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# AIRCRAFT CRASH SURVIVAL DESIGN GUIDE

## VOLUME II - AIRCRAFT CRASH ENVIRONMENT AND HUMAN TOLERANCE



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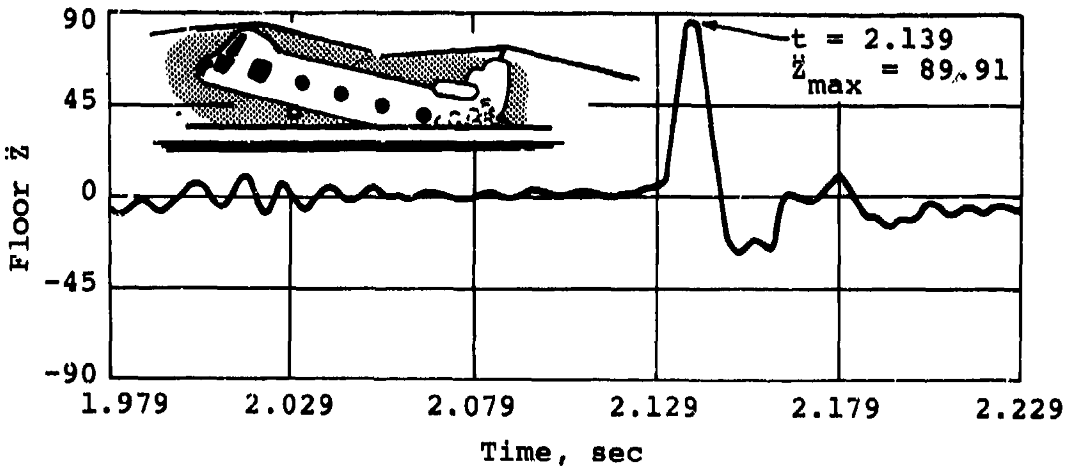
FINAL REPORT

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## APPLIED TECHNOLOGY LABORATORY POSITION STATEMENT

This revised edition of the Crash Survival Design Guide was prepared for the Applied Technology Laboratory by Simula, Inc., under the terms of Contract DAAJ02-77-C-0021. The original Crash Survival Design Guide was published in 1967 as USAAVLABS Technical Report 67-22 and subsequent revisions were published as USAAVLABS Technical Report 70-22 and USAAMRDL Technical Report 71-22. This current edition consists of a consolidation of design criteria, concepts, and anlytical techniques developed through research programs sponsored by this Laboratory over the past 20 years into one report suitable for use as a designer's guide by aircraft design engineers and other interested personnel.

This document has been coordinated with USAAVRADCOM, the U. S. Army Safety Center, the U. S. Army Aeromedical Research Laboratory, and several other Government agencies active in aircraft crashworthiness research and development.

The technical monitors for this program were Messrs. G. T. Singley III, R. E. Bywaters, W. J. Nolan, and H. W. Holland of the Safety and Survivability Technical Area, Aeronautical Systems Division, Applied Technology Laboratory.

Comments or suggestions pertaining to this Design Guide will be welcomed by this Laboratory.



NOTE: This is a revised edition of the Crash Survival Design Guide (formerly USAAMRDL Technical Report 71-22). All previous editions are obsolete and should be destroyed.

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20. ABSTRACT (Continue on reverse side if necessary and identify by block number)

This five-volume document has been assembled to assist design engineers in understanding the problems associated with the development of crashworthy U. S. Army aircraft. Contained herein are not only a collection of available information and data pertinent to aircraft crashworthiness but suggested design conditions and criteria as well. The five volumes of the Aircraft Crash Survival Design Guide cover the following topics:

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Volume I, Design Criteria and Checklists;
Volume II, Aircraft Crash Environment and Human Tolerance;
Volume III, Aircraft Structural Crashworthiness;
Volume IV, Aircraft Seats, Restraints, and Litters; and
Volume V, Aircraft Postcrash Survival.

This volume (Volume II) contains information on the aircraft crash environment, human tolerance to impact, occupant motion during a crash, human anthropometry, and crash test dummies, all of which serves as background for the design information presented in the other volumes.

## PREFACE

This report was prepared for the Safety and Survivability Technical Area of the Applied Technology Laboratory, U. S. Army Research and Technology Laboratories (AVRADCOM), Fort Eustis, Virginia, by Simula Inc. under Contract DAAJ02-77-C-0021, initiated in September 1977. The Department of the Army Project Number is 1L162209AH76. This guide is a revision of USAAMRDL Technical Report 71-22, Crash Survival Design Guide, published October 1971.

A major portion of the data contained herein was taken from U. S. Army-sponsored research in aircraft crashworthiness conducted from 1960 to 1979. Acknowledgment is extended to the U. S. Air Force, Federal Aviation Administration, NASA, and U. S. Navy for their research in crash survival. Appreciation is extended to the following organizations for providing accident case histories leading to the establishment of the impact conditions in aircraft accidents:

- U. S. Army Safety Center, Fort Rucker, Alabama.
- Civil Aeronautics Board, Washington, D. C.
- U. S. Naval Safety Center, Norfolk, Virginia.
- U. S. Air Force Inspection and Safety Center, Norton Air Force Base, California.

Additional credit is due the many authors, individual companies, and organizations listed in the bibliographies for their contributions to the field. The contributions of the following authors to previous editions of the Crash Survival Design Guide are most noteworthy:

> D. F. Carroll, R. L. Cook, S. P. Desjardins, J. K. Drummond, J. H. Haley, Jr., A. D. Harper, H. G. C. Henneberger, N. B. Johnson, G. Kourouklis, W. H. Reed, S. H. Robertson, L. M. Shaw, J. W. Turnbow, and L. W. T. Weinberg.

This volume has been prepared by Dr. D. H. Laananen. Data from the investigations of recent Army aircraft accidents were provided by the U. S. Army Safety Center. R. F. Chandler of the FAA Civil Aeromedical Institute furnished assistance in locating recent information on human tolerance, anthropometry, and crash test dummies.

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## INTRODUCTION

For many years, emphasis in aircraft accident investigation was placed on finding the cause of the accident. Very little effort was expended in the crash survival aspects of aviation safety. However, it became apparent through detailed studies of accident investigation reports that large improvements in crash survival could be made if consideration were given in the initial aircraft design to the following general survivability factors:

1. Crashworthiness of Aircraft Structure - The ability of the aircraft structure to maintain living space for occupants throughout a crash.

2. Tiedown Chain Strength - The strength of the linkage preventing occupant, cargo, or equipment from becoming missiles during a crash sequence.

3. Occupant Acceleration Environment - The intensity and duration of accelerations experienced by occupants (with tiedown assumed intact) during a crash.

4. Occupant Environment Hazards - Barriers, projections, and loose equipment in the immediate vicinity of the occupant that may cause contact injuries.

5. Postcrash Hazards - The threat to occupant survival posed by fire, drowning, exposure, etc., following the impact sequence.

Early in 1960, the U. S. Army Transportation Research Command* initiated a long-range program to study all aspects of aircraft safety and survivability. Through a series of contracts with the Aviation Safety Engineering and Research Division (AvSER) of the Flight Safety Foundation, the problems associated with occupant survival in aircraft crashes were studied to determine specific relationships between crash forces, structural failures, crash fires, and injuries. A series of reports covering this effort was prepared and distributed by the U. S. Army, beginning in 1959. In October 1965, a special project initiated by the U. S. Army consolidated the design criteria presented in these reports into one technical document suitable for use as a designer's guide by aircraft design

*Now the Applied Technology Laboratory, Research and Technology Laboratories of the U. S. Army Aviation Research and Development Command (AVRADCOM).

engineers and other interested personnel. The document was to be a summary of the current state of the art in crash survival design, using not only data generated under Army contracts, but also information collected from other agencies and organizations. The Crash Survival Design Guide, first published in 1967, realized this goal.

Since its initial publication, the Design Guide has been revised several times to incorporate the results of continuing research in crashworthiness technology. The last revision of TR 71-22 was the basis for the criteria contained in the Army's crashworthiness military standard, MIL-STD-1290(AV), "Light Fixed- and Rotary-Wing Aircraft Crashworthiness" (Reference 1). This current revision, the fourth, contains the most comprehensive treatment of all aspects of aircraft crash survival now documented. It can be used as a general text to establish a basic understanding of the crash environment and the techniques that can be employed to improve chances for survival. It also contains design criteria and checklists on many aspects of crash survival and thus can be used as a source of design requirements.

The current edition of the Aircraft Crash Survival Design Guide is published in five volumes. Volume titles and general subjects included in each volume are as follows:

Volume I - Design Criteria and Checklists

Pertinent criteria extracted from Volumes II through V, presented in the same order in which they appear in those volumes.

Volume II - Aircraft Crash Environment and Human Tolerance

Crash environment, human tolerance to impact, military anthropometric data, occupant environment, test dummies, accident information retrieval.

Volume III - Aircraft Structural Crashworthiness

Crash load estimation, structural response, fuselage and landing gear requirements, rotor requirements, ancillary equipment, cargo restraints, structural modeling.

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1. Military Standard, MIL-STD-1290(AV), LIGHT FIXED- AND ROTARY-WING AIRCRAFT CRASHWORTHINESS, Department of Defense, Washington, D. C., 25 January 1974.

Volume IV - <u>Aircraft Seats, Restraints, and Litters</u>

Operational and crash environment, energy attenuation, seat design, litter requirements, restraint system design, occupant/restraint system/seat modeling.

Volume V - <u>Aircraft Postcrash Survival</u>

Postcrash fire, ditching, emergency escape, crash locator beacons.

This volume (Volume II) contains information on the aircraft crash environment and the response of the human body to such an environment. Following a general discussion of aircraft crashworthiness in Chapter 1, a number of words commonly referred to in discussing the crash environment are defined in Chapter 2. Chapter 3 describes the crash environment itself, in terms of impact conditions, terrain, and the nature and frequency of different types of injuries. Chapter 4 discusses the tolerance of the human body and various body parts to impact loading. Chapter 5 presents data on occupant motion during a crash. Chapter 6 provides data on human anthropometry that may be useful directly, as in cockpit design, or indirectly, as in preparing input for computer simulation models such as those discussed in Volume IV. Chapter 7 describes the crash test dummies used in evaluating protective systems such as seats and restraints.

# 1. BACKGROUND DISCUSSION

This volume deals with the variables involved in the Army aircraft crash environment and the effects of forces in that environment on the human body. An understanding of the environment and of the ability of the human body to survive it is necessary for the effective design of more crashworthy aircraft. The following background discussion presents general considerations that are of importance in understanding and applying the included information.

The overall objective of designing for crashworthiness is to eliminate unnecessary injuries and fatalities in relatively mild impacts. Results from analyses and research during the past several years have shown that the relatively small cost in dollars and weight of including crashworthy features is an extremely wise investment. Consequently, new generation aircraft are being procured to rather stringent crashworthy requirements.

To maximize aircraft crashworthiness, or, in the sense being discussed, to provide as much occupant protection as possible, all aspects of the complete system must be considered. In other words, every available subsystem must be employed to the fullest extreme in order to maximize the protection afforded to vehicle occupants. When an aircraft impacts the ground, deformation of the ground absorbs some energy. This is an uncontrolled variable since the quality of the impacted surface usually cannot be selected by the pilot. If the aircraft lands in the proper attitude, the landing gear can be used to absorb a significant amount of the impact energy. After stroking of the gear, crushing of the fuselage provides the next level of energy absorption. Of course, one of the functions of the fuselage is to provide a protective shell around the occupant while energy-absorbing stroke is occurring outside the shell. The functions of the seat and restraint system are to restrain the occupant within the protective shell during the crash sequence and to provide additional energy-absorbing stroke to further reduce the loads. The structure and components immediately surrounding the occupant also must be considered. Structures such as cyclic controls, glare shields, instrument panels, and sidewalls must be delethalized in some manner if they lie within the strike envelope of the occupant.

The original edition of this design guide dealt primarily with modifications that could be made to existing aircraft to increase their crashworthiness; however, now, two approaches to improving aircraft crashworthiness are open. The first approach is to influence the design of new aircraft, and the second is to improve the crashworthiness of existing aircraft. Obviously, much higher levels of crashworthiness can be achieved in the design and development of new aircraft if crashworthiness is considered from the beginning. This is being accomplished at the present time through the use of procurement packages that include pertinent specifications that require certain levels of crashworthiness of various subsystems as well as for the entire aircraft. However, some of the available potential is still being lost due to the historical approach used in designing aircraft. The basic aircraft is designed leaving space and providing attachment provisions for subsystems. Later the subsystems are designed and then are limited by the previously established, somewhat arbitrary boundary conditions. The boundary conditions may unnecessarily limit the performance of the subsystems. The better approach is the systems approach in which all systems and subsystems are, at least preliminarily, designed at the same time. This approach enables subsystem considerations to affect the larger systems and will produce a more nearly optimum vehicle.

The same principles for improving crashworthiness can be applied to the retrofit of existing aircraft; however, the "cast-in-concrete" status of existing production structure is a more costly and difficult obstacle to overcome. When crashworthiness features must be included through retrofit, the level that can be achieved is usually reduced. Even in retrofit situations, however, the overall objective can be met; i.e., occupant protection can be maximized to eliminate unnecessary injuries.

As mentioned above, the entire system should be considered in any analysis resulting in apportionment of the crash energy to be absorbed by the various components. However, any valid systems approach will consider probable alternate crash environments wherein all subsystems cannot perform their desired functions; for example, an impact situation in which the landing gear cannot absorb its share of the impact crash energy because of angle of impact or loss of gear. To achieve the overall goal, therefore, minimum levels of crash protection have been required of the various individual subsystems, such as the seat.

