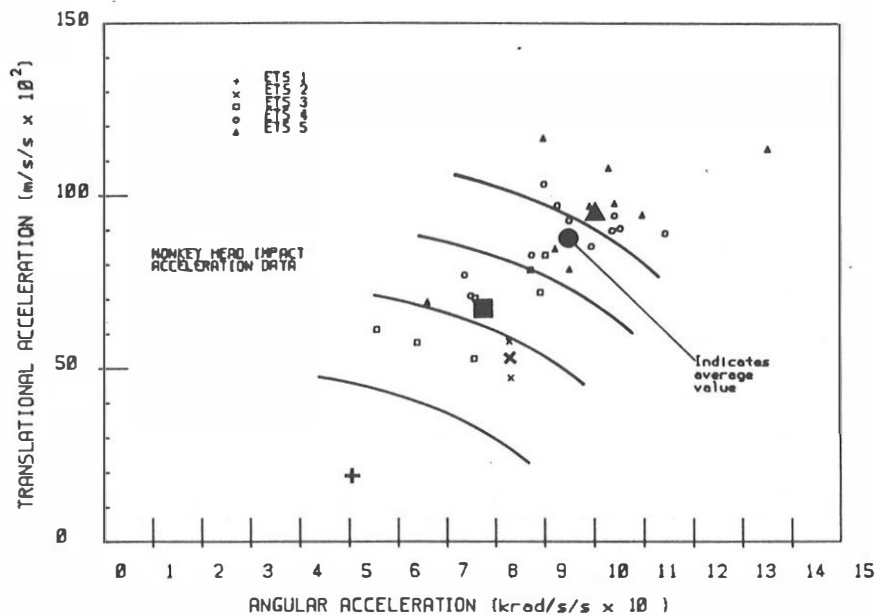

MONKEY IMPACT TESTS

In 1979, Gennarelli and co-workers (5) reported on the results of 30 rhesus monkeys subjected to an angular acceleration pulse in the saggital plane. The motion of the animal head was constrained via a linkage arrangement attached to a "helmet" fitted to the animal's head (thus avoiding direct skull contact phenomena). The level of rotational acceleration was varied by changing the effective radius through which the animal's head could rotate. The tangential component of acceleration was measured using a single linear accelerometer.

The purpose of their experiments was to classify the nature of the observed injury (i.e. no injury, frontal lobe contusion, temporal contusion) according to the mechanical imput. The injuries were graded according to an Experimental Trauma Scale (ETS) which is similar to AIS but for animals. For the present purposes, the data is of certain value.

The controlled nature of the motion causes the locus of translational versus rotational acceleration to simply be a straight line (i.e. tangential acceleration is always proportional to rotational through the fixed radius). As a consequence, peak rotational and peak translational acceleration always occur at the same instant in time. If the data supports the notion of combined loading, a correlation based simply on maximum values should exist.

To mask possible size difference effects, the data for each test was normalized to the mass of the average monkey brain. When treated in this way, and plotted in a GAMBIT plane, the results take the form shown in Figure 4. Shown there are the recorded values and the average for each ETS level. Also shown are lines that separate the various groupings into categories.

It is significant to note that the groupings of data do illustrate a trend. Furthermore, the trend is one which lends support to the idea that both rotational and translational acceleration must be considered when trying to correlate a certain injury level. The data suggests this trend since, on average, injury severity tends to increase as rotation increases for a fixed level of tangential acceleration. Additional data in the low translation-high rotation field would be helpful to further substantiate this observation. The data further suggests that a series of boundaries, similar to those proposed by GAMBIT, are in evidence.

Figure 4: Rhesus Monkey Head Data
(Data from Reference 5)

PIGLET HEAD IMPACT EXPERIMENTS

In 1984, Prasad and Daniels (6), reported on a series of 15 head impact tests on piglets. These tests were intended to provide insight into child head injury mechanisms through the use of the piglet surrogate. Instrumentation comprised a triaxial accelerometer attached to the animal's snout and a uniaxial accelerometer further out on the snout. This latter instrument allowed the computation of angular acceleration in the mid-saggital plane. Following impact, the animals were sacrificed and autopsies performed. Seven of the fifteen tests resulted in injuries to the animal's brain.

