

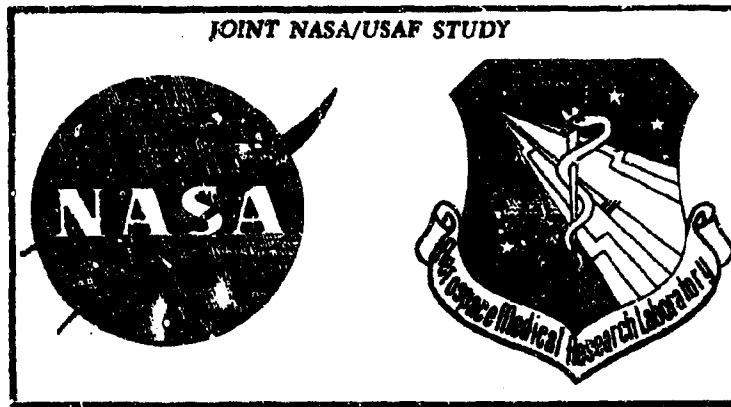
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DYNAMIC MODELS OF THE HUMAN BODY

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JOINT NASA/USAF STUDY

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Foreword

This study was initiated by the Vibration and Impact Branch, Biodynamics and Bionics Division of the Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio, with the support of the Biodynamics Section, Environmental Physiology Branch, Life Systems Division, National Aeronautics and Space Administration - Manned Spacecraft Center, Houston, Texas. The research was conducted by Frost Engineering Development Corporation, 3910 South Kalamath, Englewood, Colorado, under Contract AF 33(657)-9514. The work was performed in support of Project No. 6301, "Aerospace Systems Personnel Protection," and Task No. 630102, "Personnel Protection in Aerospace Systems," with Mr. Peter R. Payne as principal investigator. Mr. James Brinkley of the Vibration and Impact Branch was the contract monitor for the Aerospace Medical Research Laboratory, while Mr. Harris F. Scherer and Mr. Jack Rayfield were the NASA liaison representatives.

This report is one of a series of reports generated in the area of human restraint and support system dynamics under Contract No. AF 33(657)-9514. The research in this report was performed from September 1962 to February 1964.

This technical report has been reviewed and is approved.

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Section 1

INTRODUCTION

The designers of restraint systems, ejection seats, escape capsules, and space vehicle landing systems are concerned with optimizing the performance of these systems without endangering the human occupant. The immediate design problems must be solved while basic research is still being pursued, requiring the research community to establish interim tolerance limits to minimize the users' risks and the designers' limitations simultaneously.

Establishment of human tolerance limits has been a necessity for environments other than exposure to acceleration and, in virtually every case, a common problem occurs. It has been customary to set a tolerance limit as a specific and arbitrary value, even though differences in individuals and variations in exposure circumstances will produce gradations of damage, injury and/or disease on both sides of the tolerance limit unless it is set very low. Similarly, in many cases the tolerance limits must be established prior to gaining a broad enough knowledge of the physical, chemical, and biological factors involved in the situation, with the consequence that a large safety factor should be included in the limit set. The combined effect can be to impose too severe a restriction on the performance of the system; conversely, ignoring these factors can result in establishment of a tolerance level that is too high.

In the case of human tolerance to acceleration, there have been occasions where it has not seemed feasible to incorporate any considerable margin for error. Consider for example, the problem inherent in the design of early ejection seats. To accelerate such escape devices rapidly enough to clear the aircraft tail at high flight speeds with the short catapult stroke then available, the designer had to impose spinal accelerations close to 20 G. The resulting acceleration environment caused the vertebrae to compressively fracture. Since staying with an airplane after a serious malfunction or making an unaided bailout was known to be lethal in a high percentage of cases, risking some injury from the ejection appeared to be the lesser of two evils.

Implicit in the above situation are the concepts of risk and probability. Presumably, they were regarded only qualitatively, if at all, and for design purposes a single acceleration limit was set. Such a single limit implies that no one will be injured who is subjected to an acceleration below the limit value. In actual fact, such a situation does not exist, and the probability of injury increases with increasing accelerations.

Had the concept of (and the ability to predict) probability of injury existed at the time the single acceleration tolerance limit was set as a guide for ejection seat and catapult designers, a lower limit would have been established. Unfortunately, it took the experience of actual injuries to produce this result. In the interim, ejection seats acquired a reputation that caused many pilots to use them only as a last resort, when the flight conditions were too often beyond the performance capability of the seat.

By applying the analytical techniques described herein to the data available from experiment, experience, and many other sources, it is possible to predict and/or evaluate acceleration tolerances as quantitative risk levels, expressed in terms of probability of injury.

The risk or probability approach to tolerance can be a useful tool for those who establish and apply acceleration tolerance data. Two simple examples will serve to illustrate this point. As noted above, the ejection seat designer is concerned with removing the crew of a disabled aircraft from a potentially lethal situation. This implies that a reasonable risk of injury is acceptable in the ejection.

tion sequence, since the alternative is the certainty of much more severe injury and the strong possibility of death. On the other hand, the designer of a space vehicle landing system must virtually eliminate the possibility of injury during the course of normal operational landings.

A single parameter should be used to specify injury or damage, since two parameters would involve a large set of combination of numbers to describe the tolerance situation. Fortunately, this limitation to a single injury parameter is not really restrictive. Damage to biological tissue and structure from imposition of accelerations must be assumed to be caused in fundamentally the same manner as damage to mechanical systems, i.e., through the application of forces that result in such deflection or strain that failure of the structure or material being loaded finally occurs. Thus, if the characteristics of the structure can be defined, and the maximum force applied to the structure can be ascertained, it should be possible to predict whether or not failure will occur.

As used in this report, the term "dynamic model" refers to a mathematical analogy to a lumped parameter mechanical system comprised of separate mass, spring, and damping elements. The purpose of the dynamic models is to permit prediction of the peak forces exerted within the body on critical structural items. There is, of course, a relatively wide variation in the characteristics of these structures and data is not yet adequate in certain areas of interest. Equations can be developed that describe the behavior of such a system under the influence of any input acceleration or force. Once developed and proved valid, these models are a useful tool for rapid determination of the effects of variations in the disturbing forces or the components of the system or both. Adequately accurate investigations can be made quickly and economically, for the dynamic models lend themselves well to performance of laborious calculations by analog or digital computers.

Despite the complexity of the human body and the variation between individuals, the response of the human body to acceleration environments can be approximately characterized by an analogous dynamic model. Therefore, if body characteristics in these respects can be established, the techniques of dynamic analysis may be applied to the gain the benefits already obtained in the engineering disciplines.

One of the major contributions to be expected from this approach is the development of tolerance data in the form of relative probability of injury curves instead of the single-line graphs or single critical G values used heretofore. The importance of this cannot be over-estimated: although the engineer needs a single value upon which to base his design, he certainly is not qualified to establish it. Neither is this the province of the experimenter, because he is producing information that is based upon too few sample and too-special situations, in general. Instead, it is up to the user of the end product to select the limit to be used, and this in turn should be dependent upon knowledge of the risks involved. Given the relative probabilities of injury for different levels of accelerations exposure, and knowing the many factors involved in the missions that will create these exposures, he can make an intelligent value judgment.

Working from the experimental data with analog or digital computation techniques developed as an outcome of the dynamic models, these probabilities of injury can be computed. In addition, irregular and complex acceleration-time histories, impractical to duplicate experimentally, can be evaluated for their effect on the human involved. When such calculations show unacceptable probabilities of injury, the parameters involved can be varied singly or in combination to derive knowledge of the modifications necessary to make the acceleration exposure endurable. It will always be wise to verify these predictions with tests simulating at least the critical points, but the savings are still considerable.

Since this report is intended as a general treatment of the subject, numerical examples of specific acceleration and restraint system design problems are not presented.

The acceleration tolerance problems that are of major concern at present involve only seated or reclining subjects. The head, arms, or legs may or may not be included in the masses that make up a specific dynamic model, depending on whether they are factors in response of the body to the acceleration situation under study. The models of the human body used in this report generally refer to a whole body response based on a particular failure mode with no special regard to the response characteristics of each body segment.

Section II

BASIC BODY DYNAMICS

THEORETICAL DYNAMIC SYSTEM

Human body dynamics has a simple basis: Biological Structures (and the human body in particular) respond to applied loads and accelerations in exactly the same manner as any other physical systems. Thus the logical starting point in the discussion of basic body dynamics is the fundamental consideration of the response characteristics of a mechanical system.

If a rectangular acceleration pulse is applied to a simple spring-mass system, the response of the system is as illustrated in figure 1. Note that the peak acceleration experienced by the system is greater than the peak acceleration of the input pulse. This phenomenon is termed the *dynamic overshoot* of the system.

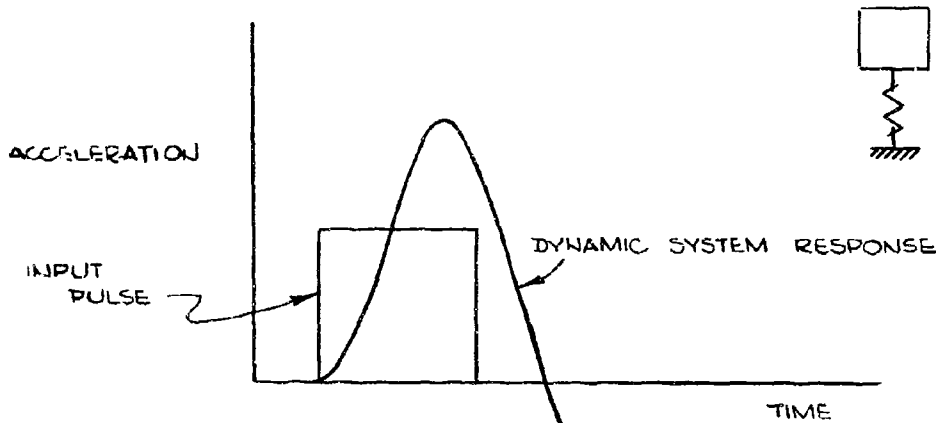


Figure 1. Response of a Simple Dynamic System to a Rectangular Pulse.

If the peak force in the spring of the dynamic model is recorded in a series of tests in which the duration of a rectangular pulse to the system is progressively increased while the acceleration level remains constant, the result will be a graph such as that shown in figure 2. At zero pulse duration the peak force in the dynamic system will also be zero since no force can be generated in zero time. As the pulse duration is gradually increased from this minimum value, the peak force will continue to increase up to a certain peak value, and then will continue at that value for longer and longer pulse durations.

The type of curve shown in figure 2 is characteristic of any single degree of freedom system. The pulse duration at which the force in the system levels out at a constant peak value occurs around the natural period of the system. Beyond this critical duration, the peak force in the system remains constant as pulse duration increases. Both theory and experiment show that in an undamped system, the actual value of the force will be twice the input force. That is, the dynamic overshoot will be 100% for pulse durations that exceed the natural period of the system.

Below this critical pulse duration, the peak force in the system will be a function of the area under the acceleration-time pulse, which is equal to the velocity change incurred by the system as a result of this acceleration.

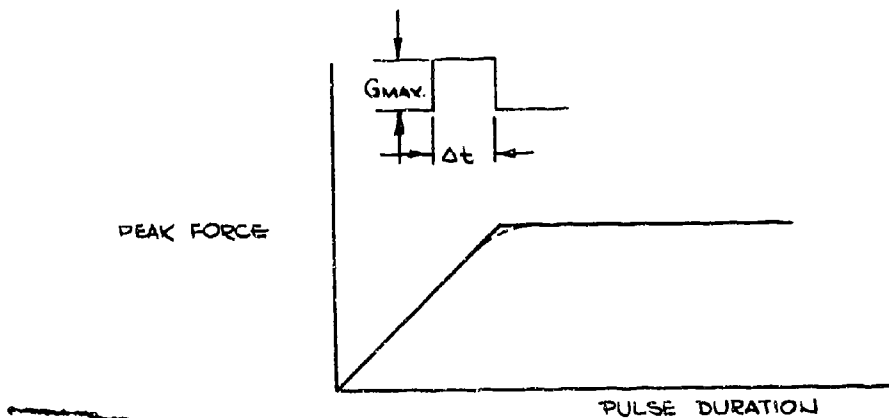


Figure 2. Peak Force in a Dynamic System for Various Rectangular Pulse Durations.

Two simple rules emerge from the foregoing: The peak force in a dynamic system is related to the *velocity change* involved in the acceleration pulse when the pulse duration is less than the natural period of the dynamic system. The peak force in the dynamic system is related to the *peak acceleration* value of the input acceleration when the pulse duration is greater than the natural period of the dynamic system. For durations comparable with the natural period, the peak force is a complex function of both velocity change and acceleration magnitude.

Now if the spring in the system always breaks at a given force level, the system will be damaged any time the force in the spring exceeds this critical value. Knowing the response of a dynamic system to input acceleration forces, it is then possible to relate the damage incurred to the input pulse parameters. For the rectangular pulse shape, it is necessary to consider only the peak acceleration and the pulse duration. Figure 3 depicts the resultant tolerance graph plotted on a log-log scale.

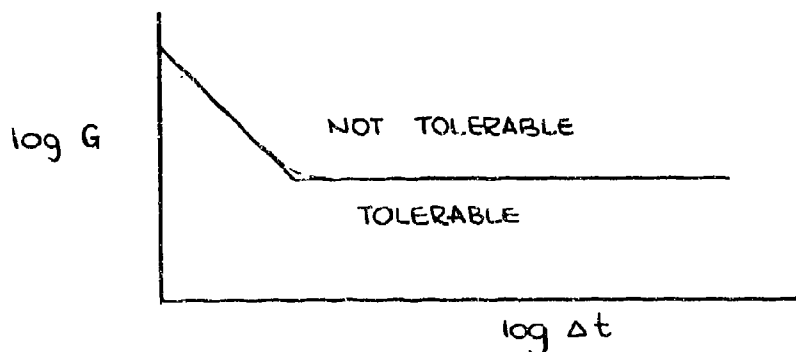


Figure 3. Tolerance Graph for a Simple Dynamic System Subjected to Rectangular Input Pulses.

In this figure, the line at the left of the graph that angles down at 45° represents the regime in which velocity change determines the peak force in the spring, while the horizontal line extend-

ing to the right of the graph represents the region in which peak acceleration is the governing factor.

Figure 3 can be considered typical of tolerance curves for single degree of freedom systems subjected to rectangular acceleration inputs. Such a case is of interest as an illustration of the analytical technique but, for practical problems regarding human body dynamics or restraint system design, it is necessary to take into account such factors as damping, statistical variabilities, input pulse irregularities, and similar factors.

Damping exists in all real materials as evidenced by the simple fact that all dynamic systems come to rest after excitation, rather than oscillating indefinitely. The concept that is most useful in human body dynamics is that of viscous damping, which is characterized by a linear proportionality between the damping force and the velocity of the various parts of the system with respect to each other.