In earlier editions of the Design Guide, the requirements to provide occupant protection in crashes up to and including the severity of the 95th-percentile survivable crash pulse was expressed. With the deployment of aircraft designed for crash safety, the link to the 95th-percentile survivable crash pulse has been dropped, and the recommended design environment is simply presented as the design pulse. Obviously, the severity of a 95th-percentile survivable crash pulse will be much greater for the new aircraft than for aircraft having no crashworthy requirements placed upon them during their development. The extent of the crash protection provided to the occupant cannot indefinitely continue to be linked to the survivability of the crash as improved crashworthiness increases the severity of the survivable crash producing a never-ending increase in the level of crashworthiness at the expense of aircraft performance. The crashworthiness levels recommended herein are felt to be a near optimum mix of requirements including consideration of cost, weight, and performance. The crash environments selected for design purposes in this volume are identical to the historical 95th-percentile survivable crash pulses.

Also in earlier editions of the Design Guide, information was provided on design of fixed-wing transport aircraft. Considering the volume of new information on crashworthy design, in an effort to ensure that the size of this document remains within reasonable limits, only the primary aircraft in the Army inventory are considered. Therefore, information given herein is intended to apply to rotary-wing aircraft and light fixed-wing aircraft, defined by a mission gross weight of 12,500 lb or less.

# 2. DEFINITIONS

## 2.1 GENERAL TERMS

The following text defines words commonly used in discussions of the aircraft crash environment:

- The Term "G"

  The ratio of a particular acceleration (a negative acceleration may be referred to as a deceleration) to the acceleration due to gravitational attraction at sea level (32.2 ft/sec²). With respect to the crash environment, unless otherwise specified, all acceleration values (G) are those at a point approximately at the center of the fuselage floor. In accordance with common practice, this report will refer to accelerations measured in G. To illustrate, it is customarily understood that 5 G represents an acceleration of 5 x 32.2, or 161 ft/sec². As a result, crash forces can be thought of in terms of multiples of the weight of objects being accelerated. Therefore, also in keeping with common practice, the term G is used in this document to define accelerations or forces.

- Survivable Accident

  An accident in which the forces transmitted to the occupant through his seat and restraint system do not exceed the limits of human tolerance to abrupt accelerations and in which the structure in the occupant's immediate environment remains substantially intact to the extent that a livable volume is provided for the occupants throughout the crash sequence.

- Survival Envelope

  The range of impact conditions--including magnitude and direction of pulses and the duration of forces occurring in an aircraft accident--wherein the occupiable area of the aircraft remains substantially intact, both during and following the impact, and the forces transmitted to the occupants do not exceed the limits of human tolerance when current state-of-the-art restraint systems are used.

  It should be noted that, where the occupiable volume is altered appreciably through elastic deformation during the impact phase, survivable conditions may not have existed in an accident that, from postcrash inspection, outwardly appeared to be survivable.

## 2.2 AIRCRAFT PARAMETERS

- Aircraft Coordinates

Positive directions for velocity, acceleration, and force components and for pitch, roll, and yaw are illustrated in Figure 1.

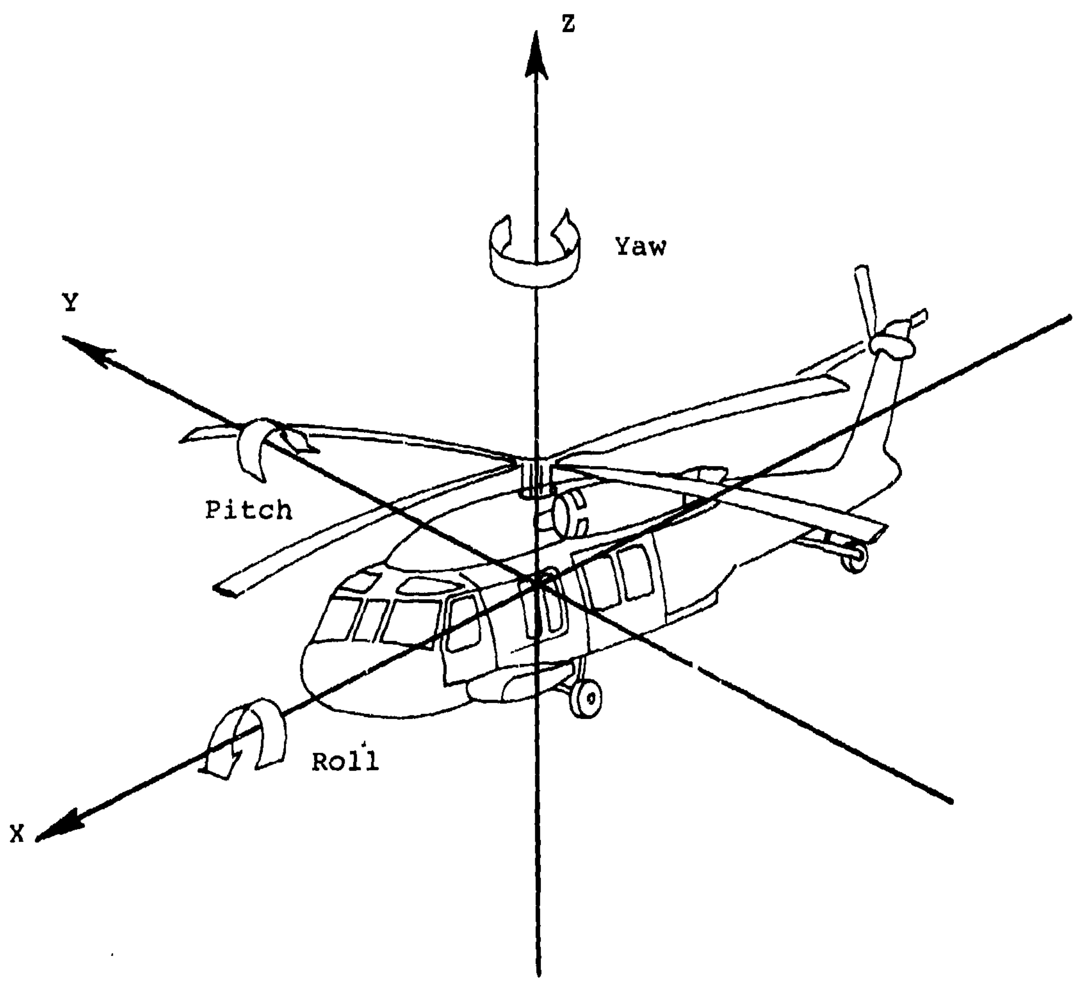



Figure 1. Aircraft coordinate and attitude directions.

- <u>Velocity Change in Major Impact (ΔV)</u>

The decrease in velocity of the airframe during the major impact, expressed in feet per second. The major impact is the one in which highest forces are incurred, not necessarily the initial impact. For the acceleration pulse shown in Figure 2, the major impact should be considered ended at time $T_2$. Elastic recovery in the structure will tend to reverse the direction of aircraft velocity prior to $T_2$.

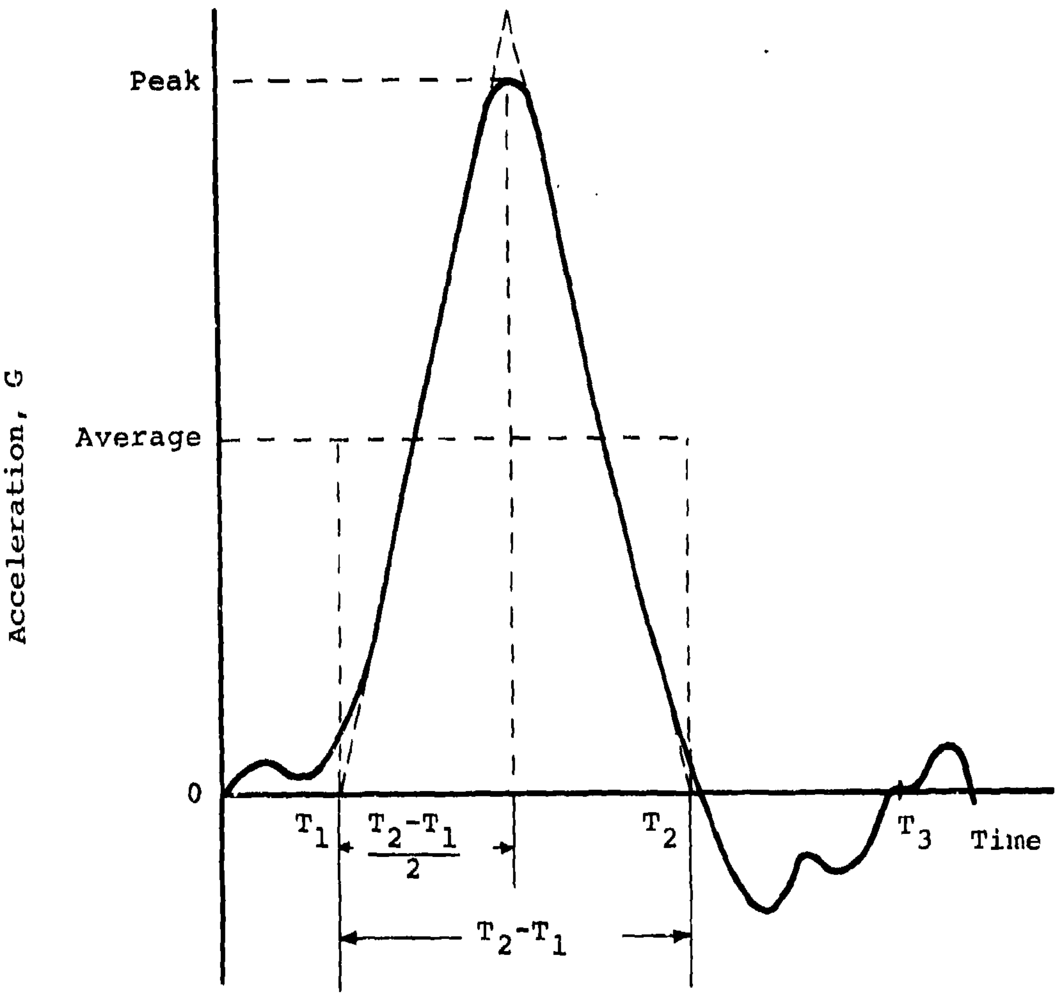



Figure 2. Typical aircraft floor deceleration pulse.

Should the velocity actually reverse, its direction must be considered in computing the velocity change. For example, an aircraft impacting downward with a vertical velocity component of 30 ft/sec and rebounding with an upward component of 5 ft/sec should be considered to experience a velocity change

$$\Delta V = 30 - (-5) = 35 \text{ ft/sec}$$

during the major impact. After the aircraft rebounds upward, gravity will accelerate it downward again, as illustrated by the negative acceleration between $T_2$ and $T_3$ in Figure 2.

- Longitudinal Velocity Change

  The decrease in velocity during the major impact measured along the longitudinal (roll) axis of the aircraft. The velocity may or may not reach zero during the major impact. For example, an aircraft impacting the ground at a forward velocity of 100 ft/sec and slowing to 35 ft/sec before rebounding into the air would experience a longitudinal velocity change of 65 ft/sec during this impact.

- Vertical Velocity Change

  The decrease in velocity during the major impact measured along the vertical (yaw) axis of the aircraft. The vertical velocity generally reaches zero during the major impact.

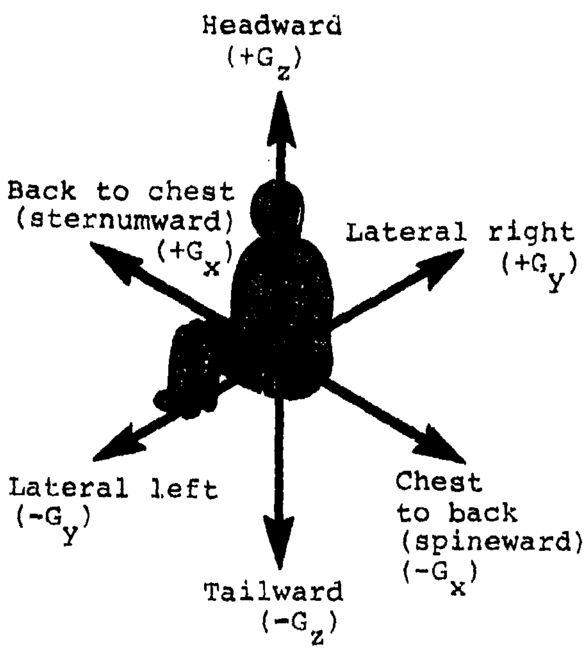
- Abrupt Accelerations

  Accelerations of short duration primarily associated with crash impacts, ejection seat shocks, capsule impacts, etc. One second is generally accepted as the dividing point between abrupt and prolonged accelerations. Within the extremely short duration range of abrupt accelerations (0.2 sec and below), the effects on the human body are limited to mechanical overloading (skeletal and soft tissue stresses), there being insufficient time for functional disturbances due to fluid shifts.

## 2.3 HUMAN BODY PARAMETERS

- Human Body Coordinates

In order to minimize the confusion sometimes created by the terminology used to describe the directions of forces applied to the body, a group of NATO scientists compiled the accelerative terminology table of equivalents shown in Figure 3 (Reference 2). Terminology used throughout this guide is compatible with the NATO terms as illustrated.




Direction of accelerative force

Vertical

Headward - Eyeballs-down
Tailward - Eyeballs-up

Transverse

Lateral right - Eyeballs-left
Lateral left - Eyeballs-right
Back to chest - Eyeballs-in
Chest to back - Eyeballs-out

Note:

The accelerative force on the body acts in the same direction as the arrows.

Figure 3. Terminology for directions of forces on the body.