The relationship between the peak angular and average (calculated over the "HIC" duration) translational accelerations, and the levels of brain injury, normalized for brain mass, are displayed graphically in Figure 5. As in the case of the monkey data, the trends seem to support the basic premise of the GAMBIT.

127
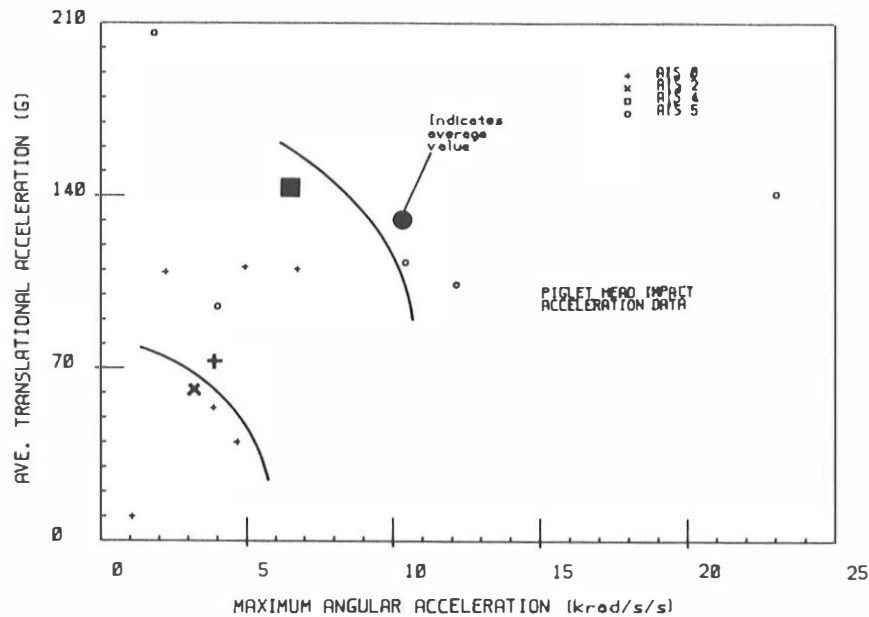
Figure 5:   Piglet Head Impact Data
(Data from Reference 6)

DISCUSSION

Simplification of Interpretation

Analyses of certain rotational/translational acceleration locii, indicates that, in spite of theoretical generalities, maximum translational acceleration and maximum rotational acceleration often do not occur at distinctly different times during the impact event. As a consequence, and considering the accuracy of the actual data, it is hard to justify the requirement to constantly monitor the accelerations throughout the entire pulse. In fact, it is probably sufficient to simply record the maximums of each variable and compute G using those precise values.

The merits of this are that it does not presume a particular instant when a "critical" situation occurs and, at the very least is a conservative approach to the definition of a combined criteria. That is, G so computed would be the value that would apply if in fact, maximum translational and rotational accelerations did occur at the same instant.

Boundary Uncertainty

No precise separation between a non-injurious and an injurious situation can be defined. The location of any such boundary must be a matter of probability. That is, a given set of inertial conditions carries with it only a certain likelihood of injury depending on a variety of variables.

128

Age of the victim, and other physical and morphological variables, will determine whether in fact a head injury will occur.

One possible way to address this matter is to replace the GAMBIT boundary with a region or band that is associated with such probabilities.

Directional Sensitivity

It has been suggested, with some justification, that the tolerance of the head to impacts from different directions are different. Such behaviour is thought to be observed by noting, for example, that for the same brain injury severity, the head accelerates at a lower level for lateral impacts. Presumably if such behaviour is associated with translational motion, it would also be true for rotation. Generally, one might have "critical" accelerations for each of the six degrees of freedom. To deal with this possibility, one could, with the same engineering justification used to define G, define "equivalent" translational and rotational accelerations. These would take the form:

$$\left(\frac{a}{a_c}\right)_{equiv} = \left[\left(\frac{a_x}{a_{cx}}\right)^2 + \left(\frac{a_y}{a_{cy}}\right)^2 + \left(\frac{a_z}{a_{cz}}\right)^2\right]^{1/2}$$

where the a_c's represents the orthogonal critical values and a_x, a_y, a_z, are the component values in the three principal directions.