The effect of damping on a simple dynamic system is illustrated in figure 4, where the amplitude of the system's oscillation decreases with each cycle until the system comes to rest. In response to an acceleration pulse, the damped system exhibits less overshoot than an undamped system, as illustrated in figure 5.

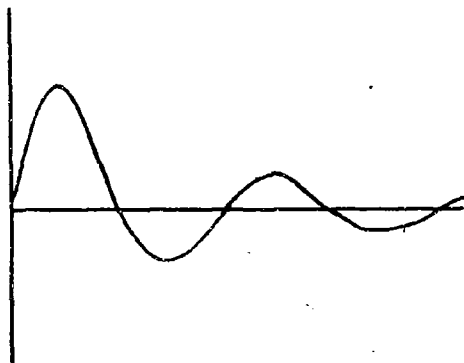


Figure 4. Damped Oscillatory Motion.

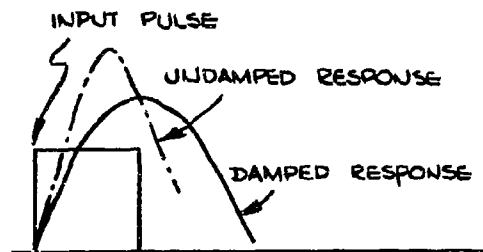


Figure 5. Effect of Damping on Response to a Rectangular Input Acceleration Pulse.

Since a damped system exhibits less overshoot than an undamped system, the effect on acceleration tolerance is quite straightforward. A smaller amount of overshoot implies that a greater input acceleration can be applied to the system before reaching the critical force level in its "spring." (This term will be used henceforth to express the elasticity characteristic of the human body.) The effect of the reduced overshoot on acceleration tolerance is shown in figure 6.

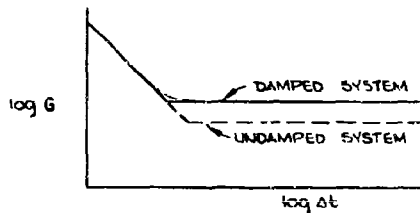


Figure 6. Effect of Damping on System Acceleration Tolerances.

The addition of a viscous damper to the dynamic model results in the damped single degree of freedom system illustrated schematically in figure 7.

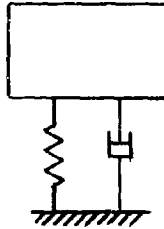


Figure 7. Damped Single Degree-of-Freedom System.

VALIDATION BY EXPERIMENT

All real materials exhibit a marked variability in breaking strength, a fact that is recognized in most applications by designing on the basis of allowable strengths which are lower than those which are obtained in tests. Statistical procedures often are used to estimate the lowest strength limit, which is then used as a design allowable. The same sort of situation exists, of course, in biological structure and must be taken into account when discussing the dynamics of the human body. The point that must be borne in mind at all times is that there is a probability of injury associated with a given input force-time history.

Not only do variabilities exist in the breaking strength of materials, but in dealing with the human body the variabilities in bone stiffnesses and body weights must be considered. The variability in these factors result in variations in the response to a given input acceleration and consequently in the peak force occurring in the human structure.

The concept of variability can be illustrated as shown in figure 8 in which statistical distributions are superimposed on the so-called tolerance line that, in this case, is the 50% probability of injury line. Another way of illustrating this is presented in figure 9 in which a series of probability of injury lines are drawn. (Note that the concept of injury probability is quite similar to that of the Lethal Dose used in any physiological and medical studies.)

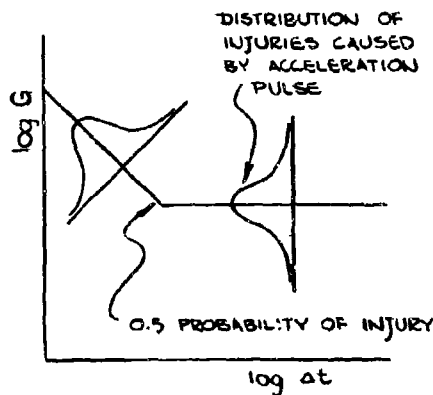


Figure 8. Statistical Distribution of Injuries.

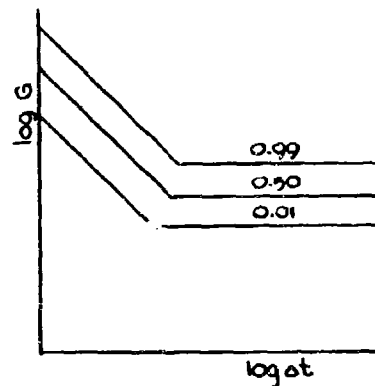


Figure 9. Probability of Injury Levels on a Tolerance Plot.

In the preceding paragraphs, the fundamental nature of physical systems consisting of elements with elasticity and damping were explored. Measurements have been made on various types of mechanical systems to determine sensitivity to impact accelerations, and the results confirm the results just presented. There is, of course, always a question as to whether biological structure in complex arrangements follows simple physical laws, although there is no obvious reason why such laws should not apply.

It should be relatively easy to demonstrate that biological specimens exhibit the same characteristics as mechanical systems, in terms of impact sensitivity. The only experimental equipment required is a drop test rig with some means of modifying the impact acceleration to obtain various peak accelerations and pulse durations. Live animal subjects can then be tested on the rig to determine injury or fatality levels. Such experiments have been conducted by Kornhauser and Gold as reported in reference 1.

A total of 329 mice were subjected to impacts in two series of tests to determine the levels at which approximately 5% and 95% of the test subjects died (5% and 95% Lethal Dose or LD levels). The drops were conducted so that the mice received positive spinal acceleration pulses that varied from slow to moderate rate of onset. The results of these mice impact tests are plotted in figure 10, adapted from reference 1. The points do in fact fall on lines that are either at 45° to the horizontal or are horizontal. This is just as predicted by the dynamic theory.

There is an unusual feature in the data that has not previously been discussed. Figure 10 shows two different 45° angle lines representing two different critical velocities and two different horizontal lines representing two different peak accelerations. Hypothetically such a situation can

result from three causes: An artifact may have existed in the test situation; the mice may have constituted a two-degree-of-freedom model; the data may be showing the occurrence of two different failure modes. There is some question as to which of the causes listed results in the shape shown for the mouse impact tests, but as verification of the applicability of dynamic theory to biological tissue as well as mechanical materials, the mouse impact test results are probably valid as an illustration of the basic theory.

There remains the basic problem of relating an input acceleration-time history to injury or some other practical limit. Reflecting for a moment on the load-deflection curve characteristics of materials, a force (or its equivalent deflection) is clearly the mechanism of damage in a material. Therefore, if interest is centered on tolerance to acceleration, the only parameter of interest is the peak force developed in the elastic material of the human body. This force is most conveniently described by a parameter called the Dynamic Response Index (DRI).

For simplicity and consistency, the DRI is expressed in nondimensional G units by dividing the peak force in pounds by the body weight in pounds. In all subsequent discussion in this report, the peak force in the body will be referred to in terms of DRI.

The best example of the DRI in the human body is for the case where the acceleration is directed parallel to the spine. In this mode of acceleration of the body, the vertebral column acts as the principal load-bearing member, with the mass of the thorax, thoracic organs, arms, shoulders, neck, and head resting on the column. Actually, these masses are attached along the entire length of the spinal column, which means that the first analogy that comes to mind is a distributed parameter dynamic model, i.e., one in which an infinite number of individual masses, springs, and dampers are connected in series. However, there is a lack of data on the precise manner in which the masses are distributed, and in addition, it is very difficult to calculate the forces in a complex

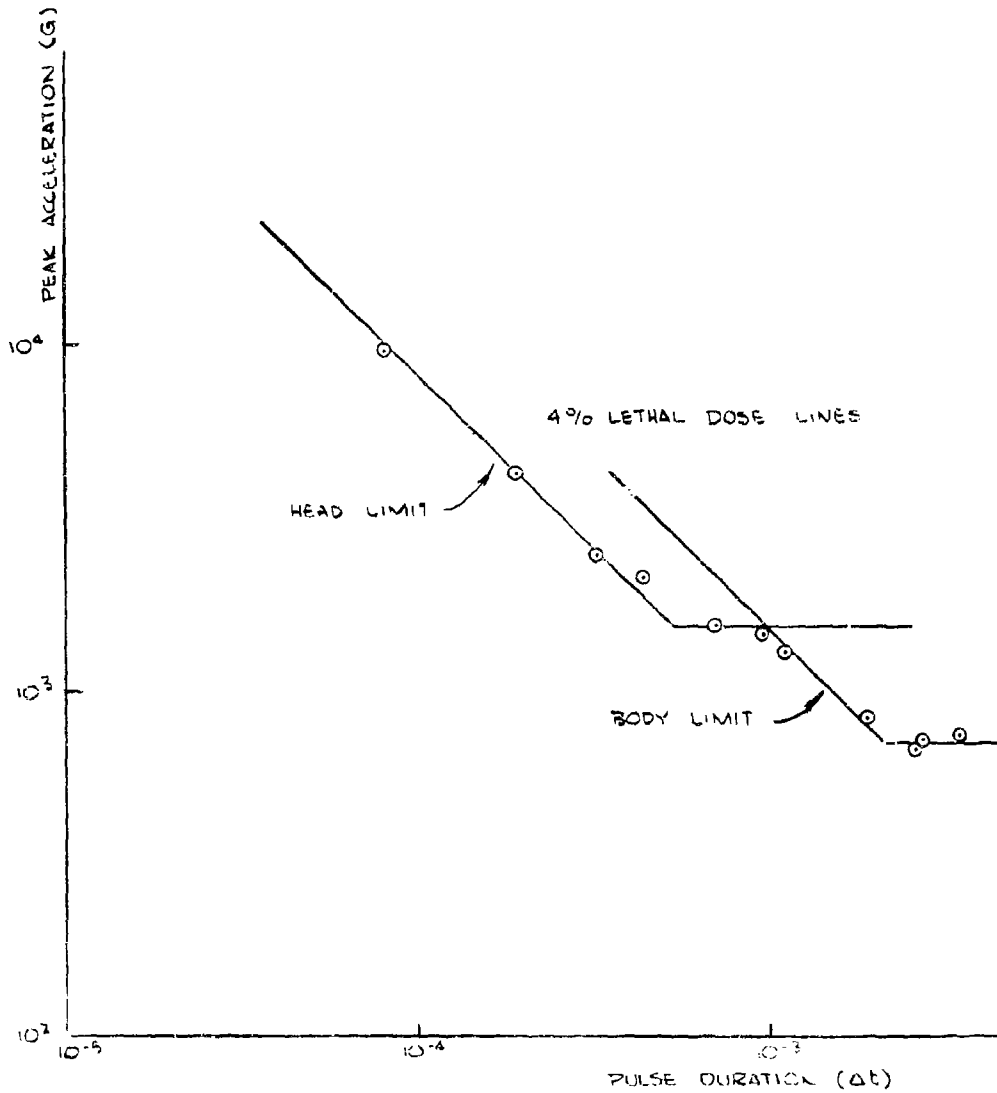


Figure 10. Four Per Cent LD Lines for Mice Drop Tests.
(From Kurnhauer and Gold)

distributed parameter system. This does not mean, however, that useful information cannot be developed.

Instead of a distributed parameter system, the upper body analogy can be simplified as a lumped parameter system, this means that all the mass is assumed to rest upon the top of the thoracic and lumbar vertebrae which constitute a simple massless spring. In a later section this analogy provides good correlation with empirical measurements of body dynamics. Thus, only damage to the vertebral column need be considered in developing the dynamic model — at least initially.

Section III

APPLICATION OF HUMAN BODY DYNAMICS

The correct interpretation and use of engineering information is predicated upon presenting it in a manner that enables users to convert and use the data readily. One of the major advantages of using the dynamic model approach to the problem of acceleration tolerance is that it permits the use of a standard and well defined vocabulary. Furthermore, the relative effect of various parameters can be evaluated, and the relative importance of the parameters investigated quantitatively. This section describes the appropriate parameters that need to be taken into account when human tolerance to acceleration is discussed.

INPUT ACCELERATION PARAMETERS

The input acceleration-time history can be characterized by three parameters if it is rectangular or triangular in shape: the time duration involved, the peak value of the acceleration applied, and the time required to get from the initial acceleration condition to the peak value (the rise time). In the simplest cases, these parameters are all represented by single values so that linear relationships are assumed. In most real acceleration-time histories, however, irregularities occur as a matter of course.

The simplest acceleration pulse to evaluate is the rectangular pulse, characterized by a duration and a plateau acceleration value, the rise time being zero.

As explained in a preceding section, there are two acceleration tolerance regimes. If the acceleration pulse duration is extremely short, then the peak force in the dynamic system is simply a function of the area under the pulse, which is the velocity change involved in the acceleration. Beyond a certain duration (several times the natural period of the system), the force in the dynamic system becomes a function of the peak acceleration rather than the area under the pulse. As a consequence of this, the curve that represents a given level of damage to the dynamic system can be drawn as in figure 11. The area below and to the left of the lines in the graph represents rectangular acceleration pulses that will cause less damage than the nominal value for the line. Conversely, points above and to the right of the lines represent pulses that will cause more damage.



Figure 11. A Tolerance Graph for Rectangular Pulse Inputs.

The curve shown above can be considered a tolerance graph only for rectangular acceleration pulses. When used in such cases, however, the curve should be labeled a damage or injury curve.

rather than a tolerance curve, and the restriction that the curve applies to rectangular pulses only should always be noted on the graph.

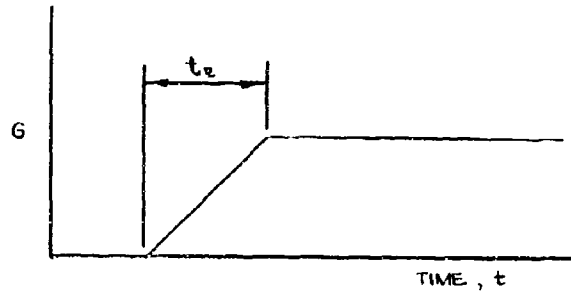


Figure 12. A Linear Ramp Pulse Showing the Rise Time (t_r).

Pulses with a ramp leading up to the peak acceleration, such as that shown in figure 12, can be specified in terms of rise time as well as peak acceleration and duration. In general, the longer the rise time the lower the peak force obtained in the dynamic system for a given plateau acceleration. With a rectangular or zero rise time pulse, the dynamic system will experience 100% overshoot if is not damped. If a pulse with a sufficiently long rise time is applied to the same dynamic system, no overshoot at all will occur. Thus, for a range of rise times, various amounts of overshoot will occur.