- Human Tolerance to the Crash Environment

Obviously, the tolerance of the human body to crash environments is a function of many variables, including the unique characteristics of each person as

2. Gell, C. F., TABLE OF EQUIVALENTS FOR ACCELERATION TERMINOLOGY, Aerospace Medicine, Vol. 32, No. 12, December 1961, pp. 1109-1111.

well as the loading variables. The loads applied to the body include decelerative loads imposed by seats and restraint systems as well as localized forces due to impact with surrounding structure. Tolerable levels of the decelerative loads depend on the direction of the load, the orientation of the body, and the means of applying the load. For example, the critical nature of the loads parallel to the occupant spine manifests itself in any of a number of spinal fractures. Forces perpendicular to the occupant spine can produce spinal fracture through flexure that results from jackknife bending over a lap-belt-only restraint. The lap belt might inflict injuries to the internal organs if it is not retained on the pelvic girdle but is allowed to exert its force above the iliac crests in the soft stomach region. Excessive rotational or translational acceleration of the head can produce concussion. Further, skull fracture can result from localized impact with surrounding structure. Therefore, tolerance is a function of the method of occupant restraint as well as the variable of specific occupant makeup.

For the purposes of this document, human tolerance is defined as a selected array of parameters that describe a condition of decelerative loading for which it is believed there is a reasonable probability for survival without major injury. As used in this volume, designing for the limits of human tolerance refers to providing design features that will maintain these conditions at or below their tolerable levels to enable the occupant to survive the given crash environment.

- <u>Submarining</u>

  A rotation of the hips under and about the lap belt as a result of a forward inertial load exerted by deceleration of the thighs and lower legs accompanied by lap belt slippage up and over the iliac crests. Lap belt slippage up and over the iliac crests can be a direct result of the upward loading of the shoulder harness straps at the center of the lap belt.

- <u>Dynamic Overshoot</u>

  The amplification of decelerative force on cargo or personnel above the floor input decelerative force (ratio of output to input). This amplification is a result of the dynamic response of the system.

# 3. AIRCRAFT CRASH ENVIRONMENT

## 3.1 INTRODUCTION

This chapter presents the results of statistical studies conducted to determine impact conditions that occurred in accidents of various types of U. S. Army aircraft. Data were accumulated through a study of U. S. Army accidents for the period 1 July 1960 through 30 June 1965 and 1 January 1971 through 31 December 1976. Additional data considered pertinent were obtained from Civil Aeronautics Board, U. S. Navy, and U. S. Air Force accident reports.

Aircraft included here are rotary-wing and light fixed-wing of mission gross weight no greater than 12,500 lb. The accident cases selected were limited to those in which one or more of the following factors applied: (1) Substantial structural damage, (2) postcrash fire, (3) personnel injuries, and (4) at least one person surviving the crash. Mid-air collisions and other accidents resulting in catastrophic uncontrolled free falls from altitudes of a hundred feet or more were not considered. Such accidents almost invariably result in random, unpredictable crash kinematics and nonsurvivable impact forces, and are of little value in establishing realistic crash survival envelopes that would be useful to the aircraft designer. Although accidents involving postcrash fire were considered where possible, analysis of impact forces in many of the accidents involving fire was impossible due to extensive burn damage to the aircraft, and these cases were not used in the study. Still other accident reports simply provided insufficient or inadequate data to permit a detailed analysis of the case.

The impact accelerations computed or estimated for each accident by the original accident investigation boards were analyzed and recalculated or estimated by the survey team. This action was taken to minimize the variation in findings caused by the diversity in training and experience of the various members of the boards who made the initial calculations. Those accidents involving high impact forces that appeared to be near the upper limits of survivability were analyzed in depth. In the less severe accidents, which made up the bulk of the accidents surveyed, the impact forces given by original investigation boards or teams were analyzed primarily to eliminate any major errors or misjudgments by the aircraft accident board. Minor individual variations in findings in these less severe accidents were accepted with the assumption that the sum of the variations would tend to be zero and would not affect the accuracy of the final data.

Altogether, 563 rotary-wing accidents and 92 fixed-wing accidents were reviewed in the preparation of earlier editions of the Design Guide. However, only 373 total cases are used in establishing the impact conditions outlined here. Impact attitude data from an additional 108 attack and 10 cargo helicopters collected during 1971-1976 also are presented.

Analysis of the data from Army accident records showed a similarity in impact conditions between rotary- and light fixed-wing STOL aircraft (O-1, U-6, U-1). The typical severe STOL crash occurred with a large vertical velocity component in addition to a longitudinal component. The small amount of underfloor structure means that high vertical accelerations result at impact. The small amount of structure below the floor is a primary reason for high vertical acceleration in rotary-wing aircraft also. The data showed that, except for the lateral direction, the similarities between rotary-wing and lignt STOL aircraft impact conditions were sufficient to allow treating them as being the same.

## 3.2 VELOCITY CHANGES IN MAJOR IMPACTS

The velocity change of the airframe during major impacts (as defined in 2.2) was difficult to obtain in the majority of the accidents analyzed because of an inability to fix the impact velocity. The impact velocities in 40 accidents were known, and the velocity changes in the major impact pulses were estimated with errors probably not exceeding plus or minus 20 percent.

### 3.2.1 Vertical Velocity Changes

The vertical velocity change is defined as the decrease in velocity during the major impact measured along the vertical (yaw) axis of the aircraft. The vertical velocity generally reaches zero during the major impact.

The curve in Figure 4 shows the distribution of vertical velocity changes experienced by rotary- and light fixed-wing aircraft in survivable accidents involving substantial structural damage or occupant injury.

The median of the distribution in Figure 4 is a vertical velocity change of approximately 24 ft/sec, with one-half of the accidents involving vertical velocity changes greater than this figure and one-half having vertical velocity changes lower than this figure. For reference, 24 ft/sec is the velocity reached by a body during a free fall of 8 ft 11 in.

Ninety-five percent of the accidents represented by the curve in Figure 4 incurred vertical velocity changes of less than

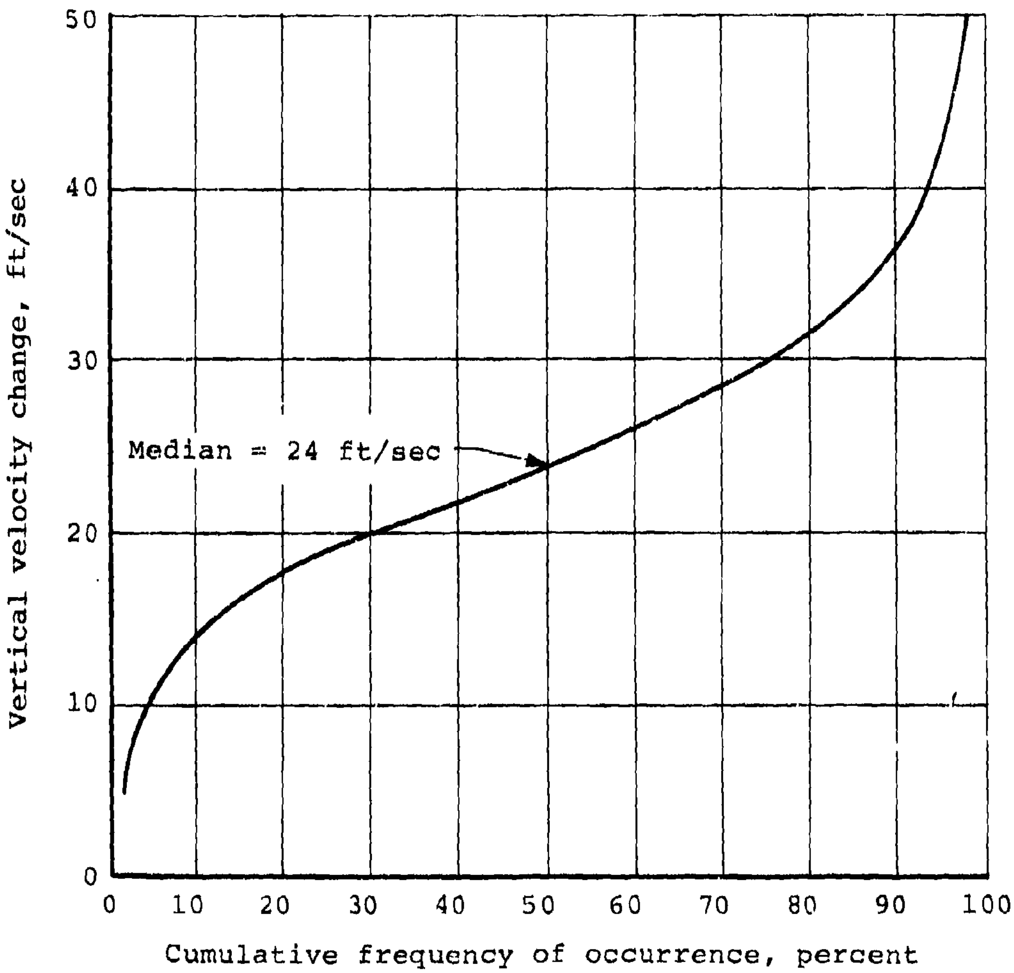



Figure 4. Vertical velocity changes for survivable rotary- and light fixed-wing aircraft accidents.

42 ft/sec. A free fall of 27 ft 5 in. is required for a body to reach a velocity of 42 ft/sec.

### 3.2.2 Longitudinal Velocity Changes

The curve in Figure 5 depicts the distribution of longitudinal velocity changes incurred by rotary- and light fixed-wing aircraft in survivable accidents involving substantial structural

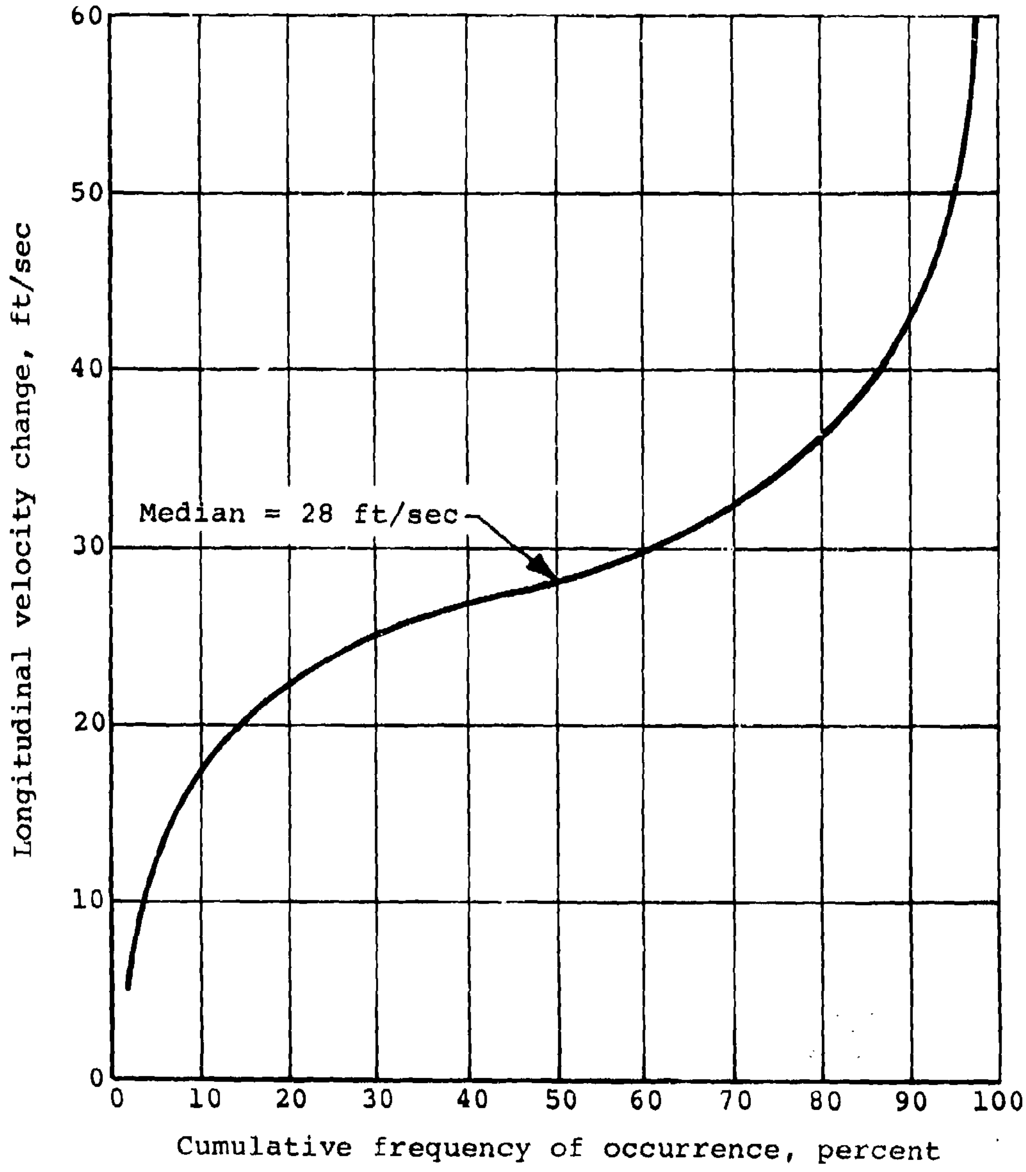



Figure 5. Distribution of longitudinal velocity changes for survivable rotary- and light fixed-wing aircraft accidents.

damage or occupant injury. The curve gives a relative frequency of occurrence of the stated longitudinal velocity changes.

The median of the distribution curve in Figure 5 is a longitudinal velocity change of approximately 28 ft/sec, with one-half of the accidents involving longitudinal changes greater than this figure and one-half experiencing longitudinal changes lower than this figure.

Ninety-five percent of the accidents represented by the curve in Figure 5 incurred longitudinal velocity changes of less than 50 ft/sec.