In general, this function could be computed as a function of time. A similar expression can be written for the rotational terms. The left hand sides of both expressions would replace the corresponding terms in the current GAMBIT formulation. Though technically appropriate (an "ellipsoid of influence" has been suggested before), such refinement is well beyond the range of currently available data.

Time Dependency

No explicit time dependence on translation or rotation has been agreed upon. However, there is the frequent suggestion that allowable average acceleration decreases with the passage of time. The Wayne State Curve is the classic example of this thinking. Should such a relation be verified for both forms of motion, the GAMBIT boundary could be considered as a 3 dimensional surface instead of a simple line. Figure 6 illustrates the concept.

Whether this "improvement" could ever be validated is a matter for future speculation.

SUMMARY AND CONCLUSIONS

For years it has been assumed that there has been a relationship between the head injury that a human being will sustain and the motion that his head undergoes as a result of some mechanical input. The GAMBIT evolved by making fairly well-accepted engineering judgements about the significance

of certain parameters. It assumes, in particular, that the onset of brain injury occurs when the combined effect of translational and rotational accelerations exceeds some limiting value.

The present study has attempted to pull together recent knowledge pertaining to brain injury that is induced in a generalized inertial field. The objective of the work has been to provide a technically realistic criterion for use with ATD's in a car crash environment.

The state-of-the-art in human brain injury criterion definition continues to not make significant progress. The existing data base is simply insufficient to expect major strides and limitations of available experimental data continues to thwart efforts in this area. As more knowledge in this area is amassed, the greater becomes our understanding of the limits of this knowledge. However, engineers typically do not wait for science to completely clarify all issues before proceeding with their designs. Thus in recognition of, and in some cases in spite of, the above limits, the GAMBIT model has been postulated. With respect specifically to the shape of the GAMBIT boundary and of the values of its intercepts, the available data is clearly limited. It is limited particularly in those regions of high translation/low rotation and high rotation/low translation; regions which would help define both the intercept values and the general shape of the curve. At present, an eliptical shape (consistent with the maximum shear stress theory) appears no more reasonable than a simple straight line.

A straight line intersecting the translational acceleration axis at 250G's forms a triangular region within which injuries to cadavers are typically not found when the rotational acceleration intercept is approximately 10,000 rad/sec/sec. The 250G limit is consistent with the failure criteria employed in helmet evaluation where headform motion is restricted to being purely linear. No similar rationalization can be provided for the pure rotational limit, however, the 10,000 rad/sec/sec does not seem inconsistent with the range of values which have been suggested at various times by others.

The revised GAMBIT thus simply becomes:

$$G = \frac{a_m}{250} + \frac{\alpha_m}{10,000} \leq 1$$

References

1.   Newman, J.A., "A Generalized Acceleration Model for Brain Injury Tolerance". Presented at the Winterlude Head Injury Workshop, Ottawa, Feburary, 1985.

2. Nusholtz, G.S., Kaiker, P.S., Muscott, G.J. and Suggit, B.R., "UMTRI Experimental Techniques in Head Injury Research", SAE Paper #851244. Presented at the Government/Industry Meeting, Washington, May, 1985.

3. Burkhard, P.M., "A Two Dimensional Accelerometer Analysis Applicable to Impacts", General Motors Report GMR-4765, June 28, 1984.

4. Nusholtz, G.S., Lux, P., Kaiker, P. and Janicki, M.A., "Head Impact Response — Skull Deformation and Angular Accelerations". SAE paper #841657, Proceedings of the 28th Stapp Car Crash Conference, Chicago, November, 1984.

5. Gennarelli, T.A., Abel, J.M., Adams, H., and Graham, D., "Differential Tolerance of Frontal and Temporal Lobes to Contusion Induced by Angular Acceleration". SAE Paper #791022, Proceedings of the 23rd Stapp Car Crash Conference, SAn Diego, October, 1979.

6. Prasad, P. and Daniels, R., "Biomechanical Analysis of Head, Neck and Torso Injuries to Child Surrogates Due to Sudden Torso Acceleration". SAE paper #840656, Proceedings of the 28th Stapp Car Crash Conference, Chicago, November, 1984.