Since a smaller amount of overshoot occurs as the rise time lengthens, the effect on an injury or damage line is as shown in figure 13. That is to say, the plateau acceleration required to cause damage will be higher than those for the rectangular pulse condition, and an increase in rise time permits the application of larger plateau accelerations. Obviously, rise time cannot affect the area where velocity change is critical, since by definition very long rise times would not permit the very short pulse durations associated with this regime.

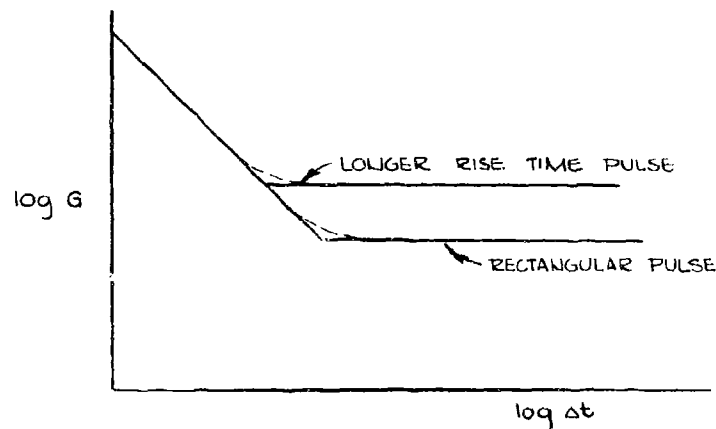


Figure 13. Effect of Rise Time on Tolerance.

A tolerance graph should really include all the rise times of interest to the engineer. Such a graph would look approximately as shown in figure 14, where each line represents a different rise time. The uppermost line represents the application of acceleration in a device, such as a centrifuge (very long rise times) while the lowest line represents the rectangular pulse condition.

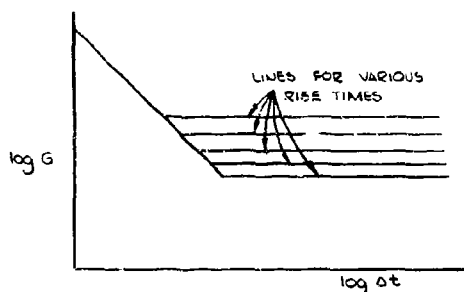


Figure 14. Illustration of Problems in Plotting Tolerance Curves for Various Rise Time Pulses.

Even with the addition of rise time to the problem of assessing acceleration pulse severity, real acceleration pulses will not always look like the trapezoid that will be drawn if only a rise time, peak acceleration, and duration are specified. Most real acceleration pulses involve irregular force-time relationships. In such cases, it is sometimes sufficient to approximate the real acceleration pulse with a mathematical function, such as a half sine pulse, versed sine pulse, an exponential acceleration onset, and so on. Several mathematical function pulse shapes have been investigated (ref. 23 and ref. 24) and in figure 15 the shape of the damage or injury curve for the half sine pulse is shown.

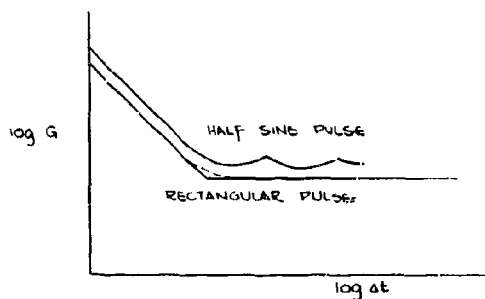


Figure 15. Tolerance Curve for a Half-Sine Pulse.

While it is possible to approximate real acceleration pulses with an appropriate mathematical function to obtain more precise results, no single graph can contain all the lines that would be required to show the damage levels for the various pulse shapes. It would become necessary then to use a number of sequence graphs to represent the potential injury or damage to a dynamic system when dealing with a number of different types of pulses and pulse shapes.

Practical acceleration-time histories usually exhibit irregularities that cannot be described by a simple mathematical function. Solutions must then be obtained numerically using a computer.

HUMAN BIOMECHANICAL PARAMETERS

As in the first approximation the natural frequency and damping coefficient of the human body, for a given mode of deflection, can be assumed to be invariant. However, the human body probably has an infinite number of deflection (or injury) modes for any given acceleration vector of which the first three to six may be significant in a lumped parameter analysis. For example, in the application of spinal acceleration to a seated human body, the fundamental mode of failure is vertebral damage, but a secondary mode in the form of head injury can occur for very short rise times. Similarly, in a transverse direction, the principal type of damage appears to be hemorrhaging in the lungs and damage to the large blood vessels around the heart. However, transverse acceleration can also result in damage to the head, liver, spleen, kidneys, and so on.

In figure 16 a multiple injury mode situation is illustrated. In the hypothetical case use for this, there are three primary injury modes. Note, however, that a number of secondary injury modes may exist, so that exceeding a given velocity change or peak acceleration can result in

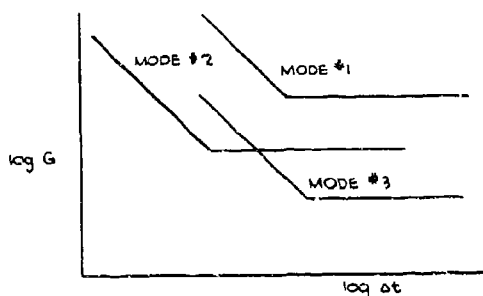


Figure 16. An Example of Multiple Injury Modes.

multiple injuries. Each injury mode represents the deflection of a piece of tissue or structure to a point of passive damage where immediate or subsequent functional or physical failure occurs. Thus, each injury mode represents a separate kind of tissue deflection. It is obviously desirable to have a dynamic model available for each injury or deflection mode.

Unfortunately, insufficient experimental data are available at present to describe the dynamic characteristics of all the various tissues involved in the injuries sustained by the human body when it is subjected to large accelerations. Far from precluding the use of the dynamic models, this situation emphasizes the use for them, because they furnish the most accurate description available of dynamic response to acceleration and a logical basis for extrapolation when it is necessary. Just as aircraft were designed and flown before aerodynamics reached its present level of sophistication, so the engineer today must design ejection seats, restraint harnesses, and other hardware on the most rational basis available, even though he lacks full details as to human injury modes. The lack of information does indicate, however, that a great deal of biomedical experimentation is required to extend and refine the data required by engineering personnel.

The multiple injury mode view of the human body can be seen from a review of injury data, such as occurs in accidental free falls. However, particular injury modes tend to occur at different force levels in different individuals and at different times.

Statistical variability exists in the mechanical properties of almost all materials. For example, if a series of samples are cut from 4130 steel ingots and tested in a tensile test machine, a distribution of breaking strengths would be found to occur. In general, the distribution would be found to follow the normal Gaussian-type of curve, although other distributions are possible. Similar results can be expected for biological materials.

Because of the variability of the mechanical characteristics of biological structures, it is desirable to consider the damage to such tissues from a probabilistic standpoint. Instead of assuming that all the material fails at a given level and none of it fails below that level, it is much more in keeping with the real world to assume that a larger and larger percentage of samples tested will fail as the applied load increases. In graphic form, the probability of injury curves for three injury modes might appear as shown in figure 17. Note that the average value where failure occurs, and also the variability in failure loads is different for the three injury modes.

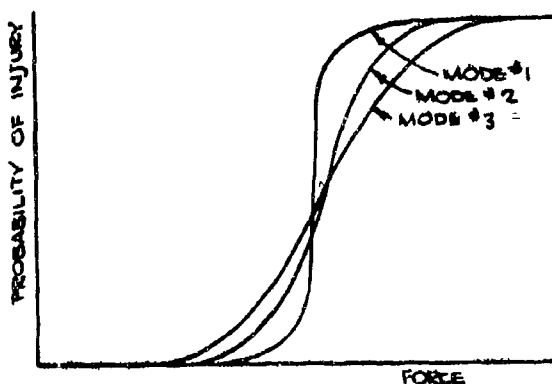


Figure 17. Probability of Injury Curves for Three Injury Modes.

Using the statistical approach to the definition of human response to acceleration, a multiple injury mode dynamic system subjected to a rectangular pulse acceleration would have the "tolerance lines" as shown in figure 18.

One known additional factor should be taken into account in defining the effect of acceleration on the human body. In Section III age has an appreciable effect on the stiffness and strength of bone. Probably age affects all biological tissue to some extent. Unfortunately, there is very little data available on this effect, other than that on hard bone in compression. This is an area that warrants further biomedical research, because the effect is sufficient to begin causing problems when the age of a person subjected to an acceleration is above 35 years. Lacking more knowledge at present, this should perhaps be taken into account by using conservative factors in establishing acceptable acceleration levels, when the application is for designs in which older people will be subjected to the accelerations.

In summary, the three items of principal importance are: (1) a number of injury modes, both primary and secondary, can occur and should be considered; (2) any given injury mode is best represented by probability of injury curves which have further value as indication of the degrees

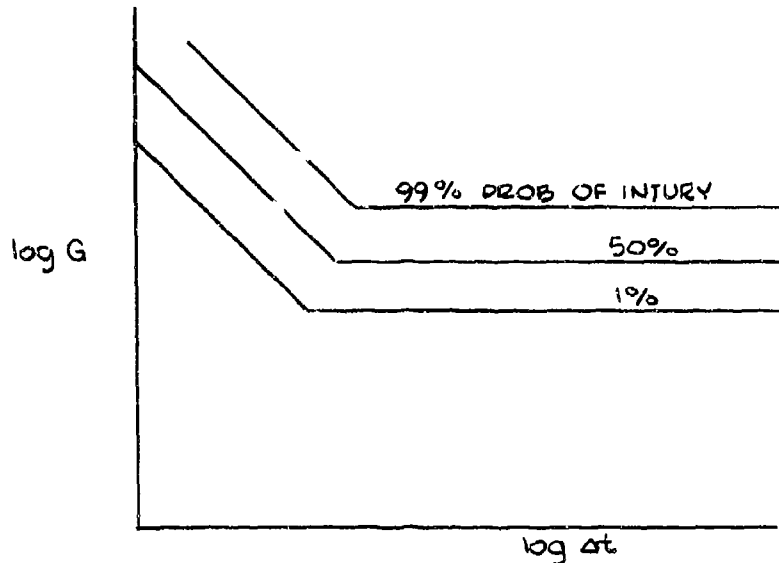


Figure 18. Probability of Injury Tolerance Lines.

of risk in operational situations; (3) the effect of age on strength of bone and tissue is of consequence in acceleration exposure conditions, so that the age of the using population should be considered in selecting acceptable acceleration levels. As the state of the art advances, further experimental and theoretical research will undoubtedly pinpoint other factors worth consideration.

RESTRAINT AND SUPPORT PARAMETERS

The preceding paragraphs have indicated some of the difficulty in describing human tolerance to acceleration as precisely as engineers would like. However, the full picture has not yet been developed, because thus far the human body has been assumed to be rigidly attached to the input acceleration support surface. Such an assumption would be correct only in those cases where a spinal or transverse acceleration is applied to the human body through a seat structure without intervening cushions. In most practical seating systems, cushions or other resilient supports are used to increase comfort and their dynamic characteristics must be taken into account. Furthermore, accelerations applied in negative spinal, negative transverse, and lateral directions result in the body impinging upon relatively elastic devices, such as, lap belts and shoulder harnesses. These devices also act to modify the acceleration transmitted to the human body.

The natural frequency, damping coefficient, and bottoming depth of elastic support devices can have a major effect in amplifying or attenuating input accelerations. Certain rigid supporting structures such as plastic foam or honeycomb materials used as one-shot energy absorbing devices have similar effect on the forces felt by the body. A description of these effects is presented in reference 2.

In general, the effect of restraint device characteristics on the ability of the human body to withstand input accelerations is best handled by analog or digital computer techniques. Certain analytical and general results have been obtained, and such results are usually presented in the format shown in figure 19, in which an amplification-attenuation scale is used.

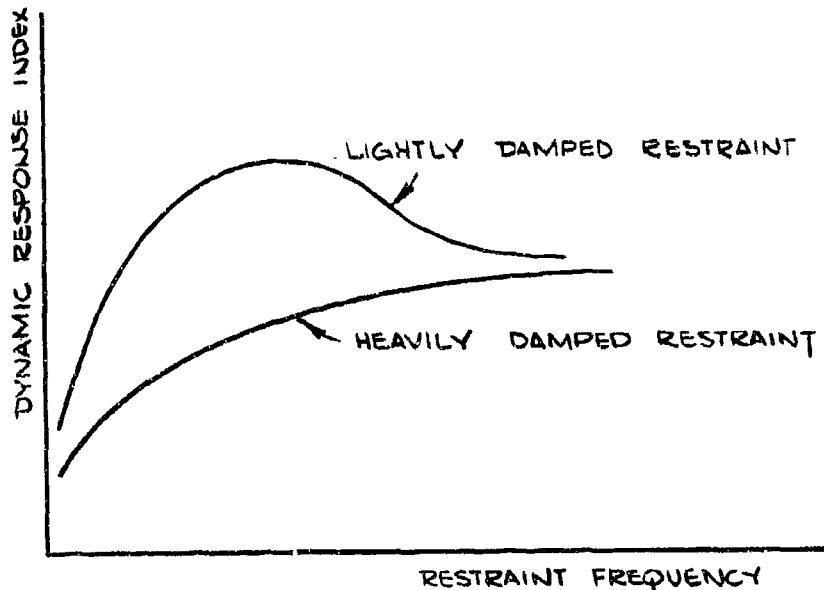


Figure 19. A Typical Restraint Effect Graph Using the Restraint Analyzer.

One method of assessing an arbitrary acceleration input is to approximate the input acceleration by using the duration, peak acceleration, and rise time values, or by using a known mathematical function such as a half sine pulse. By consulting the appropriate charts and graphs in published reports, the peak force in the dynamic model can be estimated. Then, the appropriate charts and graphs on restraint characteristics can be used to estimate the modification of this maximum force in the dynamic model as a result of restraint effects. Finally, the maximum force in the dynamic system can be compared to a probability distribution of critical forces to obtain a probability of injury. In those cases where data is available, the effect of the user population age should be included in the analysis to obtain a more nearly realistic probability of injury.

In order to use the above procedure intelligently, the engineer must understand the factors involved in the estimate, and he must be aware of the assumptions involved in the analysis. In general, it is recommended that the above procedure be used in preliminary design and similar situations where optimization procedures must be effected rather quickly and on broad scale.

A second procedure that can be used to assess an arbitrary acceleration pulse is to use an analog or digital computer. The undamped natural frequency, the damping coefficient or the damping coefficient ratio, the effect of age on the critical force, and the probability distribution of the force required to cause injury should be known for each injury or deflection mode in the system.