### 3.2.3 Lateral Velocity Changes

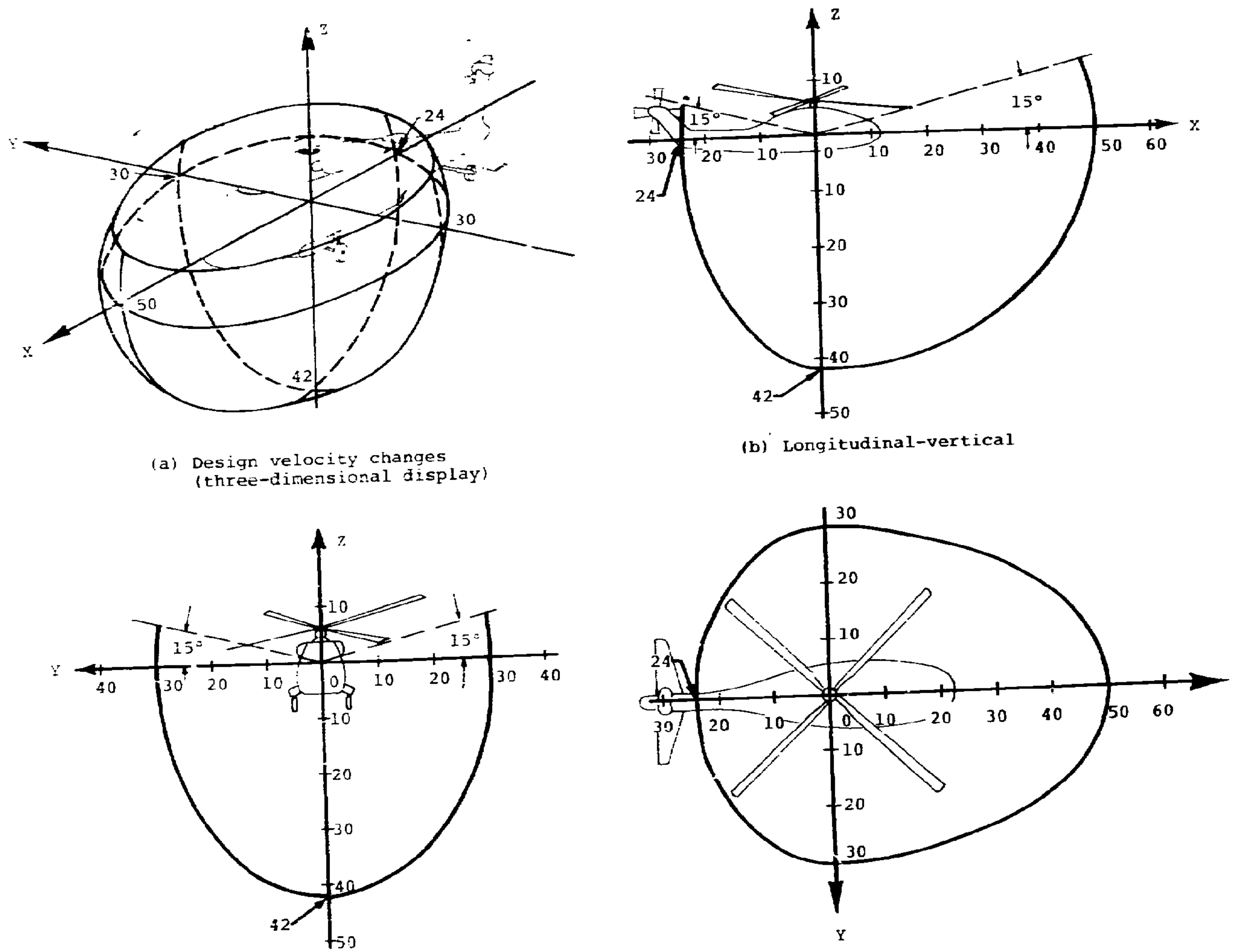
The accident survey produced insufficient data to plot an accurate distribution of lateral velocity changes. However, the engineering personnel participating in the study were able to infer from the circumstances of the accident cases studied that velocity changes in the lateral direction generally do not exceed 25 ft/sec for light fixed-wing aircraft and cargo and attack helicopters. Recent studies have indicated that 30 ft/sec is a more realistic number for other rotary-wing aircraft.

### 3.2.4 Combined Velocity Changes

The resultant velocity change for combined longitudinal, vertical, and lateral components of a 95th-percentile survivable accident of rotary- and light fixed-wing aircraft does not appear to exceed 50 ft/sec. The vertical or lateral components do not exceed their individual 95th-percentile values, i.e., 42 ft/sec vertically, and 30 and 25 ft/sec laterally for rotary- and light fixed-wing aircraft, respectively.

Figure 6 shows plots of combined longitudinal, lateral, and vertical velocity changes for helicopters, to be used in determining intermediate velocity change components. For light fixed-wing aircraft and for cargo and attack helicopters, Figure 6(b) will still be correct, but (c) and (d) must be altered for a lateral velocity change of 25 ft/sec instead of 30 ft/sec.

Since those accidents with the greatest longitudinal velocity change are not necessarily the same accidents with the greatest vertical velocity change, vertical and longitudinal percentiles shown in Figures 4 and 5 cannot be vector summed to give the combined velocity changes.




(a) Design velocity changes (three-dimensional display)

(b) Longitudinal-vertical

(c) Lateral-vertical

(d) Longitudinal-lateral

Figure 6. Design velocity changes - off-axis requirements.

In general, the components illustrated in Figure 6 are related by the equation

$$V_x^2 + V_y^2 + V_z^2 = V_R^2 \tag{1}$$

where $V_x$ = longitudinal velocity change, ft/sec

$V_y$ = lateral velocity change, ft/sec

$V_z$ = vertical velocity change, ft/sec

$V_R$ = resultant velocity change, ft/sec

and the axes are those illustrated in Figure 1. The curves have been terminated at 15 degrees, based on a study of accident reports discussed in Section 3.4. It was found that 95 percent of attack and cargo helicopter accidents occur at pitch angles between +25 degrees (nose up) and -15 degrees (nose down). Therefore, 15 degrees above the longitudinal axis appeared to be a reasonable point to which to extend the longitudinal-vertical conditions of Figure 6(b). For consistency, the 15-degree limit was extended to the other directions, where insufficient data existed.

As mentioned in Section 3.3, impact decelerations for design of the overhead structure have been determined. However, these conditions are intended for aircraft inverting after impact, not those impacting in an inverted attitude. Therefore, no design velocity changes are presented here for impact on the upper parts of the aircraft.

## 3.3 IMPACT ACCELERATIONS

Average aircraft floor decelerations were estimated by the following procedures:

- Calculations where deceleration distance (s) and velocity change (V) were known; i.e., the equation

$$G_{avg} = \frac{V^2}{2gs} \tag{2}$$

was used in those cases for which the velocity at the end of the pulse was zero.

- Comparisons of accident configurations and damage with crash test data.
- Observations of failure or nonfailure of lap belts, shoulder harnesses, and seats of known strength.
- Comparisons of injuries with generally established human tolerance limits.

### 3.3.1 Vertical Impact Accelerations

An examination of the frequency of occurrence of vertical accelerations for impacts of rotary- and light fixed-wing aircraft showed that 95 percent of survivable accidents involved average accelerations of less than 24 G. Peak accelerations can be expected to reach approximately twice the average values if triangular approximations to the crash pulses are assumed.

### 3.3.2 Longitudinal Impact Accelerations

Examination of the frequency of occurrence of longitudinal accelerations for impacts of rotary- and light fixed-wing aircraft showed that 95 percent of survivable accidents involved average accelerations of less than 15 G. Peak accelerations would be approximately twice the average values if triangular approximations to the crash pulses are assumed.

### 3.3.3 Lateral Impact Accelerations

Significant lateral accelerations were found to be present in the crashes studied, particularly in accidents where a rotary-wing aircraft autorotated into trees or where rotor blades struck trees or other obstacles during normal operation (Reference 3). Impact with trees often causes the fuselage to rotate and finally impact the ground on its side. Lateral accelerations also result in the types of secondary impacts associated with low-angle, high-velocity crashes and with nearly all severe accidents. The data on lateral accelerations obtained in the survey were not sufficient to allow construction of a distribution plot; however, it is believed that these accelerations rarely exceed 16 G for light fixed-wing aircraft and cargo and attack helicopters, although they may reach 18 G for other helicopters.

3. Gupta, B. P., HELICOPTER OBSTACLE STRIKE TOLERANCE CONCEPTS ANALYSIS, Bell Helicopter Textron; Technical Report 78-46, Applied Technology Laboratory, U. S. Army Research and Technology Laboratories (AVRADCOM), Fort Eustis, Virginia, April 1979, AD A069877.

## 3.4 IMPACT ATTITUDES

Analysis of data for survivable accidents involving attack and cargo helicopters from 1971 through 1976 resulted in the distributions of pitch and roll angles presented in Figures 7 and 8. The distribution of yaw angles appears much like the roll angle histogram of Figure 8, with 72 percent of the impacts occurring at zero or negligible yaw. Accidents reported as nonsurvivable were not included in the analysis, but they often occurred at extremes of attitude. In a number of nonsurvivable accidents, the attitude at impact was not reported. However, the number of these accidents with unknown attitude does not appear to constitute more than 10 percent of the total number of accidents. Therefore, the data indicate that existing attack and cargo helicopters are most likely to crash at a small roll angle.

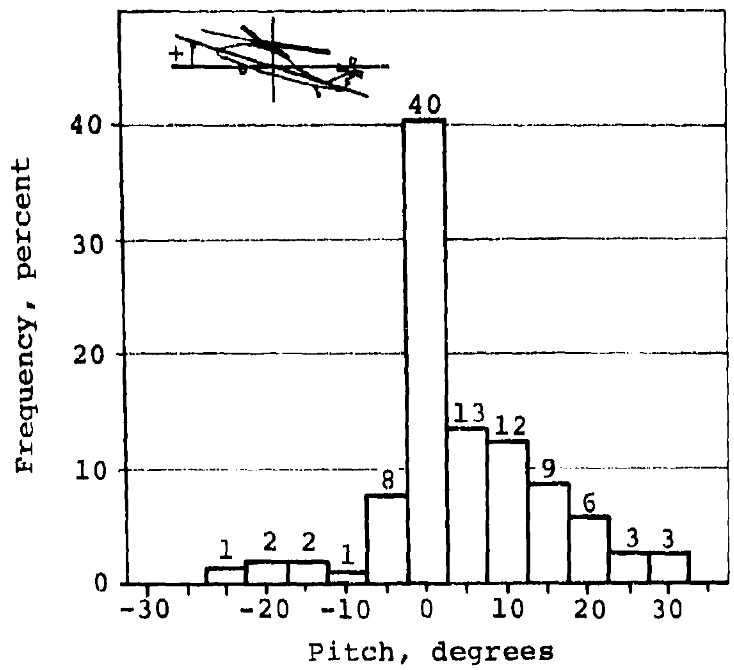



Figure 7. Frequency of occurrence of aircraft pitch angle at impact for survivable attack and cargo helicopter accidents, 1971-76.

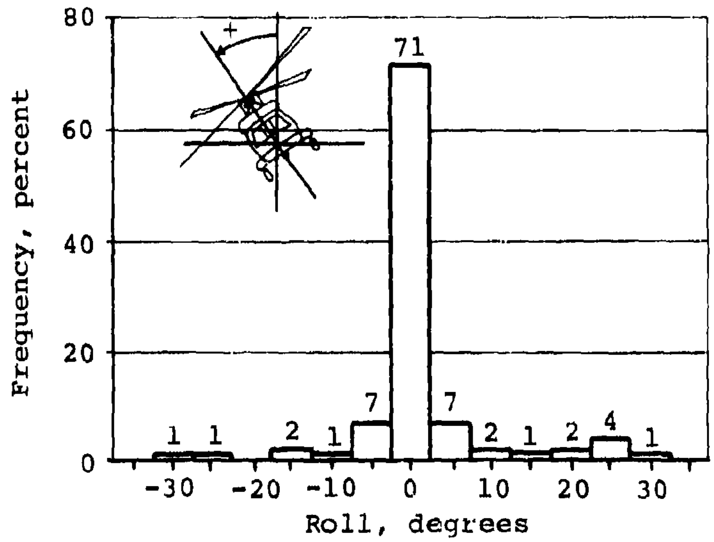



Figure 8. Frequency of occurrence of aircraft roll angle at impact for survivable attack and cargo helicopter accidents, 1971-76.

## 3.5 IMPACTED TERRAIN

The probability of impacting a given type of terrain may influence an aircraft design in several ways. For example, the types of landing gear and the escape systems installed may be based on the terrain most likely to be encountered in operation. A search of accident records at the U. S. Army Safety Center resulted in the distribution of impacted terrain shown in Table 1.

As discussed in Volume III, the lower fuselage stucture should be designed to minimize the effects of plowing or earth scooping during a longitudinal impact. For a primarily vertical impact, loose soil would prove beneficial through additional energy being absorbed by soil compaction. On the other hand, in a primarily horizontal (longitudinal and/or lateral) impact, loose soil may increase the deceleration level if the gear do not bend or fail or if the belly skin ruptures to cause an earth-scooping effect.

TABLE 1. DISTRIBUTION OF TERRAIN IMPACTED BY ARMY AIRCRAFT DURING THE PERIOD 1971 THROUGH 1976

| Terrain | Percentage of impacts |
|---|---|
| Sod | 43 |
| Trees | 30 |
| Rocks | 10 |
| Prepared surface | 7 |
| Bog | 6 |
| Water | 2 |
| Snow | 1 |
| Ice | 1 |

### 3.6 IMPACT INJURY FREQUENCIES

The data contained in this section were obtained from a search of accident records at the U. S. Army Safety Center, and are based on reports of accidents that involved a total of 4550 occupants during the period 1971 through 1976. The intention here is to inform the designer of aircraft systems as to the location of problem areas and the significance of the information contained in subsequent chapters, which deal with human tolerance to impact and the occupant environment.