Then, the applicable dynamic parameters of the restraint system, which adds an additional degree of freedom to the model, must be estimated in accordance with the best available information. The equations of motion of the resulting two degree of freedom dynamic system are then solved. The probability of injury can then be estimated within the computer, or by use of an appropriate distribution graph.

In order to simplify the process as much as possible, a fixed circuit analog system using the human body dynamics data presented in this report has been developed for the sponsoring agencies. This device, called the Frost Restraint Analyzer, is illustrated in figure 20. The dynamic model circuits in the restraint analyzer are fixed, but the restraint system dynamic characteristics can be varied by means of front panel controls. Selector switches permit the operator to select the various body models and restraint models. The readout on the restraint analyzer is in terms of Dynamic Response Index.

In addition to analog techniques, a digital computer program has been developed for use in restraint optimization problems. A simplified flow chart for this program is shown in figure 21. Both the analog and digital approaches are discussed in detail in reference 2.

The information and procedures presented in the preceding paragraphs of the body of this report are applicable to the general problem of using dynamic models to describe the response of the human body to acceleration forces. For the general reader, the presentation was intended as an introduction to the problems, techniques, and solutions involved in protecting humans from acceleration exposures. In the sections that follow, more detailed information is provided, particularly with reference to quantitative estimates of human body dynamic characteristics.



Figure 20. Frost Restraint Analyzer.

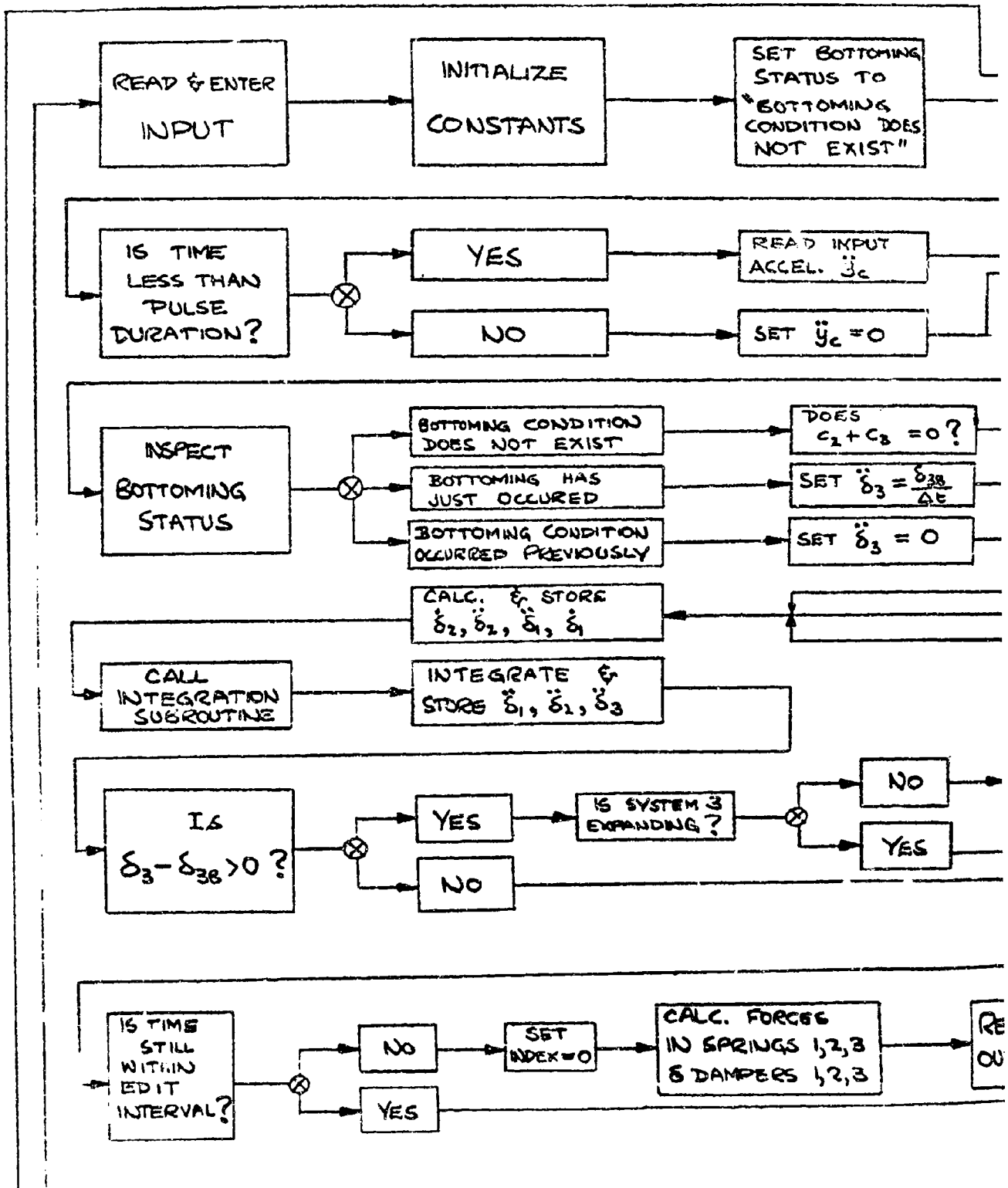
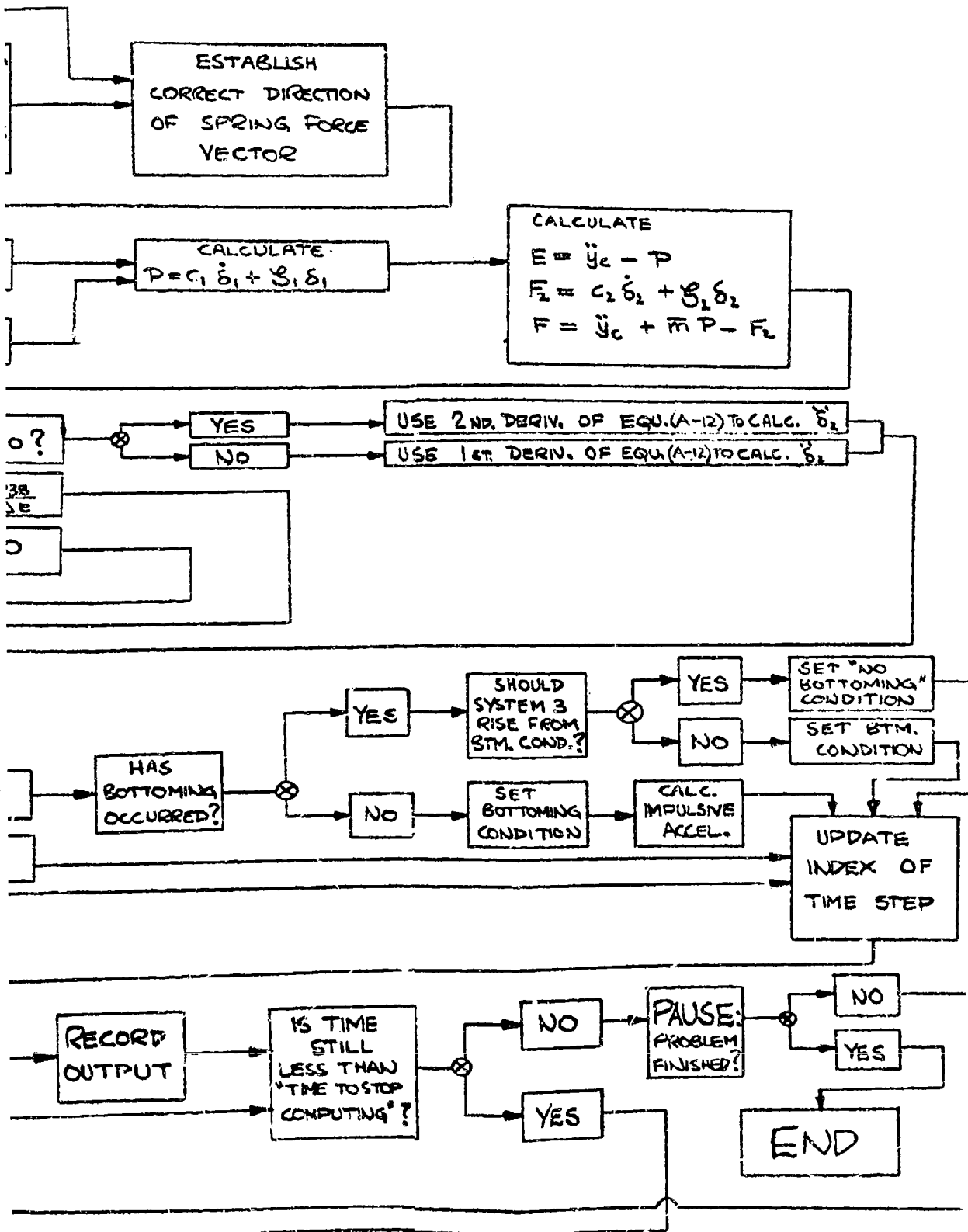


Figure 21. Flow Chart of Digital Computer Program.



B

Section IV

SOURCES OF DATA ON BODY DYNAMICS

Before describing the dynamic models of the human body that have been developed, it is useful to review the available data on acceleration exposures so that the information used as a basis for the models can be evaluated in proper perspective. This section describes in brief form the types of and usefulness of data that are available.

To assess the ability of the human body to withstand acceleration and, subsequently, to establish levels of estimated injury probability, it is necessary to acquire and analyze data on actual exposures to acceleration. In general, there are two sources of data: Controlled experimentation using volunteer subjects and accidental or involuntary exposures. It is, of course, undesirable to injure volunteer subjects in tests situations. Therefore, the data resulting from tests tend to consist of subjective responses or reports of no injury. Accident data, on the other hand, quite often include injury levels ranging from mild to lethal, but the circumstances surrounding the exposure are usually ill-defined.

Tests have been conducted at various laboratories using live human subjects, human cadavers, and live animals. The tests involving live human beings are of primary interest since they provide the most valid and directly applicable data in terms of assessing human response to acceleration.

DROP TESTS

By far the simplest kind of test to conduct and evaluate is a drop test. The principal sources of data on drop tests have been the Civil Aeromedical Research Institute (CARI) at the FAA Aeronautical Center, Oklahoma City, Oklahoma; the Aerospace Medical Research Laboratory (AMRL) at Wright-Patterson Air Force Base, Ohio; and Stanley Aviation Corporation, Denver, Colorado. Both the CARI and AMRL programs involved the investigation of restraint material effectiveness. Stanley Aviation Corporation conducted a series of drop tests using a prototype B-58 escape capsule, as part of the demonstration of operational acceptability of the system, including its shock attenuators and restraint harness.

The results of the CARI tests were reported in references 3 and 4. Detailed subjective comments on the part of the test subjects were recorded for each drop in the program, and these subjective report data have been used in the development of the spinal model reported in Section VI.

As a general rule, subjective comments are known to be extremely variable and often apparently useless. However, if a sufficient number of subjective reports are available, an averaging process can be used to reduce the variability to some extent. In the case of the drop tests conducted at CARI, over 200 tests were conducted, which is a sufficient number to permit averaging.

Subjective comments were collected in both the AMRL and Stanley Aviation drop test programs. However, in the latter program, the subjective comments were not published, nor were the number of drops at each condition sufficient to permit suitable averaging. Also, since the tests involved the use of a prototype operational escape system, there are the additional complicating factors of ill-defined shock attenuator characteristics, unknown impact surface characteristics, and the accelerations imposed were multidirectional.

Drop tests conducted at AMRL facilities at Wright-Patterson Air Force Base have tended to consist of evaluations of proposed operational restraint devices. Relatively few drops have been

conducted under any one condition, and it is difficult to obtain averaged subjective comments in such a case.

SLED TESTS

Major problems exist in assessing the results of short-term acceleration sled tests. First, the applied acceleration is usually of a somewhat irregular shape and, in some cases, it is difficult to establish the true shape of the pulse because of noise in the record. The second complicating factor is the presence of viscoelastic restraint elements in the test. This should not (and does not) prevent the analysis of the data from the test, but the data reduction process is made much more complicated since the dynamics of the restraint system must be taken into account in assessing the forces felt by the subject's body.

The real problem in analyzing sled test data or any other data involving an irregular and arbitrary input acceleration results from an interesting dilemma. Much of the sled test data are intended for use in establishing human tolerance to a given acceleration condition. However, the data cannot be reduced to valid usefulness for this purpose unless a dynamic model of the body is available. Fortunately, there are ways to get around this difficulty, as will be shown.

The major effort in testing live human subjects on sleds has been conducted at Edwards Air Force Base and Holloman Air Force Base. The results are recorded in references 5, 6, 7, and 8. Initial investigations were conducted under the direction of Colonel John P. Stapp and involved considerable risk to the volunteer subjects, including Colonel Stapp. The information resulting from these sled tests has been of great value in preliminary definitions of human tolerance to transverse accelerations, and still remains the only data available on transverse acceleration exposures of sufficient duration to be classed as short-term.

VIBRATION TESTS

If a repetitive input force is applied to the human being, then human tolerance to vibration is the controlling factor. Vibration testing also permits determination of the resonant frequencies and damping of the human body. The principal work in ascertaining human response to vibration inputs has been conducted at the Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio (refs 9, 10, 11, 12). Latham (ref 25) and others have also studied human tolerance to vibration. One of several summaries of the available information has been given by Goldman and Von Gierke in reference 26.

In general, three methods are used in vibration testing. The first is voluntary human exposure to vibrations of various frequencies with amplitude increasing up to a subjective limit; the second is by measurement of input impedance of the human body; the third is measurement of amplitude transmissibility. It will be seen later that resonant frequency measurements using impedance methods and subjective reports provide an excellent cross check on other calculations of the frequency of the human body in the spinal direction. In addition, impedance and transmissibility data permit estimates of damping to be made.

STRUCTURAL TESTS

In addition to tests on live human being, one of the more useful sources of data is the testing of human cadavers. Tests on the mechanical characteristics of the body's hard structure have contributed significantly to an understanding of the mechanical behavior of the body under certain conditions. The problem of tolerance to accelerations in the spinal direction while seated has been of major importance since the advent of the ejection seat, and as a result a number of studies have been performed on vertebrae obtained from human cadavers.

The most useful studies of the mechanical properties of the vertebra are contained in references 13, 14, and 15. Ruff, in reference 13, measured the strength and deflection of vertebra. Static loading of sections of the vertebral column were used to obtain stress-strain curves. Failure or injury was defined as the first peak in the stress-strain curve, which is the yield point in the curve and corresponds to permanent or irreversible deformation. Subsequently, Perey (ref 14) reported on a series of studies in which end plate fracture was shown to occur at a significantly lower load than the vertebral body deformation reported by Ruff. Perey also reported the proportional limit load for a series of cadaver vertebra.