Table 2 displays the frequencies of major and fatal injuries to various body parts as percentages of all major and fatal injuries reported. The injuries are broken down according to aircraft type in order to point out the effect of this variable where it is significant. It can be noted that the frequency of serious vertebral injuries is lower for light fixed-wing aircraft and cargo helicopters than for the other helicopter types. A reasonable explanation for this observation is that the vertical component of acceleration experienced by the occupants of light fixed-wing aircraft and cargo helicopters during impact is lower than for the other aircraft. As pointed out in Chapter 4, the vertical component of acceleration is likely to cause vertebral damage. Fixed-wing aircraft, often

TABLE 2. FREQUENCY OF MAJOR AND FATAL INJURIES TO EACH BODY PART AS PERCENTAGES OF TOTAL MAJOR AND FATAL INJURIES (U. S. ARMY AIRCRAFT, 1971 THROUGH 1976)

| | Percentage of injuries (total injuries in parentheses) | | | | | | |
|---|---|---|---|---|---|---|---|
| | Helicopters | | | | | | |
| Body part | Utility (678) | Observation (271) | Cargo (70) | Attack (93) | All helicopters combined (1114) | Light fixed-wing (104) | All aircraft combined (1218) |
| Head | 19.6 | 18.5 | 22.9 | 21.5 | 19.7 | 19.2 | 19.[illegible] |
| Face | 9.1 | 9.2 | 10.0 | 11.8 | 9.4 | 14.4 | [illegible].9 |
| Neck | 2.5 | 3.0 | 1.4 | 3.2 | 2.6 | 0.0 | 2.4 |
| Upper extremities | 10.9 | 12.9 | 17.1 | 15.1 | 12.1 | 11.5 | 12.1 |
| Thorax | 13.7 | 12.2 | 7.1 | 8.6 | 12.5 | 19.2 | 13.0 |
| Abdomen | 7.8 | 5.9 | 10.0 | 3.2 | 7.1 | 5.8 | 7.0 |
| Pelvis | 3.8 | 2.6 | 0.0 | 1.1 | 3.0 | 1.9 | 2.9 |
| Vertebrae | 15.3 | 19.2 | 8.6 | 21.5 | 16.5 | 12.5 | 16.2 |
| Lower extremities | 17.1 | 16.6 | 22.9 | 14.0 | 17.1 | 15.4 | 16.9 |

subject to a stall-spin type of crash, may have a lower frequency of flat impacts than helicopters. Cargo helicopters, being larger, have a greater crush distance beneath the floor that can attenuate the vertical impact forces.

The seriousness of head injuries is demonstrated by the data of Table 2 where they appear as the leading cause of major and fatal injuries. Their significance is further amplified by the data of Table 3, where head injuries are shown to account for approximately 31 percent of all fatal injuries, regardless of aircraft type.

The information presented in Tables 2 and 3 is summarized in Figure 9.

The influence of rollover conditions on the incidence of fatalities was studied in 198 major Army aircraft accidents during 1970-71. The aircraft studied were utility, attack, observation, and cargo helicopters. The accident data indicated that the probability of a major accident resulting in fatal injuries increases if a rollover occurs. Of 104 rollover accidents, 51 accidents, or 49 percent, resulted in fatalities; of 94 no-roll accidents, 30 accidents, or 32 percent, resulted in fatalities (Reference 4).

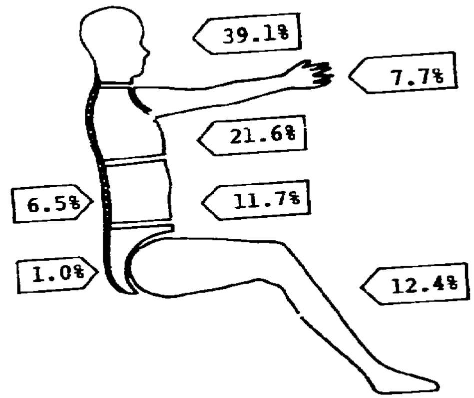
## 3.7 ACCIDENT INFORMATION RETRIEVAL SYSTEM

Crash survival design criteria are based on the realities of actual aircraft accidents. It is necessary to know such facts as aircraft attitude and velocity at impact, and the magnitude, duration, and direction of the impact forces. These data are currently derived by accident investigation teams who must estimate the impact conditions from their study of the wreckage and terrain, witness accounts, medical reports, etc. These same teams also must try to reconstruct preimpact events and determine possible primary and contributing cause factors so that preventive measures can be taken to decrease the incidence of aircraft accidents. Such accident reconstruction, even when performed by the most skilled investigators, can produce no more than estimated numbers and probable causes. Accurate aircraft precrash and crash data are needed to establish a sound basis for more precise design criteria and operational techniques to reduce accidents and improve crash survivability.

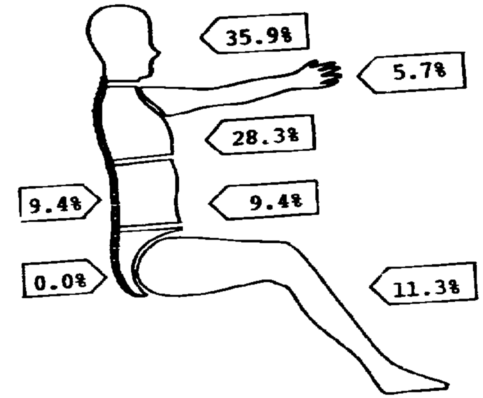
4. Haley, J. L., and Hicks, J. E., CRASHWORTHINESS VERSUS COST: A STUDY OF ARMY ROTARY-WING AIRCRAFT ACCIDENTS IN PERIOD JAN 70 THROUGH DEC 71, paper presented at Aircraft Crashworthiness Symposium, University of Cincinnati, 6 - 8 October 1975.

TABLE 3. FREQUENCY OF FATAL INJURIES TO EACH BODY PART AS PERCENTAGES OF ALL FATAL INJURIES (U. S. ARMY AIRCRAFT, 1971 THROUGH 1976)

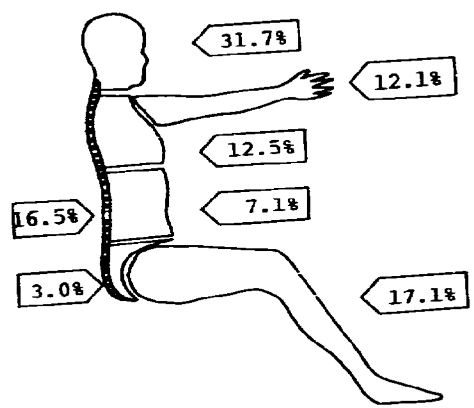
| Body part | Percentage of injuries (total injuries in parentheses) | | | | | | |
|---|---|---|---|---|---|---|---|
| | Helicopters | | | | | | |
| | Utility (262) | Observation (75) | Cargo (34) | Attack (32) | All helicopters combined (40[illegible]) | Light fixed-wing (53) | All aircraft combined (456) |
| Head | 32.4 | 25.3 | 35.3 | 34.4 | 31.5 | 30.2 | 31.4 |
| Face | 4.6 | 4.0 | 8.8 | 6.3 | 5.0 | 5.7 | 5.0 |
| Neck | 1.9 | 6.7 | 0.0 | 3.1 | 2.7 | 0.0 | 2.4 |
| Upper extremities | 7.3 | 6.7 | 8.8 | 12.5 | 7.7 | 5.7 | 7.4 |
| Thorax | 23.3 | 22.7 | 8.8 | 18.8 | 21.6 | 28.3 | 22.4 |
| Abdomen | 11.8 | 12.0 | 14.7 | 6.3 | 11.7 | 9.4 | 11.4 |
| Pelvis | 0.8 | 2.7 | 0.0 | 0.0 | 1.0 | 0.0 | 0.9 |
| Vertebrae | 7.3 | 6.7 | 5.9 | 0.0 | 6.5 | 9.4 | 6.8 |
| Lower extremities | 10.7 | 13.3 | 17.6 | 18.8 | 12.4 | 11.3 | 12.3 |



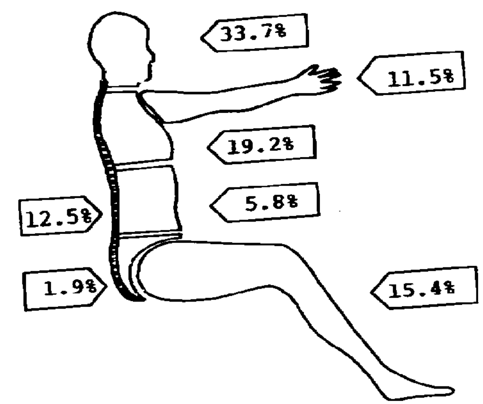

a. Frequency of fatal injuries, helicopters.




c. Frequency of fatal injuries, light fixed-wing.




b. Frequency of major and fatal injuries, helicopters.




d. Frequency of major and fatal injuries, light fixed-wing.

Figure 9. Frequency of injuries to body parts in U. S. Army aircraft accidents, 1971-76.

To answer this need, the Applied Technology Laboratory of the U. S. Army Research and Technology Laboratories (AVRADCOM) developed a program and awarded a contract for the conceptual analysis and preliminary design of an Accident Information Retrieval System (AIRS) suitable for helicopter installation. As part of this effort, present state-of-the-art electromechanical crash recorders were examined and found to be both too large and too heavy for practical helicopter installation. Results of the study, documented in Reference 5, indicate that a solid-state system of approximately 10 lb in weight and 200 in.$^3$ volume is feasible for installation in new Army aircraft, such as UTTAS or AAH. The further development and eventual installation of the AIRS would measurably reduce the cost and time of accident investigation while enhancing the validity of the accident data obtained. The data that such a system would provide should be of great value in further refining crashworthy design criteria.

5. Burrows, L. T., AN ECONOMICAL APPROACH TO AN ACCIDENT INFORMATION RETRIEVAL SYSTEM (AIRS), Applied Technology Laboratory, U. S. Army Research and Technology Laboratories (AVRADCOM); paper presented at the 15th Annual Survival and Flight Equipment Symposium, Las Vegas, Nevada, December 1977.

# 4. HUMAN TOLERANCE TO IMPACT

## 4.1 INTRODUCTION

The objective of this chapter is to provide the designer of aircraft systems with a summary of available information on tolerance of the human body to forces of the type experienced in a crash environment. Knowledge of human tolerance is vital to understanding and effectively applying the principles of crashworthy design that are defined elsewhere in this guide.

A great deal of research has been conducted in the field of biodynamics, and general guidelines and approximate end points have been determined. However, there still remain many areas of uncertainty and disagreement, and much more research is needed to provide accurate, proven figures. An obvious difficulty is that the usual test method of stressing a specimen beyond the point of failure in order to establish tolerance limits is not possible when one is dealing with injurious and fatal ranges of forces acting on the human body. Experiments using human volunteers have necessarily been conducted, for the most part, at subcritical levels. The few instances that have inadvertently approached the critical range of forces have provided valuable, but often unverifiable, data. Test animals such as chimpanzees, monkeys, bears, pigs, and mice have been used in attempts to establish a better definition of the injurious and fatal ranges of forces. Human cadavers also have been utilized as test specimens. While these approaches have provided valuable data in many areas of investigation, a means of reliably extrapolating these results to the tolerance of a live human is not yet available.

As discussed in Volume IV, mathematical models of the human body have been used successfully in studying the overall kinematic response of the body to crash forces and in evaluating crashworthiness of vehicle interiors. Mathematical models directly related to injury prediction are discussed in this chapter. Anthropomorphic dummies, as mechanical models, have been refined to remarkable levels of physical resemblance to the human body. However, interpreting the results from mathematical and mechanical simulators remains a problem, as tolerable levels of the predicted variables are neither well defined nor widely agreed upon.

The following sections discuss the factors that affect human tolerance to impact and summarize the existing data on tolerable levels for various body parts. An in-depth review of research in human tolerance before 1970 can be found in Reference 6.

## 4.2 FACTORS AFFECTING HUMAN TOLERANCE

The tolerance of the human body to impact forces depends on a number of variables, including characteristics of the individual such as age, sex, and general state of health. Military systems can be expected to be used by personnel who are generally younger and in better physical condition than the general population for which much tolerance data has been obtained. Thus, in some cases, a degree of conservatism may be built into the application of tolerance criteria in designing Army aircraft. However, whole-body tolerance criteria have been based on experiments involving subjects seated with "correct" upright posture. Because a helicopter pilot is unlikely to maintain such posture in flight, particularly when near the ground, tolerable levels of such variables as $+G_z$ acceleration may be significantly reduced under actual crash conditions.

The overall probability of survival in a crash depends to a large extent on the manner of restraint. It would be extremely difficult to prevent the arms and legs from contacting the cabin interior during a severe impact, but the use of upper and lower torso restraints to prevent such critical body parts as the head and chest from striking surrounding structure can significantly reduce the probability of serious or fatal injury under given crash conditions.

The method of body restraint, of all the factors affecting human tolerance, offers the designer the greatest opportunity for effective application of crashworthy design. The effectiveness of the restraint system is dependent upon the area over which the total force is distributed, the location on the body at which the restraint is applied, and the degree to which it limits residual freedom of movement (Reference 7).

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6. Snyder, R. G., HUMAN IMPACT TOLERANCE, Paper 700398, International Automobile Safety Compendium, Society of Automotive Engineers, New York, 1970, pp. 712-782.

7. Rothe, V. E., et al., CREW SEAT DESIGN CRITERIA FOR ARMY AIRCRAFT, TRECOM Technical Report 63-4, U. S. Army Transportation Research Command, Fort Eustis, Virginia, February 1963.