In addition to the strength characteristics of vertebrae, Perey studies the stiffness of vertebrae for several age groups. Yorra, in reference 15, presents typical stress-strain curves for the equivalent of one lumbar vertebra plus the intervertebral disc. This information can be used to estimate the natural frequency of the human body, and the effect of age on frequency. Furthermore, the degree of nonlinearity in the spinal direction can be estimated. The method used in calculating the natural frequency and the results of the calculations are presented in Section V for the dynamic model of the human body in the seated position and spinal direction.

Aside from the work on the mechanical characteristics of the vertebral column, very little research has been performed to obtain data on stiffness or damping of major body structures. Some studies sponsored by the United States Air Force are under way at present on tissue strength, but because of the inherent difficulties in mechanical measurements on biological tissue and the relatively more complex nature of thoracic and abdominal organs and their support systems, it can be anticipated that useful data will not be obtained as readily as was possible with the spinal column.

Recent studies at Wayne State University using seated human cadavers (ref 16) are of considerable interest since they illustrate the capability of predicting dynamic response by use of dynamic models and then obtaining confirmation in an experimental situation. Previous work at Wayne State, particularly in the area of head impact, has resulted in the definition of several important dynamic parameters of the skull-brain system (ref 17).

ANIMAL TESTS

There are certain objections to the use of data from cadavers, particularly since it is possible that storage and preservation techniques can influence the mechanical properties of biological tissue. For certain mechanical characteristics, it is possible to cross check information obtained from cadavers with experiments on live human beings. Another approach, of course, is to use live biological specimens other than humans. A variety of animals have been used in acceleration experiments, including hogs, rhesus monkeys, chimpanzees, mice, and bears. The advantage of using animal subjects is that injurious and even lethal exposures can be attempted providing the experimentation is conducted within the rules for animal experimentation. Thus, the health and well being of human subjects need not be risked in severe condition testing.

The information obtained from animal experimentation must be viewed with some skepticism if it is to be applied to the problem of assessing and setting human tolerance limits, however. The forces generated in an animal body will probably not be the same as those generated in the human body for an equivalent input acceleration, because of obvious dynamic differences. Also, the force required to cause injury in the skeletal structure, organs, or other portions of the animal body may not be the same as the forces required to cause injury in the human body. Before valid extrapolations can be made from animal experimentation to human conditions, the precise relationship between the dynamics and critical force in the animal's body and those in the human body must be

known. Unfortunately, this type of definition has not yet been accomplished for any animal subject type.

Animal testing does have benefits in certain situations, however. As an example, the experiments conducted by Kornhauser (ref 1) cited earlier in this report resulted in reasonable confirmation of the shape of the tolerance curve predicted by a simple dynamic model. Thus, Kornhauser was able to show that a system consisting of biological tissue was equivalent to a mechanical system insofar as tolerance to acceleration was concerned.

ACCIDENT DATA

Up to this point, the discussion of acceleration exposure data has been restricted to information collected under test conditions in which the investigator presumably is able to control many of the conditions under which the acceleration is received. Such data is of great value since the parameters affecting the outcome of the exposure are either known or controlled. Human subjects cannot be exposed to large acceleration levels in order to determine the magnitude of the forces involved in injury, however. Cadaver or animal tests can result in a definition of injury levels or modes but extrapolation of the data to live human beings is fraught with difficulties.

One way in which information can be obtained on injury-producing acceleration exposures to obtain data relating to situations in which human beings have been accidentally exposed to very large forces. By far the simplest data to collect and analyze is that resulting from situations in which live human beings have fallen from known heights and struck solid surfaces, or where the drop height and deceleration distance is known so that the acceleration and duration can be estimated. The classic study in this regard is that by DeHaven (ref 18) in which survival of free falls was documented. More recently, an analysis of free falls occurring in mountain climbing has been performed by one of the authors of this report and the results are presented in Section V. DeHaven collected cases that consisted essentially of miraculous survivals, which results in a somewhat overly optimistic estimate of tolerance to this kind of acceleration, but perhaps can serve to define absolute limits.

There exists very little adequately documented accident data other than the very spectacular information on free falls. Obviously, the many cases in which people slip on ice, fall in the home, and otherwise injure themselves accidentally during the course of everyday living have not been well documented or recorded, although a program to collect data is now under way at the Civil Aeromedical Research Institute in Oklahoma City (ref 19).

EJECTION SEAT DATA

There does exist one other major source of exposure to large accelerations which can provide information for assessing acceleration tolerance. Ejection seats have been in use in military aircraft since World War II, and in some cases, the ejections have involved acceleration forces sufficient to cause an appreciable number of injuries to pilots and crew members who eject from disabled aircraft. An excellent study of ejection injury is contained in ref 20, which summarizes the medical record involved in the use of ejection seats by the Royal Air Force and Royal Navy. The US Navy and US Air Force also have data relating injury to type of ejection seat. Unfortunately, all of the ejection seat data constitute a very complex data reduction problem.

The initial spinal forces generated by an ejection seat are a function of a ballistic catapult or a rocket catapult that is subject to the usual statistical variations in propellant performance plus variations caused by temperature. The weight of the ejectee also influences the rise time and peak acceleration achieved by the seat. Added to these uncertainties are the factors of seat cushion

thickness, material, and age, restraint harness configuration and material, restraint harness pre-load or slack, and the age and previous loading history of the restraint elements.

Since the nominal force-time history of an ejection seat is rarely known with any precision and the dynamic characteristics of the restraint and support devices are totally unknown, the ejection seat data that are currently available do not constitute a very promising source of information, unless suitable data reduction techniques are developed to permit a reasonable assessment of the available information. This could then be used as a cross check against data developed from other sources, such as tests with live human beings and with cadavers.

Section V

A DYNAMIC MODEL OF THE HUMAN BODY IN THE SPINAL DIRECTION

A dynamic model for the spinal injury mode of the human body is not complicated, for the vertebral column can be considered the resilient load-bearing member, and the mass of the upper body can be considered to be resting on top of the column. If this simple mechanical analogy to the body is valid, then it ought to be possible to calculate the natural frequency by obtaining information on the stiffness of the vertebral column and the magnitude of the mass resting on it. The latter data should not be difficult to obtain, and since static load-deflection tests have been conducted on vertebrae, it should be possible to obtain stiffness estimates.

ESTIMATES OF MASS AND SPRING STIFFNESS

Yorra (ref 15) obtained load-deflection curves for several lumbar vertebrae. The average stiffness obtained for the fourth lumbar (L4) vertebra was 11,750 pound/inch. Unfortunately, the load-deflection curves for other vertebrae were not obtained.

From a knowledge of materials in general the stiffness of any given structural element may be reasonably assumed to be directly related to the breaking strength of that element. Therefore, it is possible to obtain the stiffness of all the vertebrae if their breaking strength is known.

The most valid data on vertebral breaking strength in compression is that obtained by Ruff (ref 13) in Germany during World War II. He measured the breaking strength of individual vertebrae between the fifth lumbar (L5) and eight thoracic (T8) with a minimum of three samples for each vertebra tested. In addition, Ruff devised a method for estimating the percentage of total body weight carried by each vertebra from T5 through L5. Table I presents the raw data

TABLE I
RUFF'S VERTEBRAL STRENGTH DATA

Vertebra	Raw Data		Average in lbs.
	Breaking Strength in Kg.	Average	
T8	640, 540, 609	590	1315
T9	610, 720, 700	677	1493
T10	800, 660, 770, 730	740	1632
T11	750, 720, 860, 755	771	1700
T12	900, 690, 800, 800	797	1757
L1	720, 840, 900, 800, 800	812	1790
L2	990, 800, 830	873	1925
L3	900, 940, 1100	980	2161
L4	1100, 900, 950	983	2168
L5	1020, 1000, 1200	1073	2366

thus obtained on the breaking strengths. The percentage of body weight data and the breaking strength in pounds is shown in columns 2 and 4 respectively of table II.

It is possible to extrapolate the percentage of body weight carried from T5 up through T1 if there is a relatively consistent increment of 3% from vertebra to vertebra in the portion of the spinal column actually tested by Ruff. The resulting value of 9% of body weight for T1 is reasonable since this is the approximate weight, in terms of percentage, for the head and neck. Thus, the percentage of body weight carried by each vertebra from T1 through L5 can be listed. It is then an easy matter to calculate the actual weight in pounds carried by each vertebra for a 160 lb man as shown in column 3 of table II.

The actual weight carried by each vertebra and the compressive breaking strength of each vertebra is known for T8 through L5. Therefore, the breaking load (expressed as G-force/static weight) can be calculated quite easily and is shown in column 5 of table II. If it is assumed that the breaking load of vertebra from T1 through T7 is relatively constant at 25 G, then the breaking strength in pounds for these vertebra can be calculated since the actual weight carried in pounds is listed in column 3. The results are shown for T1 through T7 in column 4 of table II.

TABLE II
CALCULATION OF VERTEBRAL STIFFNESS

Vertebrae	% of Body Weight Carried	Weight Carried 160 lb Man	Breaking Strength in lbs.	Breaking Load in G	% of L4 Breaking Strength
T1	9	14.4	360	25.0	16.6
T2	12	19.2	480	25.0	22.1
T3	15	24.0	600	25.0	27.7
T4	18	28.8	720	25.0	33.2
T5	21*	33.6	840	25.0	38.7
T6	25*	40.0	1000	25.0	46.1
T7	29*	46.4	1160	25.0	53.5
T8	33*	52.8	1315*	24.9	60.7
T9	37*	59.2	1493*	25.2	68.9
T10	40*	64.0	1632*	25.5	75.3
T11	44*	70.4	1700*	24.2	78.4
T12	47*	75.2	1757*	23.4	81.0
L1	50*	80.0	1790*	22.4	82.6
L2	53*	84.8	1925*	22.7	88.8
L3	56*	89.6	2161*	24.1	99.6
L4	58*	92.8	2168*	23.4	100.0
L5	60*	96.0	2366*	24.6	109.1

*Single asterisk represents data collected experimentally by Ruff; unmarked values are calculated or assumed as explained in the text.

The procedure outlined above is used simply to obtain the breaking strength in pounds for each vertebra from T1 through L5. Column 6 of table II presents the breaking strength of each element in the vertebral column in terms of the percentage of the breaking strength of vertebra L4. Since the stiffness of vertebra L4 was obtained by Yorra as noted previously, it is then possible to esti-

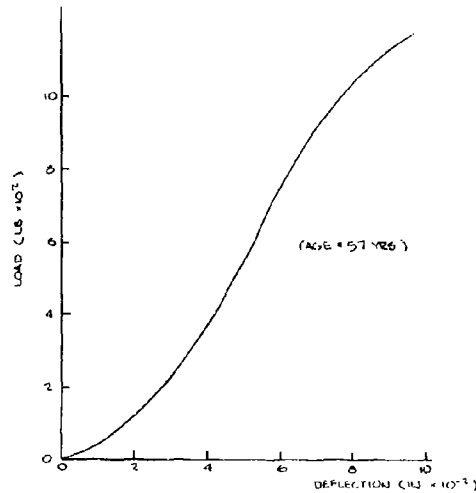


Figure 22. Load-Deflection Curve for L4 Vertebra.

mate the stiffnesses of the other vertebrae on the basis of the breaking strength percentages. Having established the relative breaking strength of the vertebrae in a spinal column, it is necessary to estimate the stiffness of vertebra L4 from the Yorra data, reproduced in figure 22.

A nonlinear spring has two kinds of equivalent linear stiffness. The first is the tangent to the load-deflection curve. This is the apparent stiffness of the spinal column that would be observed in small amplitude vibration tests on a shake table.

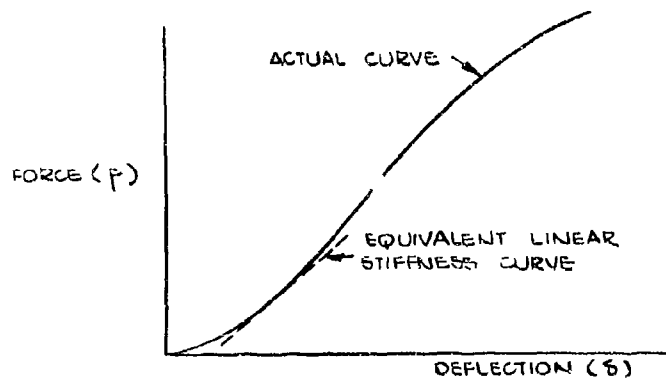


Figure 23. Estimate of Small Amplitude Equivalent Linear Stiffness Using Tangent Method.

The second type of stiffness is illustrated in figure 24. The straight line approximation to the nonlinear curve is drawn so that the areas between the straight line and the curved line are equal up to the load or deflection value of interest. This estimate of the stiffness is the one that would be observed in a test where large impact forces were applied to the body and it was then permitted to oscillate freely. For this reason, this estimate of stiffness is termed the free oscillation stiffness.

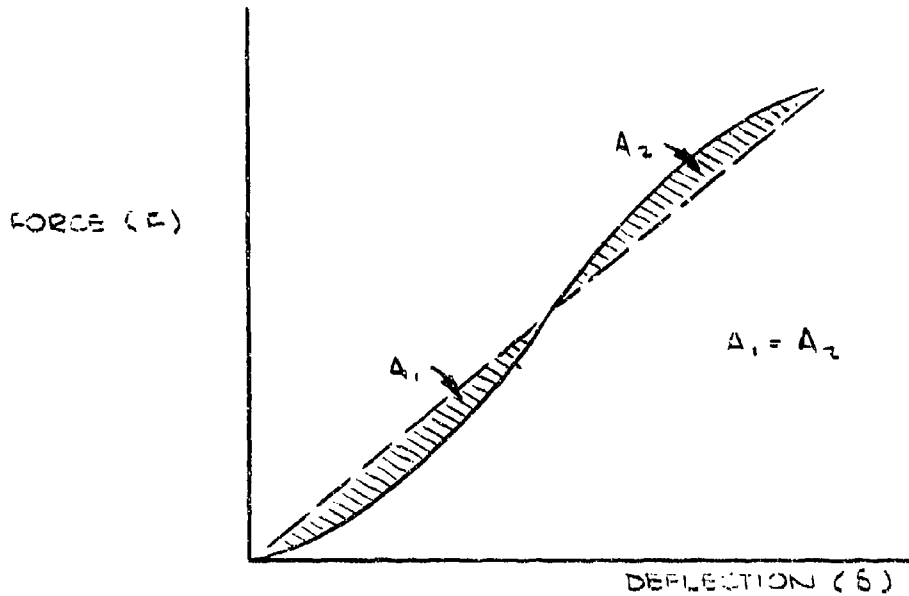


Figure 24. Estimate of Free Oscillation Equivalent Linear Stiffness Using Equal Area Method.

To recapitulate, the stiffness of all the vertebra in the spinal column can be estimated from the stiffness of vertebra L4, and the relative compressive breaking strength of all the elements can be approximated by using both the free oscillation and small amplitude methods of obtaining equivalent linear stiffnesses. With this data in hand, it is necessary to obtain an estimate of the weight of the body segments resting on the spinal column. References 21 and 22 provide the estimates for the weights of various body segments including organs in the thoracic and abdominal cavities.

For the purposes of this analysis the mass resting on the spinal column was assumed to consist of the neck, head, upper arms, lower arms, hands, and the stiff structure of the thorax. From available data the organs in the thorax were estimated to weigh approximately 3.41 pounds and the blood circulating through the thorax at any given time weighs approximately 4.06 pounds. The total upper body weight, including organs and blood, was estimated at 67.52 pounds from the available anthropometric data. Subtracting the weight of the organs and blood in the thoracic cavity* resulted in an estimate of 60.5 pounds for the relatively rigid mass of the upper body. These data are shown in Table III.

*Coermann (ref 9, 10) and others have shown that the fundamental natural frequency of the Viscera is about 3 cps. We should therefore expect thoracic organs to be relatively seismic at the higher spinal deflection frequency.

CALCULATION OF NATURAL FREQUENCY

The equation for the undamped natural frequency of a simple spring-mass system is

$$f_n = \frac{1}{2\pi} \sqrt{k/m}$$

where f_n = undamped natural frequency in cps

k = stiffness of spring in lbs/ft

m = mass in slugs (weight/32.2)

TABLE III
BODY SEGMENT WEIGHTS

	<i>Total Upper Body Weight (lbs)</i>	<i>Less Viscera (lbs)</i>
Head and Neck	13.20	13.20
Trunk		
	67.52	60.05
Structure	19.45	19.45
Organs	3.41	
Blood	4.06	
Upper Arms	18.90	18.90
Lower Arms	6.20	6.20
Hands	2.30	2.30

As noted, the weight of the upper body for a 50 percentile man was estimated to be 60 lbs, which results in a mass of 1.865 slugs. Using this value of 1.865 slugs and the various stiffness estimates from the load-deflection curve shown in figure 22, the effect of vertebral nonlinearity on the free oscillation frequency and small amplitude frequency are obtain. These data are shown in figure 25.

Three resonant frequencies in the range of 1 to 50 cycles per second have been observed on live human subjects as indicated in reference 9 for example. Of immediate interest is the resonance observed in the vicinity of 6 to 7 cps. When corrected to obtain the undamped natural frequency, a value of 6.1 cps is obtained. Assuming equivalent conditions for the cadaver data just analyzed, the predicted frequency is 5.7 cps. This is a rather significant variation, over 5%, so that the reason for the difference should be investigated.

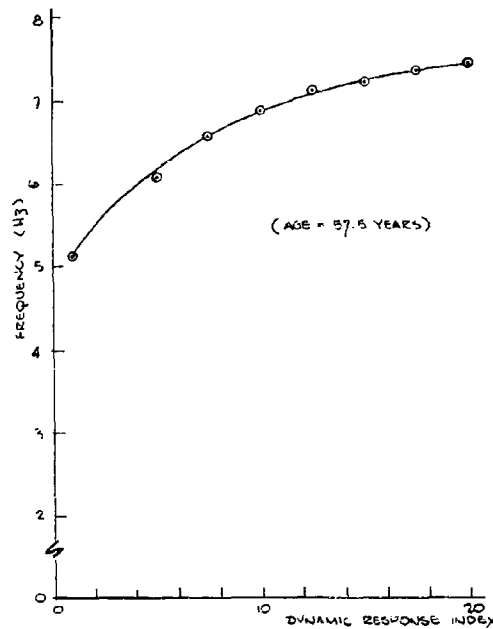


Figure 25. Effect of Vertebral Nonlinearity on "Free Oscillation" Frequency.

The cadaver used by Yorra to obtain the load-deflection curve used in this study was 57.5 years old at death. The subject used to obtain the estimate of 6.1 cps was 47 years old at the time of the test reported. Therefore, one of the variations between the two data points is age, and it is appropriate to investigate its effect on the stiffness of the vertebral column.

THE EFFECT OF AGE ON SPINAL CHARACTERISTICS

Perey in ref 14 provides a large amount of data on vertebral end plate breaking strength versus age as obtained from static tests of wet, fresh cadaver vertebrae. In addition, Perey reports the proportional limit strength for the vertebrae in two different age groups.

In figure 26, Perey's data for end plate fracture is plotted as the end plate breaking strength in psi vs age. Since Perey also measured the area of the end plate upon which loads impinge, it is possible to convert these data to ultimate stress. This has been done and the results are shown in figure 27, together with the proportional limit strength of the vertebra. When both lines are extrapolated they meet at the zero breaking strength value, which indicates some degree of consistency between two different sets of data. In addition to the Perey data, a single estimate of permanent set for the lumbar vertebra is shown in figure 27. As would be expected, this permanent set value is above the proportional limit value.

Not only did Perey obtain proportional limit strength values for lumbar vertebrae, but he also obtained the stiffness for each of the vertebra tested. In ref 14, the data are listed under three age groups: Less than 50 years, 51-60 years, and 61 plus years. A plot of breaking strength against stiffness is shown in figure 28, where the less than 50 years old group and 51 to 60 year old group

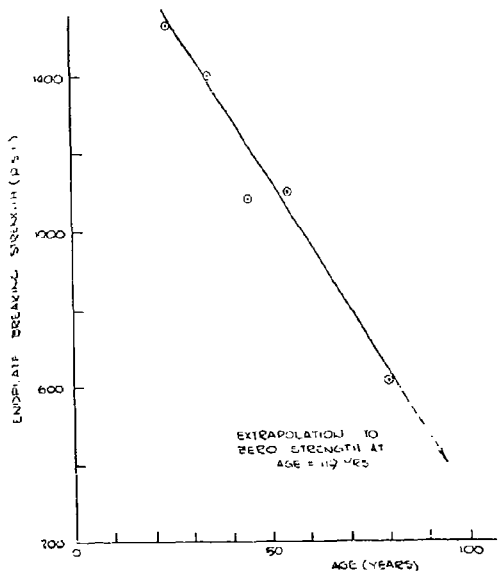


Figure 26. Variation in Cadaver Vertebral End Plate Breaking Strength with Age.

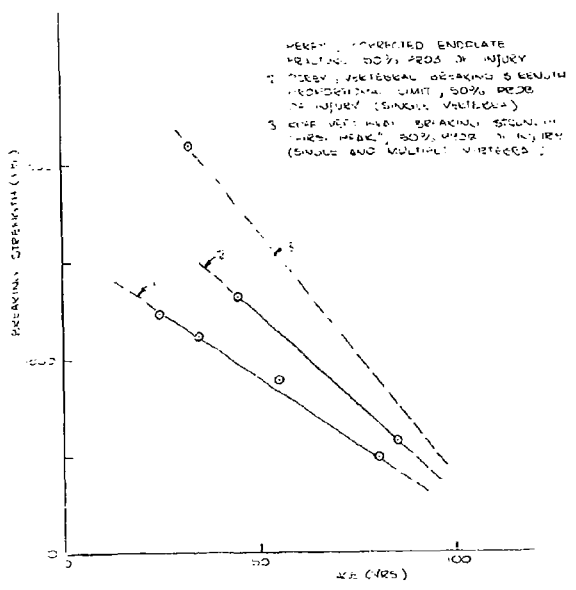


Figure 27. Effect of Age on Vertebral Breaking Strength.

overlap to some extent. However, the over 60 years group is definitely located in the low stiffness and low breaking strength levels.

A statistical correlation (known as a Pearson product moment) was calculated on the strength-stiffness data in order to confirm the fact that the two variables were indeed closely related. The result of the calculation was a correlation coefficient of 0.82. A coefficient of 1.00 would indicate

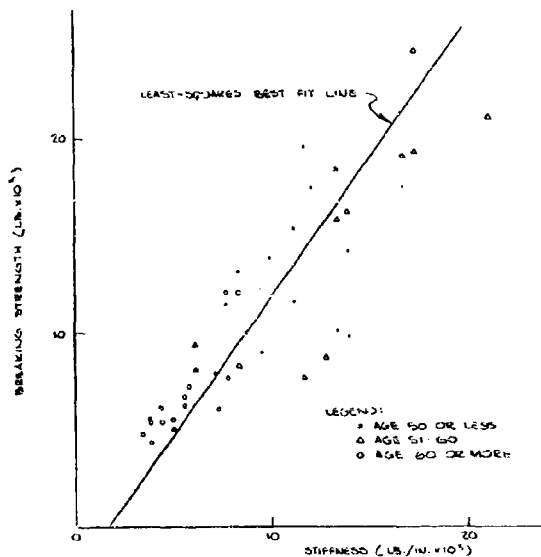


Figure 28. Relationship Between Breaking Strength and Stiffness From Fresh Cadaver Vertebrae.

a perfect relationship, and a value of 0.00 would indicate pure randomness in the data with no discernible interrelationship between variables. The value of 0.82 obtained in this case can be considered reasonable validation of the relationship between strength and stiffness.

In order to define the exact variation in stiffness with breaking strength, a best fit line was calculated from the data by the least-squares method. The result is a line with the following equation: $k = 1.672 + 0.726\delta_B$ where k is equal to stiffness in pounds per inch and δ_B is equal to the breaking strength in pounds. This is the line plotted in figure 28.

The surprising aspect of these data is the indication of bones becoming less stiff with age rather than more brittle. Bone consists of a mineral called apatite dispersed through a matrix of the protein collagen and is termed a two-phase material. Apatite is characterized by high stiffness and compressive strength and a rather low tensile strength. Collagen has a low stiffness but high tensile strength. Two materials combine to form a composite material having good elastic characteristics and high compressive and tensile strength, the combination being similar in some respects to such materials as prestressed concrete or fiberglass. Presumably a change in relative composition occurs with age and accounts for the phenomenon described.

Aging, then, apparently results in bones becoming less stiff. The effect on the natural frequency of the human body can be calculated from the relationship shown in figure 28. It is only

necessary to obtain a set of correction factors so that the stiffness values obtained from the 57.5 year old cadaver reported by Yorra can be modified for other ages. Since an approximate relationship between breaking strength and stiffness, and breaking strength and age has been devised, the effect of age on stiffness can now be estimated. The result of this evaluation of age on frequency is shown in figure 29.

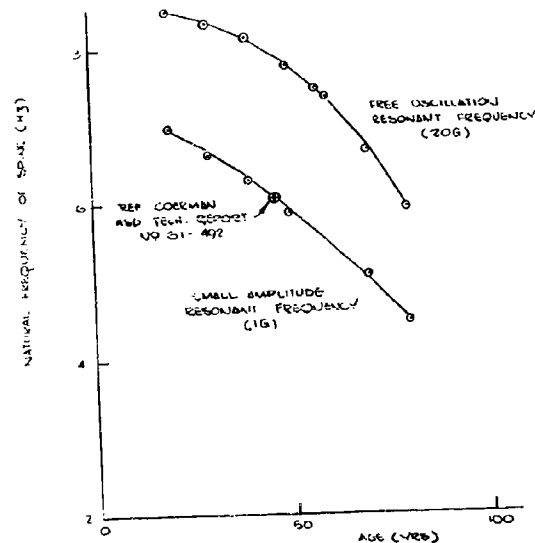


Figure 29. Variation in Human Spinal Natural Frequency With Age.

The predicted frequency for a 47 year old man (fig 25) now becomes 6.08 cps compared to the 6.1 cps observed in the impedance tests as reported in ref 9. In addition, an estimate of resonant frequency for a group with an average age of 28.5 years is reported in ref 11. The estimate obtained from subjective comments is 6.6 cps and the value predicted from the cadaver data is 6.75 cps. In both cases, the agreement is good, particularly considering that the analysis of the cadaver data cannot be considered to be extremely precise.

At this point, it is necessary to review the implications of the preceding discussion. Using a very simple dynamic analogy to the human body, the natural frequency of the body has been calculated from cadaver data. The agreement between the calculated values and the values obtained in vibration tests is very good, particularly when corrected for age and nonlinearities. However, the undamped natural frequency of the dynamic system is only part of the overall picture. It is also necessary to obtain an estimate of the damping in the human body in the spinal direction in order to obtain valid estimates of response to input accelerations.

ESTIMATES OF DAMPING

Impedance measurements with the human body reported by Coermann in ref 9 enable preliminary estimates of the damping coefficient \bar{c} to be made for relatively small amplitude oscillations, for frequencies up to 20 cps. The values derived by Coermann are not as accurate as they

could be, however, because he used an approximate equation for impedance. In addition, he used only the peak impedance to determine the coefficient \bar{c} , whereas the ratio of the resonant frequency to the maximum impedance frequency can also be used to obtain a second reading. For one subject, the data can be summarized as shown in Table IV assuming a subject mass of 5.75 slugs. For practical purposes, the variations between sitting erect and relaxed may be neglected. Thus, the final figure become $\bar{c}=0.31$.

TABLE IV
Variation of Damping Coefficient and Natural Frequency
with Body Position

$\bar{c} = 0.31$ in the sitting position
 $\bar{c} = 0.35$ in the standing position
 $f_n = 6.1$ cps sitting erect
 $f_n = 5.67$ cps standing erect

Using this value, the effect of the nonlinearities in the spring can be estimated. For practical purposes, the free oscillation case can be considered sinusoidal, although the frequency naturally varies with the amplitude achieved. Thus, if we assume linear damping characteristics, on the assumption that the materials involved are essentially viscoelastic, the damping coefficient will vary directly with the apparent linear frequency.

Table V shows the values calculated for various ages and various peak DRI values in the spinal column, and the results are shown in figure 30.

TABLE V
Variation of Natural Frequency and Damping Coefficient
with Age and Peak DRI

Peak DRI	1	2	5	10	15	20	
$\omega_t/2\pi$	5.15	5.48	6.14	6.9	7.3	7.47	(cps) (age 57.5)
"	5.06	5.38	6.03	6.78	7.17	7.34	" (age 60)
"	5.64	6.0	6.71	7.55	7.98	8.17	" (age 40)
"	5.86	6.23	6.99	7.85	8.3	8.5	" (age 20)
"	5.81	6.18	6.91	7.78	8.23	8.42	" (age 27.9)
\bar{c}	.374	.352	.314	.279	.264	.258	(age 60)
"	.336	.315	.282	.2505	.237	.2315	(age 40)
"	.323	.3035	.271	.241	.228	.2225	(age 20)
"	3.25	.306	.274	.243	.23	.2245	(age 27.9)

In this and all subsequent calculations, the age of 27.9 years appears since this is the age reported by Hertzberg in ref 23 to be the average age of the Air Force flying population. This value is based upon data taken in 1950, so that this average age may have shifted to some extent in the intervening years.