The greater the contact area between the body and the restraint system, the greater the human tolerance. The restraint system should be located on the body at those points that are best able to withstand the loads exerted by the decelerative force and that are best able to further distribute the force to the remainder of the body. These points are primarily the pelvic girdle and the shoulder structure. An additional restraint around the rib cage has been shown to increase tolerance to spineward, eyeballs out ($-G_x$) accelerations. Restraint systems located over soft tissue tend to be much less effective, often resulting in crushing of the viscera between the restraint system and bony structures. Residual freedom of movement should be limited to an absolute minimum consistent with the necessary comfort and movements required by the duties of the occupant.

When restrained only by a lap belt, the occupant's tolerance to abrupt acceleration is relatively low. In forward-facing seats, a longitudinal impact will cause a rotation of the upper torso over the belt, a whipping action of the head, and often, impact of the upper torso on the legs, resulting in chest, head, and neck injuries. Head injuries due to impacts with the surrounding environment are very common for occupants restrained only with lap belts. When longitudinal forces are combined with a vertical component, there is a tendency for the occupant to slip under the belt to some degree. This motion, often referred to as submarining, can shift the belt up over the abdomen. The longitudinal component of the pulse then causes the upper torso to flex over the belt, with the restraining force concentrated at some point on the spine and not on the pelvic girdle. In this configuration, tolerance is extremely low.

The addition of a shoulder harness greatly reduces injuries from head impacts and helps to maintain proper spinal alignment for strictly vertical impact forces. However, this standard configuration may be unsatisfactory for impacts with both vertical and longitudinal components. Pressure by the upper torso against the shoulder straps causes these straps to pull the lap belt up into the abdomen and against the lower margin of the rib cage. This movement of the lap belt allows the pelvis to move forward under the lap belt, causing severe flexing of the spinal column, as shown in Figure 10. In this flexed position, the vertebrae are very susceptible to anterior compression fractures and, if the lap belt slips off the top of the pelvic bone structure (over the top of the iliac crests), severe injury can occur as a result of viscera crushing. A lap belt tiedown strap prevents raising of the lap belt by the shoulder harness and may nearly double the tolerance to impact forces.

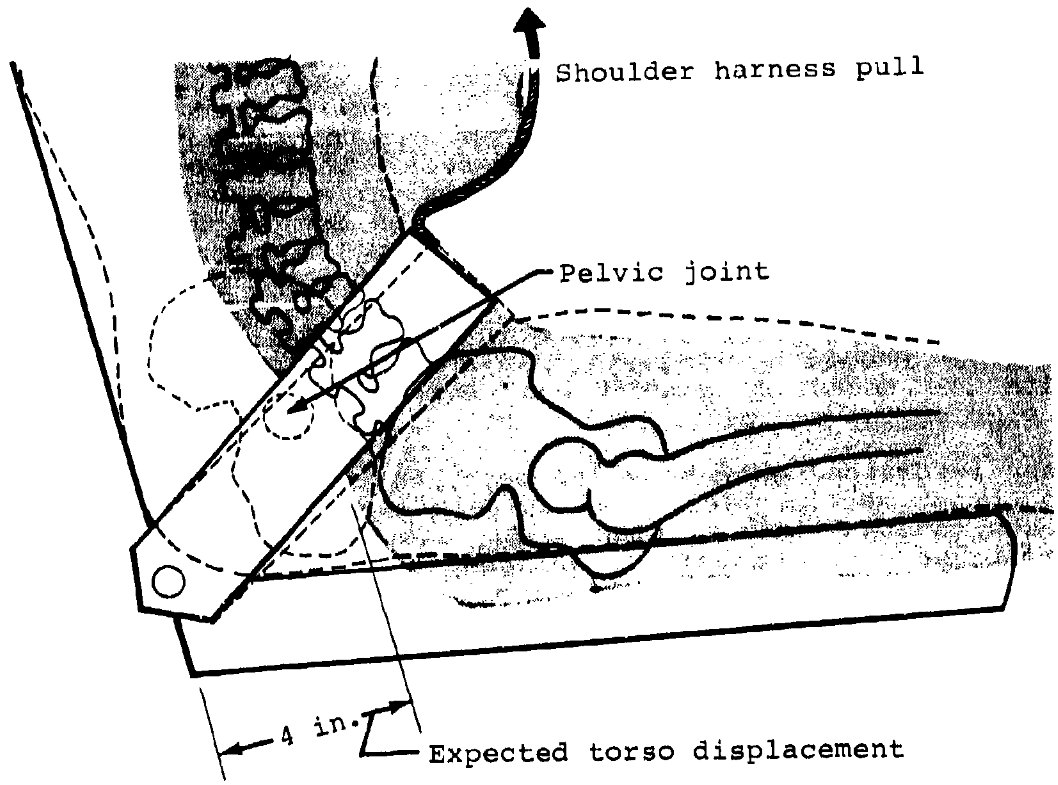


Figure 10. Pelvic rotation and submarining caused by high longitudinal forces combined with moderate vertical forces.

The restraint system used in new Army aircraft crewseats, as defined by MIL-S-58095(AV), Reference 8, consists of a lap belt, lap belt tiedown strap, and two shoulder straps connected by a single-point release buckle.

The amount of slack in the restraint system can affect tolerance to a given acceleration pulse. In general, the lower the elongation properties of the link between the occupant and the

8. Military Specification, MIL-S-58095(AV), SEAT SYSTEM: CRASHWORTHY, NON-EJECTION, AIRCREW, GENERAL SPECIFICATION FOR, Department of Defense, Washington, D. C., 27 August 1971.

seat, the greater the occupant's tolerance to an abrupt acceleration. A loose restraint system also will result in the occupant's receiving a significantly greater magnification of the accelerative force applied to the seat than would occur with a snug system. The inertia of the occupant will cause him to maintain a near constant velocity, independent of the decreasing velocity of the seat, until the slack in the restraint system is taken up. As this point is reached, the velocity of the occupant is abruptly reduced to that of the seat, resulting in relatively high G levels, even exceeding those of the seat. This is often referred to as dynamic overshoot. Dynamic overshoot is a complex phenomenon involving the elasticity, geometry, mass distribution, and, thus, the natural frequency of the occupant, and the restraint and seat systems. An example is discussed in detail in Volume IV.

## 4.3 WHOLE-BODY ACCELERATION TOLERANCE

The paragraphs in this section describe experimental results applicable to acceleration of well-restrained (including full-torso restraint) seated occupants. The tolerance levels presented here define for the designer the limits of the environment that must be provided to enable survival of aircraft occupants in a crash.

### 4.3.1 Spineward ($-G_x$) Acceleration

The magnitude and duration of the applied accelerative force have definite effects on human tolerance, as shown in Figure 11 (Reference 9). As indicated by this curve, a spineward chest-to-back accelerative force of 45 G has been tolerated voluntarily by some subjects when the pulse duration is less than 0.044 sec. Under similar conditions, when the duration is increased to 0.2 sec, the tolerable magnitude is reduced to about 25 G. Accordingly, Figure 11 shows that the tolerable limits on acceleration loading are a function of duration.

The whole-body tolerance data displayed in Figure 11 were collected for a variety of full-torso restraint and, in some cases, head restraint. With less optimum restraint, some debilitation and injury will occur at this acceleration level, or, in other words, the tolerable level will be significantly reduced.

9. Eiband, A. M., HUMAN TOLERANCE TO RAPIDLY APPLIED ACCELERATIONS: A SUMMARY OF THE LITERATURE, NASA Memorandum 5-19-59E, National Aeronautics and Space Administration, Washington, D. C., June 1959.

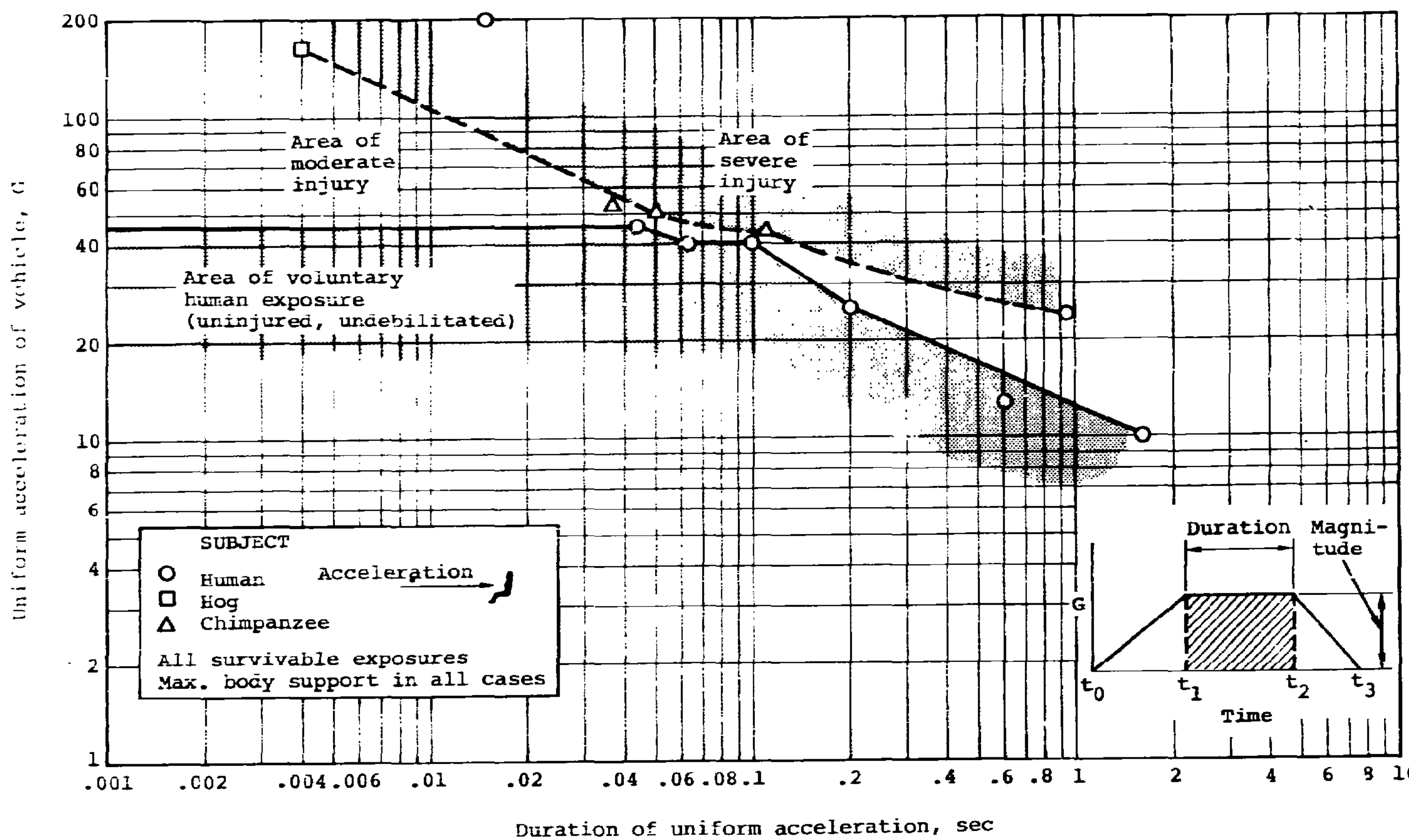


Figure 11. Duration and magnitude of spineward acceleration endured by various subjects. (From Reference 9)

With respect to whole-body deceleration, the rate of onset of the applied force also has a definite, although not yet well understood, effect on human tolerance. Under some impact conditions, the rate of onset appears to be a determining factor, as indicated by the diagram in Figure 12 (Reference 9). Lower rates of onset were more tolerable than higher rates under the test conditions present. Under other impact conditions, such as extremely short durations that occur in impacts from free falls, rates of onset as high as 28,000 G/sec were survivable and appeared to have little effect on human tolerance (Reference 10). It appears that, in certain ranges, the effects of the rate of onset are related to the natural frequencies of the body and of the various body organs (Reference 11).

### 4.3.2 Sternumward ($+G_x$) Acceleration

The human tolerance limit for sternumward, eyeballs-in ($+G_x$) acceleration has not been accurately established. Due to the high degree of restraint provided by a full-length seat back in this configuration, it can be safely assumed that tolerance is greater than for spineward acceleration. A maximum of 83 G measured on the chest with a base duration of 0.04 sec was experienced on one run in a backward-facing seat. However, the subject was extremely debilitated, went into shock following the test, and required on-the-scene medical treatment (Reference 12). Human tolerance to sternumward acceleration, therefore, probably falls somewhere between this figure of 83 G for 0.04 sec and 45 G for 0.1 sec, which is the accepted end point for the $-G_x$ (eyeballs-out) case. As in the case of $+G_x$ accelerations, the human subject experiments employed head restraint so that under aircraft operational conditions tolerable levels would be lower.

### 4.3.3 Headward ($+G_z$) Acceleration

The human body is able to withstand a much greater force when the force is applied perpendicular to the long axis of the

10. Snyder, R. G., HUMAN TOLERANCE TO EXTREME IMPACTS IN FREE FALL, Aerospace Medicine, Vol. 34, No. 8, August 1963, pp. 695-709.

11. Stapp, J. P., JOLT EFFECTS OF IMPACT ON MAN, Brooks Air Force Base, San Antonio, Texas, November 1961.

12. Beeding, E. L., Jr., and Mosely, J. D., HUMAN DECELERATION TESTS, Air Force Missile Development Center; AFMDC Technical Note 60-2, Holloman Air Force Base, New Mexico, January 1960, AD 23148.

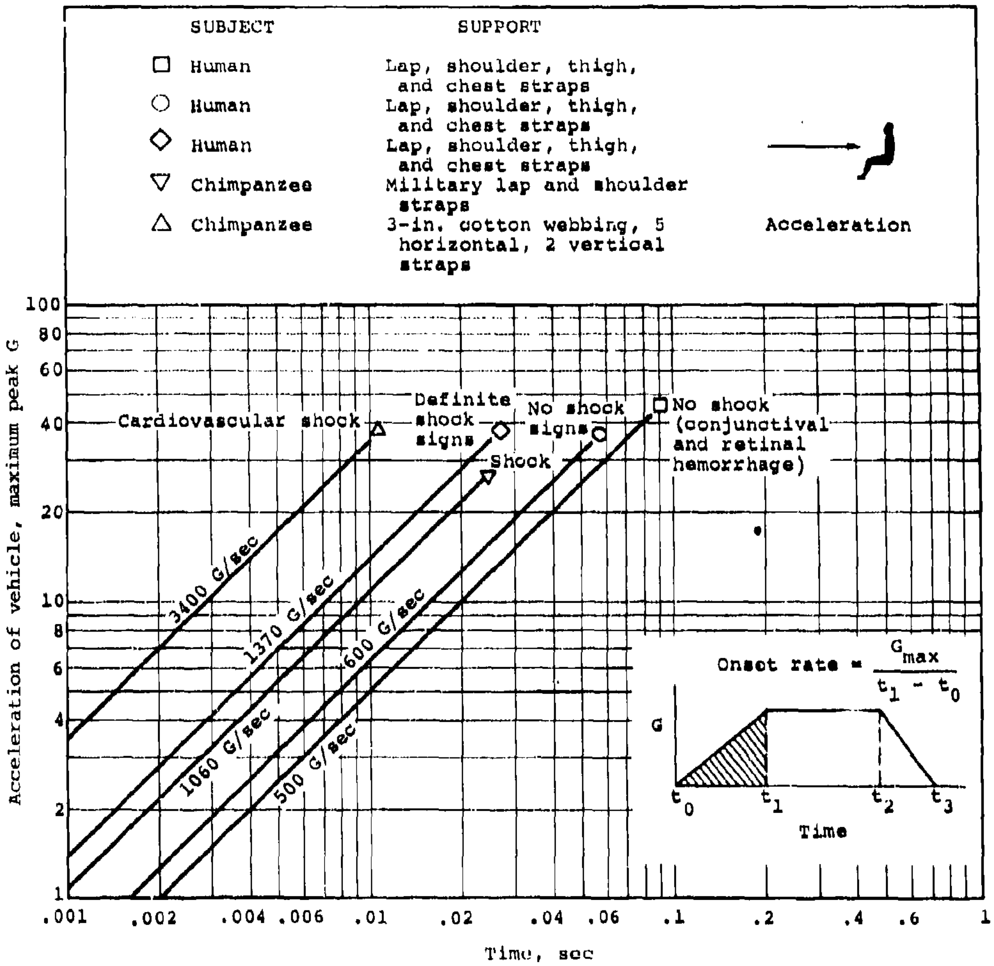


Figure 12. Initial rate of change of spineward acceleration endured by various subjects. (From Reference 9)

body in a forward or backward direction ($G_x$) than when applied parallel to the long axis ($G_z$). This is shown by a comparison of the curves in Figures 11 and 13. A primary reason for the significantly lower tolerance to headward ($+G_z$) loading is the susceptibility of the lumbar vertebrae, which must support most of the upper torso load, to compression fracture. Also, the skeletal configuration and mass distribution of the body are such that vertical loads cannot be distributed over as large an area as can loads applied forward or backward ($G_x$).

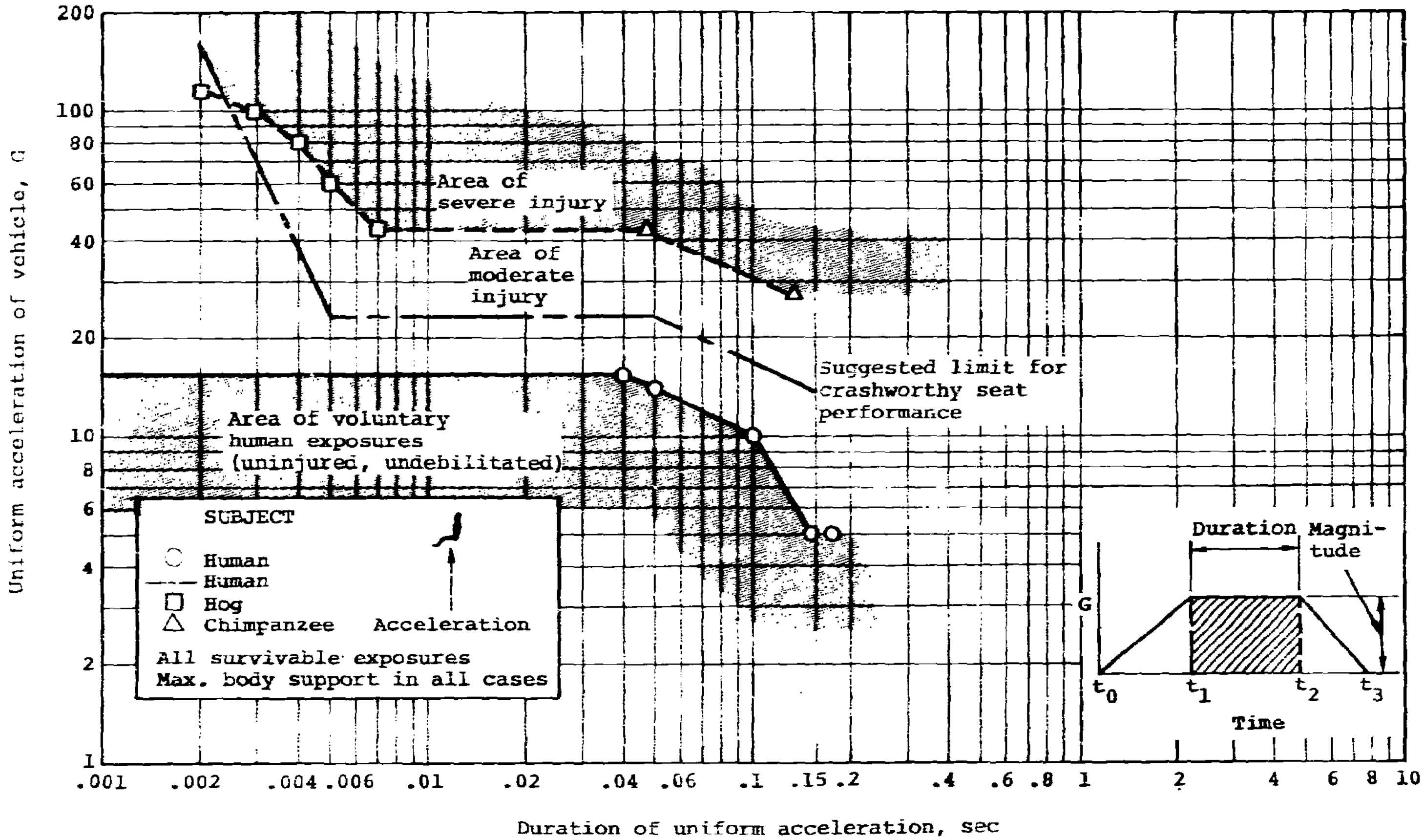


Figure 13. Duration and magnitude of headward acceleration endured by various subjects. (From Reference 9)

These vertical loads, therefore, result in greater stress per unit area than do sternumward or spineward loads. Finally, along the direction of the long axis, the body configuration allows for greater displacement of the viscera within the body cavity. Forces applied parallel to the long axis of the body, headward or tailward ($G_z$), place a greater strain on the suspension system of the viscera than do forces applied sternumward or spineward ($G_x$), thereby increasing the susceptibility of the viscera to injuries.

As in the case of the longitudinal direction (Figure 12), rate of onset also affects tolerance to vertical accelerative loads; however, insufficient data were available to establish the limits. (Figure 14 presents one set of available data.)

#### 4.3.4 Tailward ($-G_z$) Acceleration

The human tolerance limit for tailward, eyeballs-up ($-G_z$) acceleration is approximately 15 G for a duration of 0.1 sec. The shoulder harness/lap belt restraint has been used in all human testing with tailward accelerations. Most experiments also have included a lap belt tiedown strap, and the 15 G tolerance limit is based on this latter configuration.

#### 4.3.5 Lateral ($G_y$) Acceleration

Very little research has been conducted on human tolerance to lateral ($G_y$) accelerations. Two studies, one involving restraint by a lap belt alone (Reference 13) and another involving restraint by the lap belt/shoulder harness configuration (Reference 14), provide the principal available data. In both cases, a side panel provided additional restraint. With restraint by the lap belt alone, volunteers were able to withstand a pulse with an average peak of approximately 9 G for a duration of approximately 0.1 sec. At this level, the tests were discontinued due to increasing concern about lateral spinal flexion. In the experiments with restraint by lap belt and shoulder harness, volunteers were able to withstand a pulse with an average acceleration of approximately 11.5 G for

13. Zaborowski, A. V., HUMAN TOLERANCE TO LATERAL IMPACT WITH LAP BELT ONLY, Proceedings, Eighth Stapp Car Crash and Field Demonstration Conference, Society of Automotive Engineers, New York, 1964.

14. Zaborowski, A. V., LATERAL IMPACT STUDIES - LAP BELT SHOULDER HARNESS INVESTIGATIONS, Proceedings, Ninth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1965.

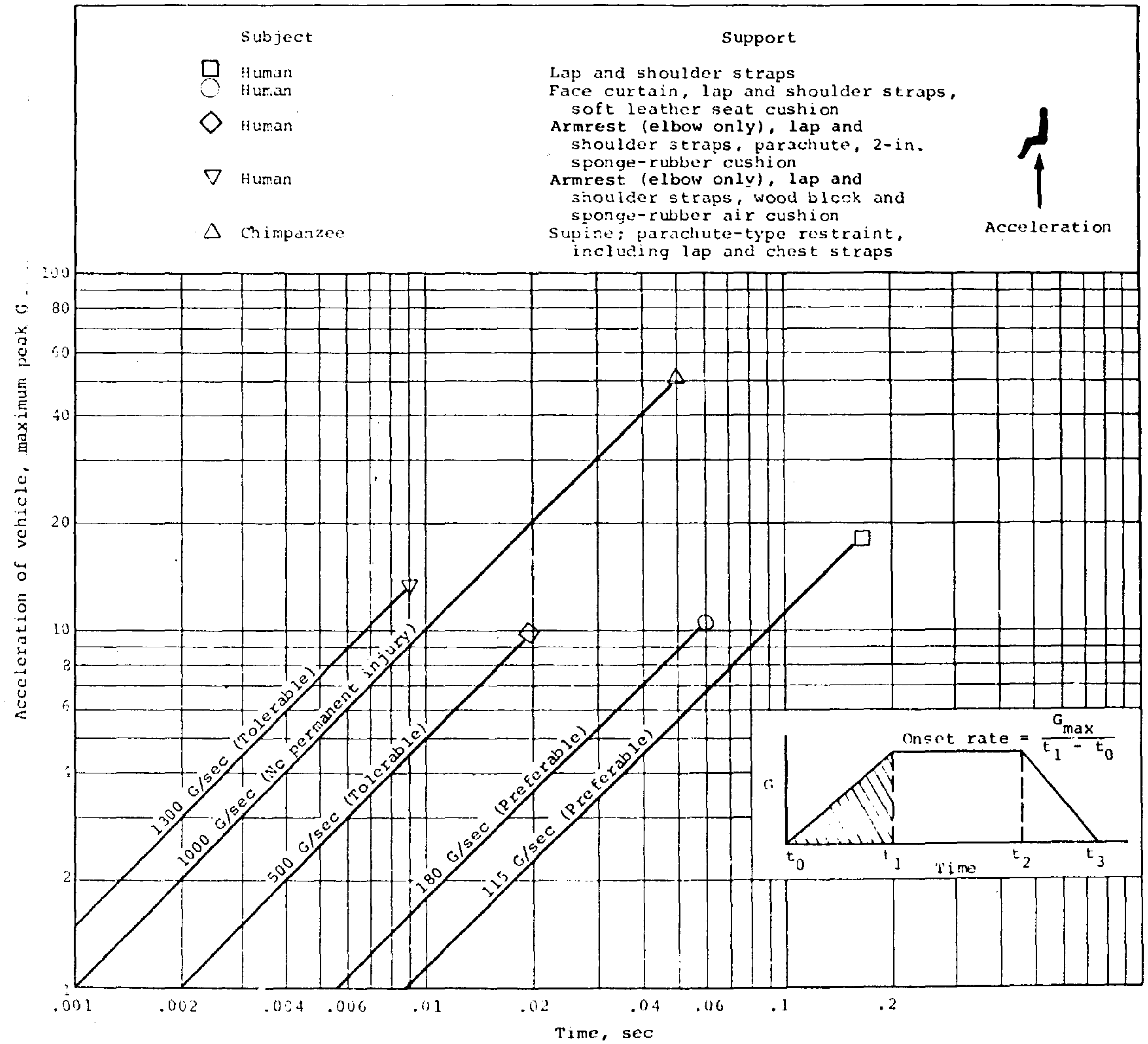


Figure 14. Initial rate of change of headward acceleration endured by various subjects. (From Reference 9)

a duration of approximately 0.1 sec with no permanent physiological changes. Tests were discontinued at this level due to possible cardiovascular involvement experienced by one of the two subjects tested. No end points for human tolerance to lateral impacts were proposed in the reports of these experiments. The only reasonable conclusions from these data at this time are that a pulse of 11.5 G with a duration of 0.1 sec is readily sustained by subjects restrained by a lap belt and shoulder harness and that the human survival limit is at some point beyond this level, probably at least 20 G for 0.1 sec.

The above values are supported by a series of human volunteer experiments conducted to measure the inertial response of the head and neck to $+G_y$ whole-body acceleration (Reference 15). Acceleration inputs ranged from long-duration pulses with magnitudes of 2 to 7.5 G to short-duration pulses of 5 to 11 G.