Using Ruff's data (ref 13) as a basis for estimating the force at which permanent damage occurs to the vertebral column, it is now possible to define the complete dynamic model of the human body in the spinal direction. Ruff's data indicates that there is a 50% probability of permanent damage at a steady-state value of 20.2 G and at an age of 32.5 years. The results are presented in table VI and in figure 31.

TABLE VI
DYNAMIC MODEL OF THE HUMAN BODY-POSITIVE
SPINAL DIRECTION

Single Degree of Freedom, Linear Model

Age (Years)	20	27.9*	40	60
Natural Frequency $\omega/2\pi$ (cps)	8.55	8.45	8.13	7.08
Damping Ratio $(\bar{c})^{**}$.228	.230	.237	.264
50% Injury Levels:				
Steady State -- \ddot{Y} crit (G)	23.05	21.30	18.45	13.78
Impulsive Velocity Change***-- ΔV (fps)	21.00	19.75	18.12	16.78

*Average age of USAF flying population.

**At a peak DRI of 15 G.

***Lower ΔV values may cause injury due to shock wave effects in cerebrospinal fluid or in viscera.

HEAD INVOLVEMENT IN SPINAL TOLERANCE

In addition to the data already presented, figure 31 shows a somewhat lower tolerance level below 0.01 seconds due to involvement of the head. This value was obtained from a series of drop tests conducted by Swearingen using live human subjects. These studies are reported in ref 3 and ref 4, and the subjective comment data from this test series was made available to the authors. Of immediate interest is the fact that it was possible to use the subjective comments to obtain an estimate of the average subjective end point in an impact situation.

The tests involved human subjects dropped from various heights on a vertical test rig. Both the seated and standing positions were investigated, although this review is restricted to the seated position. All of the testing involved decelerations parallel to the subject's spine except for a few special cases which are not considered here.

Most of the tests were conducted with a hydraulic snubbing device in the base of the test rig. Total available stroke on the snubber was 1 inch. A few tests were conducted with a rigid seat and base, that is, with the snubber locked out. These are of particular interest here, since they constitute ultrashort period impacts in which velocity change governs the response.

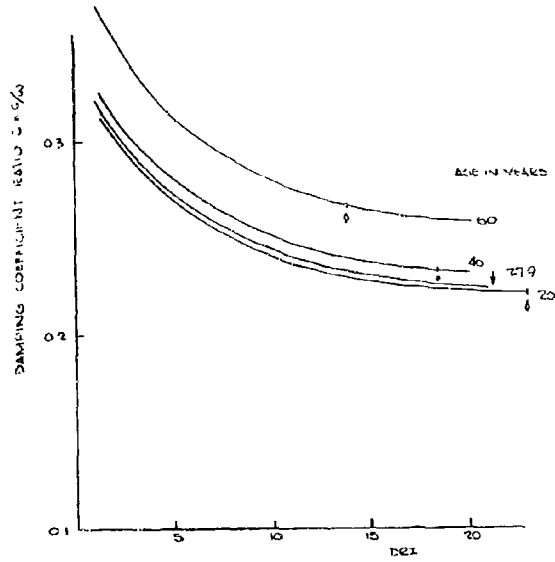


Figure 30. Variation of Free-Oscillation Damping Ratio With Age and Peak DRI.

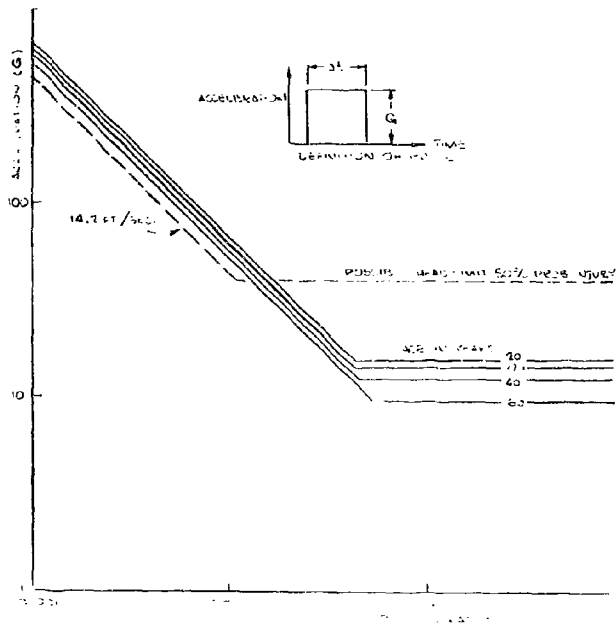


Figure 31. Spinal Positive Injury Curve for 50% Probability of a Compressive Fracture Due to a Rectangular Acceleration Pulse.

The method for using the subjective reports in the drop test situation involved the development of a subjective severity scale. All of the subjective reports were scanned in order to become familiar with the terminology used by the subjects, after which the following scale was established

Level 1	No subjective report
Level 2	Slight jolt or shock or feeling of discomfort
Level 3	Definite jolt or shock, mild ache, severe discomfort
Level 4	Severe jolt or shock, definite ache, slight and brief pain
Level 5	Definite pain

Note that the term "shock" is used in a lay sense and not in a medical sense. This field of multipliers was then established as follows:

- Multiplier of one — Symptom restricted to one portion of the body (other than the head).
- Multiplier of two — Symptom was reported for several portions of the body or for the head alone.
- Multiplier of three — Symptom reported for entire body or for several portions of the body plus the head.

By using the multipliers, each level of severity was weighed on two bases, namely, the extent of the subjective feeling and the involvement of the head. Each acceleration exposure was rated using the scales and multipliers just described. However, of major importance is the condition in which a rigid seat was dropped onto a rigid base.

The average subjective end point from this test series is equivalent to an impact velocity of 11 feet per second and the maximum recorded severity using the subjective severity scale was equivalent to a velocity change of 14 feet per second. See figure 32. The age of the subjects used in this test series range from 25 to 40 with an average in the 30-35 year range. Using the dynamic model previously developed, the critical velocity change in an impact test should have been between 18 and 20 feet per second.

An explanation for the variation in the two sets of data can be found in the subjective reports. In many cases, the impact test results indicated involvement of the head or the abdominal or thoracic viscera. We hypothesized that very short rise time acceleration inputs may result in the generation of shock waves in media such as the spinal column, the cerebrospinal fluid or in the essentially liquid viscera.

Unfortunately, the generation of shock waves is principally a matter of input acceleration rise time so that further tests at various rise times would be required to define fully the mechanisms involved. At any rate, it should always be kept in mind that head and viscera involvement may occur in such short rise time acceleration pulses. At present, this does not appear to be of practical significance, since most real acceleration pulses have reasonably long rise times. However, this could present a problem in a future application.

CONCLUSIONS

The dynamic model of the human body that has just been described is felt to be applicable to those situations in which the acceleration is applied in such a manner that the body is pushed down into the seat. This is the most common and critical mode of loading the body in that ejection seats, escape capsules, and landing impacts apply their principal accelerations in this direction. It is possible, of course, to apply an acceleration in the opposite direction so that the forces tend

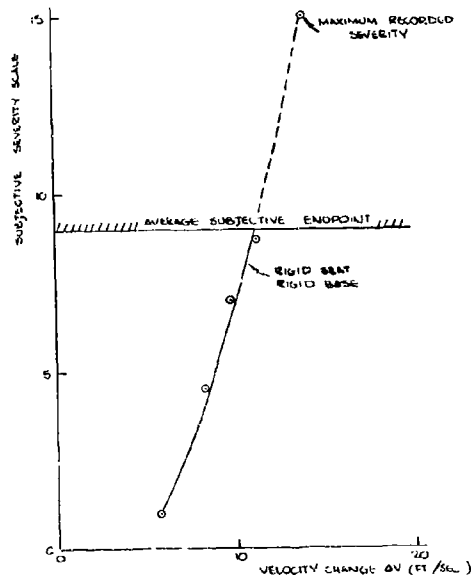


Figure 32. Variation in Subjective Severity with Velocity Change for Four Test Conditions in FAA Tests.

to lift the body out of the seat. This is frequently called negative acceleration or negative spinal acceleration. The model just described for the positive spinal direction is not directly applicable to the negative spinal direction.

There are two major differences between the positive spinal and negative spinal acceleration cases. In the first place, the human spinal column is arranged in such a manner that the strength of the individual vertebrae increase for those lower in the spinal column. This is a sort of natural adaptation of the bone material since the lower vertebrae are subjected to larger loads caused by the obvious fact that a greater percentage of the total body is above the lower vertebrae. The relationship between the load applied to the individual vertebra and position in the spinal column and breaking strength and position in the spinal column is shown in figure 33.

Now, if the loading is reversed, the largest loads will be applied to the weakest vertebrae. For those vertebrae upon which data is available, the relationship is that as shown in figure 34. However, the data is missing for the top thoracic vertebra and cervical vertebra. As a result, there is no precise way to estimate the point at which the first failure would occur. At any rate, the non-

symmetrical nature of the spinal column in terms of element breaking strength obviously makes the negative acceleration case a far different situation from the positive spinal case.

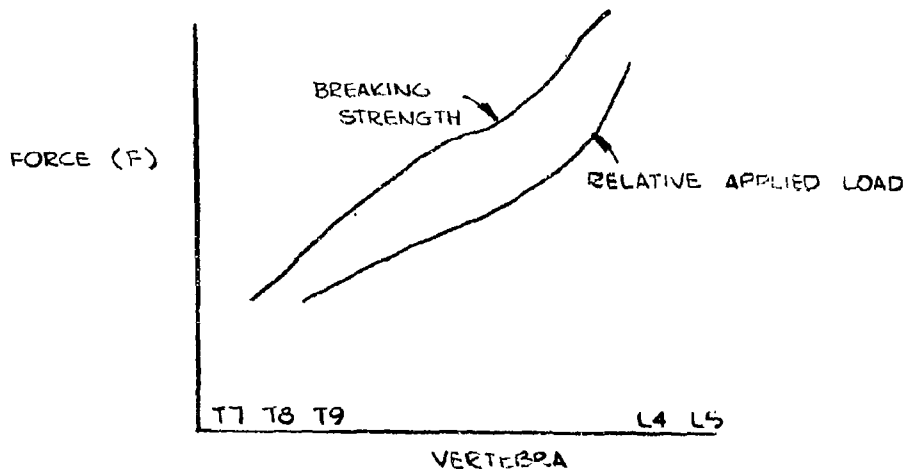


Figure 33. A Comparison of Vertebral Breaking Strength and Applied Loads (From Ruff, ref 13).

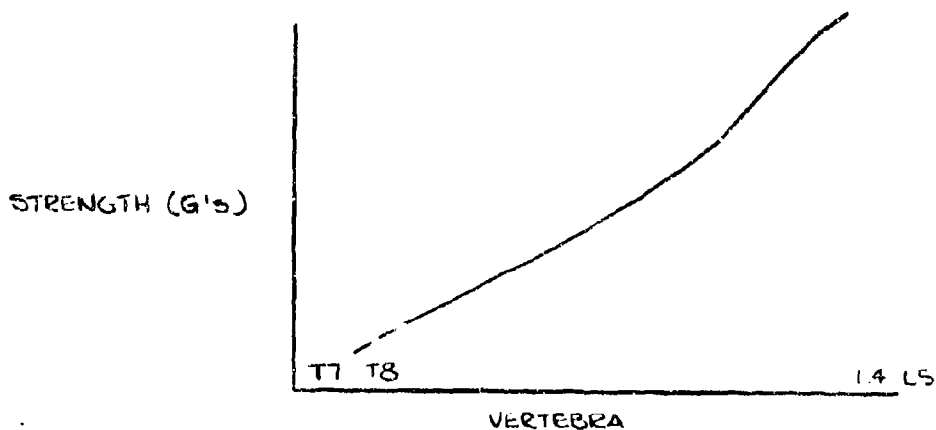


Figure 34. Vertebral Strength if Full Load is Reacted Through Shoulders in a Negative Spinal Acceleration.

The second difference between the positive and negative spinal cases is that in the latter the restraint system effects are of major importance in determining how the load is applied to the spinal column. If only a lap belt is used, then the spinal column will be placed in tension during a negative spinal acceleration. On the other hand, if only a shoulder harness is used, then the spinal column will be placed in compression and the situation cited in the previous paragraph will occur. Of course the load will probably be divided in some manner between a shoulder har-

ness and a lap belt, in which case the loading on the spinal column is a function of the relative deflections of the restraint elements.

From the preceding discussion, the basic dynamic model of the human body in the spinal direction that has been described is really applicable only to the positive spinal situation. In using this spinal model it must be remembered the restraint harness has an effect on the shape of the vertebral column. The present positive spinal model of the human body assumes good positioning of the vertebral column, so improper positioning must be taken into account in any real situation by reducing the load values at which damage occurs. One other factor to be kept in mind is the effect of multiple vector acceleration forces. If a spinal acceleration also has a component in the transverse direction, the force may tend to bend the vertebral column and cause column buckling to occur more readily. Since the spine is made up of series of pin-jointed short columns, it has low resistance to buckling. The prediction of maximum loads that will injure the vertebrae must be expected to be less for such cases.

To summarize the situation with respect to the dynamic model of the human body in the seated position subjected to a positive spinal acceleration, the following information is available:

- (1) The natural frequency, damping coefficient, and critical force levels in the spinal direction have been estimated, assuming proper positioning, and agree with data from several different types of experiment.
- (2) The effect of age on both the natural frequency and critical force has been estimated from cadaver data and appears to be consistent with results from other types of experiment.
- (3) The nonlinearity of the spinal column has been evaluated.
- (4) The lack of information on negative spinal loadings has prevented development of a suitable dynamic model although some of the factors involved are presented and no serious handicap to the eventual development of such a model is foreseen.
- (5) The degrading effect of arbitrary vector accelerations must be considered in operational situations.

Section VI

A DYNAMIC MODEL OF THE BODY IN THE TRANSVERSE DIRECTION

In the transverse direction, that is, with the acceleration applied through the chest, either front-to-back or back-to-front, the situation is somewhat more complicated than in the spinal direction. At least, it is much more difficult to specify precisely the mechanisms involved in the body's response to acceleration. This is because there are a number of organs and structural elements involved in the response to transverse accelerations.

There are two sources of data concerning the response of the body to transverse accelerations. In the impact regime, there are a number of recorded accidental free falls in which the victim impacted in the full supine or full prone positions. DeHaven (ref 18) pioneered in the analysis of such data. The second source of data is from experimental sled runs conducted with the subject sitting facing either forward or aft (ref. 5). There is a relatively large amount of the latter data available but, for the obvious reasons, only a few injury points are available.

DeHaven recorded some remarkable cases of survival involving rather enormous free fall distances. His data indicated that a human being would be injured, but could survive an impact velocity of 80 feet per second. DeHaven used data involving jumps or falls from buildings, as a general rule, but there is now another source of free fall data that he did not consider; the accident reports of the American Alpine Club (ref 24, 25, and 26). These data were analyzed to obtain an independent estimate of the critical impact velocity.

Briefly, the method used by the authors in analyzing the American Alpine Club statistics was to assign an injury severity rating to each reported accident, using the available information in the accident report and rating the severity with the scale developed by Aviation Crash Injury Research at the Cornell Aeronautical Laboratories. Figure 35 shows the individual accidents plotted against the velocity changes involved.

From figure 35, note that a few accidents involved head impact and that, in general, these accidents resulted in a higher injury severity rating than equivalent accidents where no head impact was involved. Also note that a few impacts at extreme velocities resulted in death, but that death in these cases is a qualitative and not a quantitative measure. In other words, injury severity rating number 10 represents a fatality resulting from lethal injuries in three or more regions of the body. Obviously, a case could involve just three regions of the body or could involve total mechanical demolition of the body. At any rate, the extreme velocity and death points can be ignored from the standpoint of the remainder of this analysis, because of this scaling artifact.

A straight best-fit line was calculated using the method of least squares and is shown in figure 36. Using this line as representative of the data recorded, the relationship between injury severity and impact velocity could be specified quite precisely. The midpoint of the injury severity scale was taken as the point representing 50% probability of injury in the same sense as used in other areas of human body dynamics. Using this value, an estimate of 53 feet per second is found for the 50% probability of injury level. Note that this is 27 feet per second below the value obtained from DeHaven's results. The disparity is shown in figure 37 where the DeHaven points are plotted relative to the 53 feet per second line derived from the mountaineer free fall data.

The obvious question that arises is what is the reason for the disparity between the two results? A relatively simple explanation is at hand. The DeHaven data represents a series of mirac-

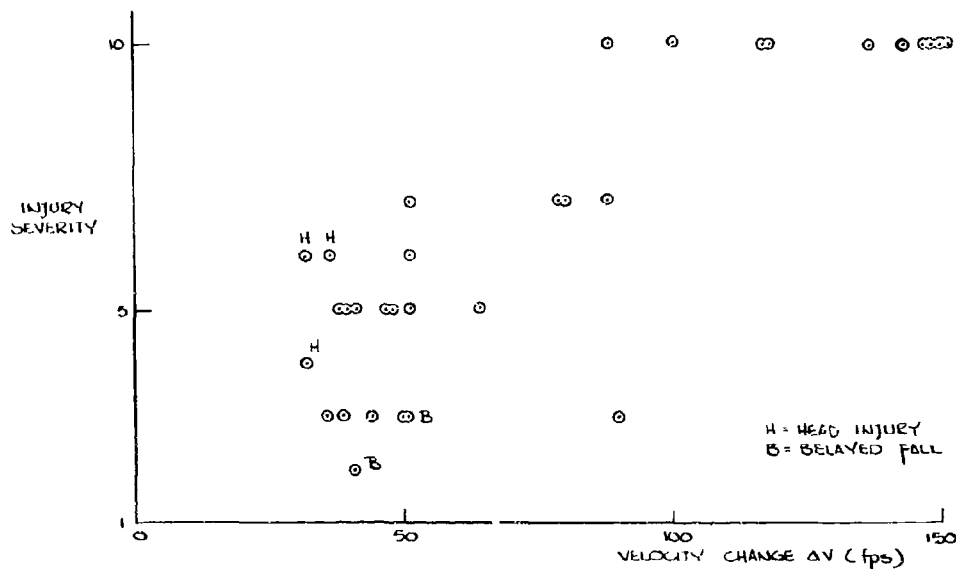


Figure 35. Mountaineering Free Fall Data from American Alpine Club Accident Reports.

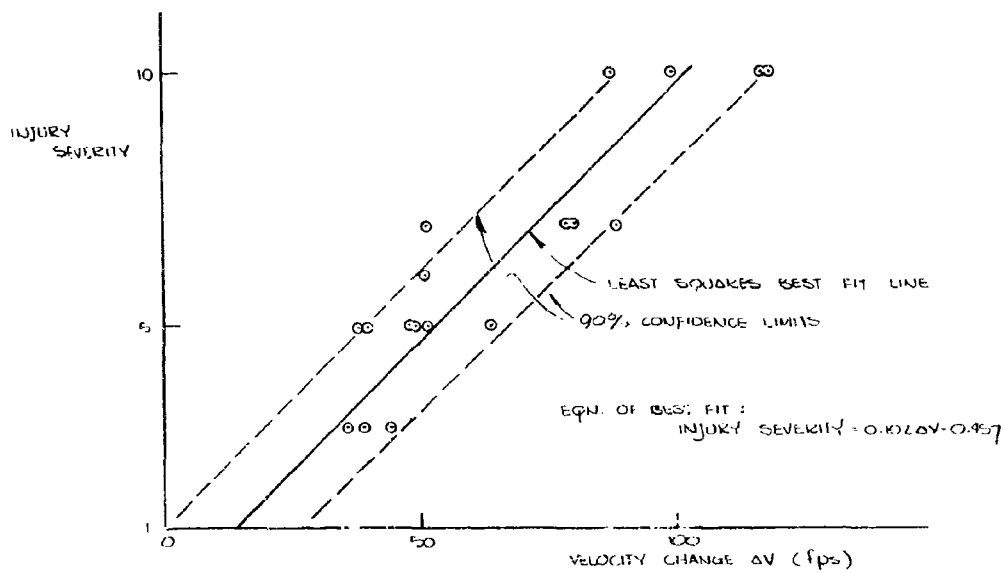


Figure 36. Calculated Best Fit Line and 90% Confidence Limits for Injury of Severity Versus ΔV .

ulous survivals of free falls from great heights. In other words, DeHaven used only data points in which the individuals survived the free fall incident, which constitutes a biased sample from a statistical standpoint. The American Alpine Club data, on the other hand, represents cases in which individuals were injured or killed at various impact velocity levels. Thus, the mountaineer free fall data can be considered more representative in a statistical sense and the estimate of 53 feet per second for the 50% probability of injury level is considered to be a reasonably good estimate of the actual capability of the human body to withstand transverse impact.

There is a relatively large amount of data on the exposure of the human body to short-term transverse accelerations, but very few injuries have been recorded in such tests. Therefore, it is somewhat more difficult to specify precisely the probability of injury levels in this regime. The available data points are plotted in figure 38 relative to the 53 feet per second tolerance line and

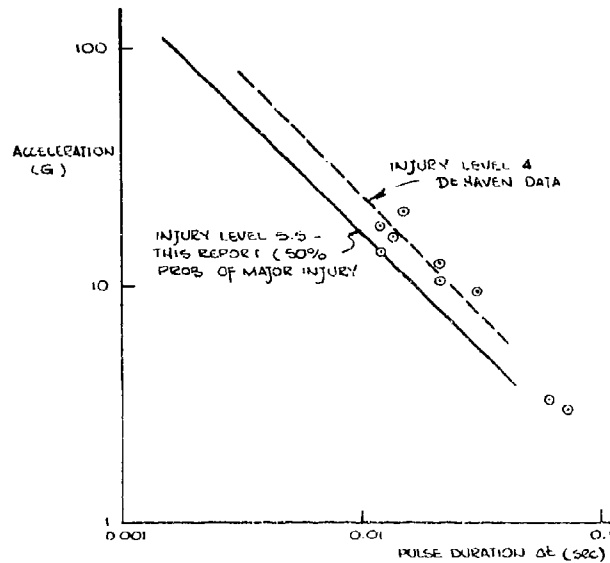


Figure 37. DeHaven Free Fall Data Versus Results of AAC Accident Analysis.

also relative to the "minor or no injury" line. Using a pulse duration of approximately 0.04 seconds as indicative of the transition point between the impact region and the regime in which acceleration peak is the governing factor, all of the sled tests data points involving injury fall above the extended minor or no injury line, and below the 50% probability of injury line. Thus, the short-period data, although sparse, tend to confirm the velocity change estimates. Because of the relative lack of data and unknown validity of the existing data, the results of this analysis must be viewed in a somewhat negative light, that is, the data is consistent within itself but does not provide a high degree of confidence or very much precision.

The corner duration, defined as the pulse duration at which the important parameter changes from peak acceleration to critical velocity change, can be calculated by dividing the critical velocity by the critical peak acceleration (in ft/sec²). Using the previously determined value of 53 feet per second for the velocity change value and 40 G as an estimate for the peak acceleration

regime, then the corner duration becomes 0.0411 seconds. If we take the damping ratio value of $\bar{c}=0.3$ from impedance measurements (ref 10) we get the following results for the transverse direction:

PARAMETER	SMALL AMPLITUDE IMPEDANCE (Semisupine Position)	IMPACT
Subject age	47.0 years	47.0 years
Damping coefficient \bar{c}	0.3	0.23
$\omega/2\pi$	7.41	9.68 cps

Note that, due to nonlinearities of the system, the effective frequency under impact loading is 30% higher than the value from the impedance measurements. This is less than for the spinal case, where an increase of over 40% is indicated in section V of this report.

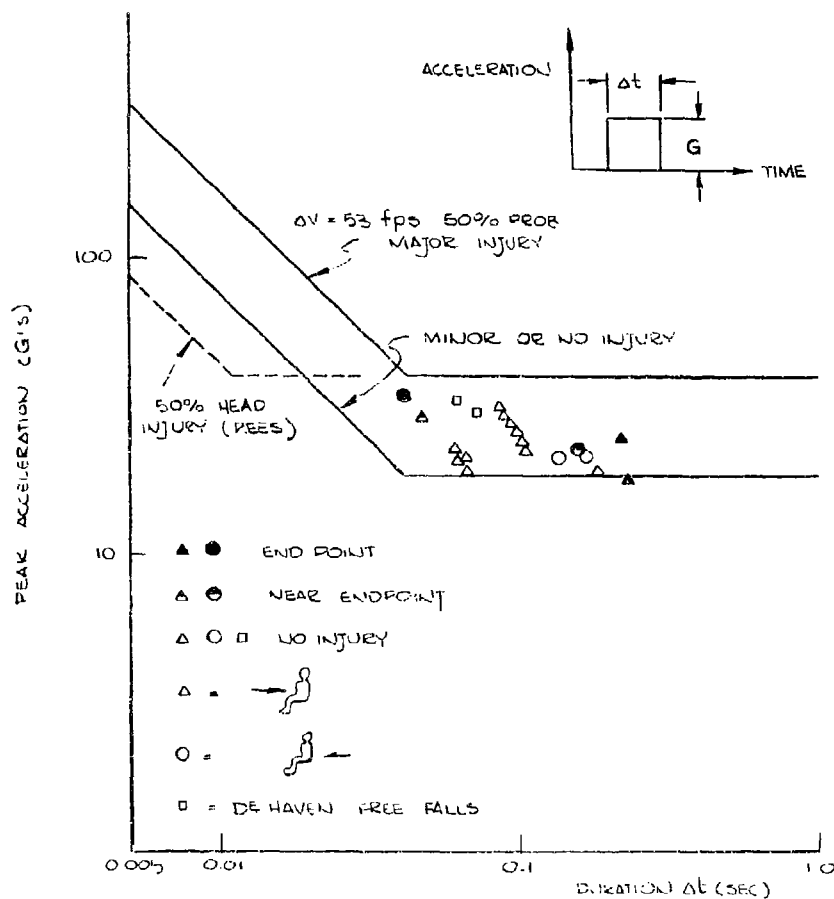


Figure 38. Human Body Injury Levels Resulting From Transverse Accelerations.

The dynamic model of the human body in the transverse direction can now be summarized as shown in table VII.

TABLE VII
DYNAMIC MODEL OF THE HUMAN BODY – TRANSVERSE
SINGLE DEGREE OF FREEDOM LINEAR MODEL

Natural Frequency— $\omega/2\pi$ —cps	9.68
Damping Ratio— \bar{c}	.23
50% Injury Levels:	
Steady State— \ddot{Y}_{crit} —G	40
Impulsive Velocity Change— ΔV —fps	53

Because of the lack of data, the above transverse body model must be assumed to be invariant with age, although additional data may show a variation with age such as that reported in section V. Similarly, the lack of data prevents even a crude estimate of the variation in probability of injury levels. However, the injury severity levels calculated from the mountaineer free fall data can be extrapolated to the short-term region for approximate estimates of injury severity. The investigator must familiarize himself with the ACIR injury severity scale, which can be found in an appendix to ref 28.

Section VII

OTHER DYNAMIC MODELS OF THE HUMAN BODY

In the preceding sections, dynamic models for the positive spinal direction and for both the positive and negative transverse directions were described. In the transverse dynamic model, the body is assumed to be symmetrical. When more data are available, it may well be that two transverse models will be required, one for the positive direction and one for the negative. The single model developed at the present time may be used for engineering purposes until more refined models can be constructed.

There is virtually no data available concerning the response of the human body to lateral accelerations. One series of tests has been conducted on rhesus monkeys by the Aerospace Medical Research Laboratory at Wright-Patterson Air Force Base, in anticipation of further testing with human beings. Such human testing has been conducted, but it was an evaluation of an operational restraint system and did not result in end point data, so it still is not possible to develop other than a hypothetical lateral dynamic model of the human body. For engineering purposes, the transverse dynamic model of the body may be used since no other information is available. However, the critical loads in the lateral direction may turn out to be significantly less.

In addition to the six major acceleration vectors applied to the whole body, the response of the head to acceleration inputs must be taken into account under certain circumstances. Because of the lack of suitable experimental data, the confidence in existing dynamic models of the human head is fairly low. However, the natural frequency and steady-state acceleration tolerance of the head are greater than for the torso, in either the transverse or spinal directions. (This assumes a reasonable amount of cushioning material between the head and the input acceleration surface provides sufficient attenuation to raise the head injury levels above those for the torso.) Thus, there is very little head injury data available from operational or accident statistics.

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13. ABSTRACT Some of the dynamic characteristics of the human body were analyzed and mathematical analogs that can be used to predict the response of the body to acceleration environments were developed. The background of the acceleration tolerance problem, the basic concepts of body dynamics, and the available experimental data were also reviewed. Both spinal and transverse dynamic models are presented together with the data used in obtaining frequency, damping, and breaking strength estimates. Other dynamic models are discussed more briefly since very little is known about the dynamic response to lateral and negative spinal accelerations. A discussion of the factors influencing the production of injury from exposure to accelerations is presented to indicate the difficulty in defining human tolerance to acceleration using the classical approach of graphs or simple critical G values that depend only upon the duration of the acceleration. The concept of relative probability of injury is developed in order to take into account the variations in the human structure and other factors that influence human tolerance to accelerations.		

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