## 4.4 HEAD IMPACT TOLERANCE

As indicated by the accident data discussed in Chapter 3, over 30 percent of the Army aircraft crash fatalities result from head injuries. The injuries may result either from impact of the head on some aircraft structure or equipment, or from head acceleration without impact. In the case of mechanical impact, tolerance conditions often are based on the presence or absence of skull fracture. However, concussion can result from nonimpact motion of the head, whether in flexion or hyperextension (References 16 and 17). Concussion that may be nonfatal in itself can temporarily immobilize an individual and reduce his chances of survival by subjecting him to postcrash hazards such as fire or drowning.

Fatal head injury has been shown to result from such severe brain damage as laceration of brain tissue or shear of the

15. Ewing, C. L., et al., DYNAMIC RESPONSE OF THE HUMAN HEAD AND NECK TO $+G_y$ IMPACT ACCELERATION, Proceedings, Twenty-First Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1977, pp. 547-586.

16. Hollister, N. R., et al., BIOPHYSICS OF CONCUSSION, Wright Air Development Center; WADC Technical Report 58-192, Wright-Patterson Air Force Base, Ohio, 1958.

17. Ommaya, A. L., and Hirsch, A. E., TOLERANCES FOR CEREBRAL CONCUSSION FROM HEAD IMPACT AND WHIPLASH IN PRIMATES, Journal of Biomechanics, Vol. 4, 1971, pp. 13-21.

brain stem (Reference 18). Head impact studies with anesthetized monkeys and dogs have been conducted to relate the severity and duration of concussion to intracranial pressure change (Reference 19). Moderate to severe concussion effects were observed in the range of 30 to 90 $lb/in.^2$ intracranial pressure change concurrent with head impact.

According to an hypothesis developed by Holburn, shear stresses induced by head rotation also can produce concussion (Reference 20). Kornhauser, in Reference 21, indicated a relationship between damaging velocity and damaging accelerations as follows:

$$\dot{\theta} = \frac{\ddot{\theta}}{\omega} \tag{3}$$

where $\dot{\theta}$ = damaging rotational velocity, rad/sec

$\ddot{\theta}$ = damaging rotational acceleration, $rad/sec^2$

$\omega$ = natural frequency of rotation of brain, rad/sec

Ommaya, et al., developed scaling factors needed to predict concussion thresholds for man from data taken on subhuman primates (Reference 22). This study showed that $\ddot{\theta}$ can be represented by

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18. Goldsmith, W., BIOMECHANICS OF HEAD INJURY, In Biomechanics-Its Foundation and Objectives, ed. by Fung, Y. C., Perrone, N., and Anliker, M., Prentice-Hall, Inc., Englewood Cliffs, New Jersey, 1972, pp. 585-634.

19. Gurdjian, E. S., Roberts, V. L., and Thomas, L. M., TOLERANCE CURVES OF ACCELERATION AND INTRACRANIAL PRESSURE AND PROTECTIVE INDEX IN EXPERIMENTAL HEAD INJURY, Journal of Trauma, Vol. 6, 1966, pp. 600-604.

20. Holburn, A. H. S., MECHANICS OF BRAIN INJURIES, British Medical Bulletin, Vol. 3(6) 1945, pp. 147-149.

21. Kornhauser, M., PREDICTION AND EVALUATION OF SENSITIVITY TO TRANSIENT ACCELERATION, Journal of Applied Mechanics, Vol. 21, 1945, p. 371.

22. Ommaya, A. K., et al., SCALING OF EXPERIMENTAL DATA ON CEREBRAL CONCUSSION IN SUB-HUMAN PRIMATES TO CONCUSSION THRESHOLD FOR MAN, Proceedings, Eleventh Stapp Car Crash Conference, Society of Automotive Engineers, New York, October 10-11, 1967.

the equation

$$\ddot{\theta} = \frac{c}{m^{2/3}} \tag{4}$$

where m = mass of the brain, gm

c = an experimentally derived constant, $gm^{2/3}$ rad/sec$^2$

The investigators found c = 21,600 $gm^{2/3}$ rad/sec$^2$, and further showed that the relationship of Equation (3) produced reasonable agreement between predictions and empirical data. Limiting values thus predicted to produce a 50-percent probability of concussion in a man having a brain mass of 1300 gm are as follows (Reference 23):

$$\ddot{\theta} = 1800 \text{ rad/sec}^2$$

$$\dot{\theta} = 50 \text{ rad/sec}$$

In general, assessing the probability of injury by observation of oversimplified parameters, such as peak acceleration of the imposed acceleration-time environment, usually is not constructive. The problem has been to define some form of parameter that is indicative of the degree of severity of a particular input excitation. Various indicators have been developed, based on experiments, and several of these for the head are presented and discussed in Sections 4.4.1 through 4.4.5.

### 4.4.1 Weighted Impulse Criterion (Severity Index)

It can be seen from human tolerance data presented previously that high forces or accelerations can be tolerated for only very short periods of time, while lower values of these quantities can be tolerated for longer periods of time. This same relationship for head injury in forehead impacts, which was established on the basis of impact tests performed at Wayne State University on animals and human cadavers, is illustrated

23. Ommaya, A. K., et al., COMPARATIVE TOLERANCE FOR CEREBRAL CONCUSSION BY HEAD IMPACT AND WHIPLASH INJURY IN PRIMATES, 1970 International Automobile Safety Conference Compendium, Society of Automotive Engineers, New York, 1970.

in Figure 15 (Reference 24). In these tests, longitudinal impacts of the subject's forehead against unyielding flat surfaces were conducted, and the acceleration-time history of the specimen head was measured at a point on the skull diametrically opposite the point of impact. The curve shown in Figure 15 was based on the observation of linear skull fracture.

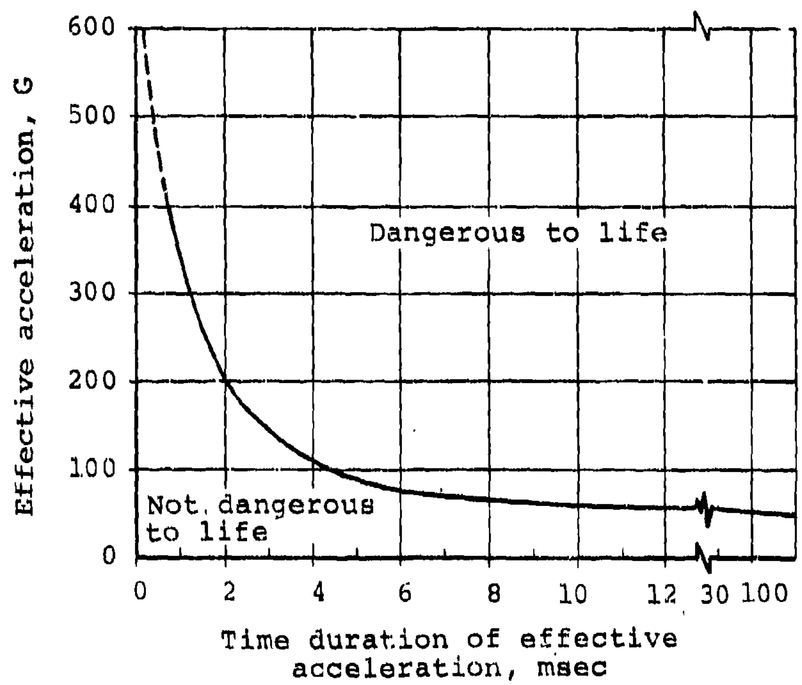


Figure 15. Wayne State Tolerance Curve for the human brain in forehead impacts against plane, unyielding surfaces. (From Reference 24)

Based on such data as that collected at Wayne State, Gadd suggested a weighted impulse criterion as an evaluator of injury potential (Reference 25). This severity index is defined as

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24. Patrick, L. M., Lissner, H. R., and Gurdjian, E. S., SURVIVAL BY DESIGN - HEAD PROTECTION, Proceedings, Seventh Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1963.

25. Gadd, C. W., USE OF A WEIGHTED-IMPULSE CRITERION FOR ESTIMATING INJURY HAZARD, Proceedings, Tenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1966.

$$SI = \int_{t_o}^{t_s} a^n dt \tag{5}$$

where SI = severity index

a = acceleration as function of time

n = weighting factor greater than 1

t = time

Based on data in References 9 and 24, the exponent n has been determined to be 2.5 for head and facial impacts. For the load, Reference 26 suggests tolerable SI levels of 1000 for distributed loading and 400 for localized loading. Severity index values exceeding 600 produced concussion in load impacts sustained by U. S. Army aircrewmen in aircraft accidents (Reference 24). A lower value of n also has been suggested for regions of softer tissue, which behave viscoelastically.

The severity index can be calculated by dividing the time base of the acceleration time curve into sufficient segments to define the acceleration curve. The G value then read from the curve for the center of the increment is raised to the 2.5 power, and the result is multiplied by the time increment. The sum of all the values obtained gives the severity index. A severity index sample calculation is shown in Figure 16, which is taken from Reference 26.

### 4.4.2 J-Tolerance

Slattenschek, after noting different head deceleration curves when different types of windshield glass were used, developed a method of assessing multiple impact tolerance by a "J-tolerance" value (Reference 27). A second-order vibrational

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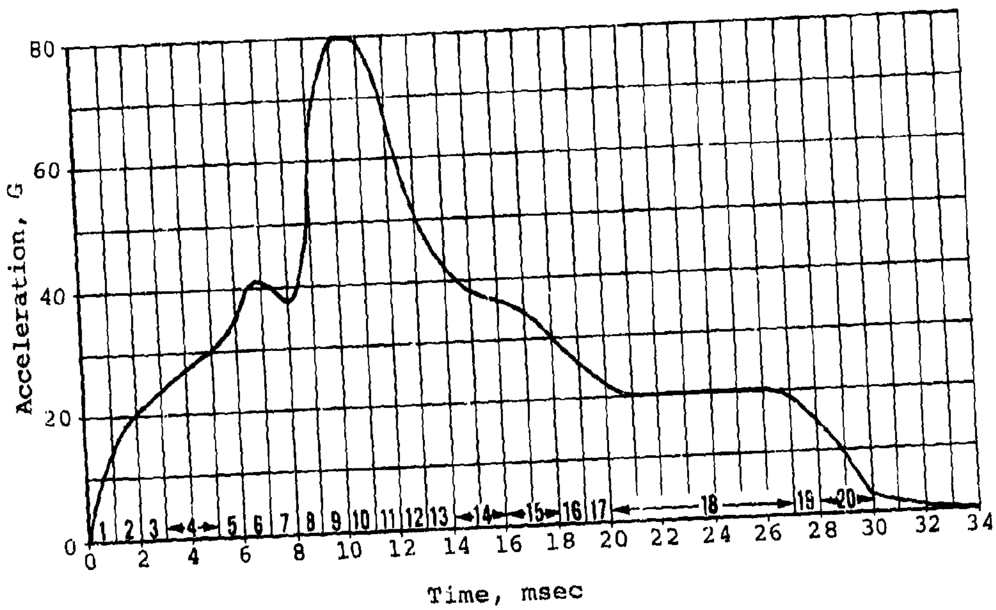
26. SAE Information Report, HUMAN TOLERANCE TO IMPACT CONDITIONS AS RELATED TO MOTOR VEHICLE DESIGN - SAE J885a, SAE Handbook, Part 2, Society of Automotive Engineers, Warrendale, Pennsylvania, 1979, pp. 34.114-34.117.

27. Slattenschek, A., and Tauffkirchen, W., CRITICAL EVALUATION OF ASSESSMENT METHODS FOR HEAD IMPACT APPLIED IN APPRAISAL OF BRAIN INJURY HAZARD, IN PARTICULAR IN HEAD IMPACT ON WINDSHIELDS, Paper 700426, 1970 International Automobile Safety Conference Compendium, Society of Automotive Engineers, New York, 1970, pp. 1084, 1112.




Calculations

| Increment number | Time increment (sec) | Midpoint G value | $G^{2.5}$ | Incremental SI (Time x $G^{2.5}$) |
|---|---|---|---|---|
| 1 | 0.001 | 7 | 130 | 0.13 |
| 2 | 0.001 | 18 | 1,400 | 1.40 |
| 3 | 0.001 | 23 | 2,500 | 2.50 |
| 4 | 0.002 | 27 | 3,800 | 7.60 |
| 5 | 0.001 | 33 | 6,300 | 6.30 |
| 6 | 0.001 | 40 | 10,000 | 10.00 |
| 7 | 0.001 | 38 | 8,800 | 8.80 |
| 8 | 0.001 | 47 | 15,000 | 15.00 |
| 9 | 0.001 | 75 | 48,000 | 48.00 |
| 10 | 0.001 | 80 | 57,000 | 57.00 |
| 11 | 0.001 | 73 | 46,000 | 46.00 |
| 12 | 0.001 | 56 | 23,000 | 23.00 |
| 13 | 0.001 | 43 | 12,000 | 12.00 |
| 14 | 0.002 | 37 | 8,300 | 16.60 |
| 15 | 0.002 | 33 | 6,200 | 12.40 |
| 16 | 0.001 | 27 | 3,800 | 3.80 |
| 17 | 0.001 | 24 | 2,800 | 2.80 |
| 18 | 0.007 | 20 | 1,800 | 12.60 |
| 19 | 0.001 | 17 | 1,200 | 1.20 |
| 20 | 0.002 | 10 | 330 | 0.66 |
| | | | Severity Index | 287.79 |

Figure 16. Sample calculation of a Severity Index. (From Reference 26)

model, based on the Wayne State Tolerance Curve, is used to determine the tolerable amplitude of brain motion. The response of the simple, damped, spring-mass system shown in Figure 17 is given by

$$\ddot{x} + 2D\omega\dot{x} + \omega^2 x = -b(t) \qquad (6)$$

where $x$ = relative displacement of mass

$\dot{x}$ = relative velocity of mass

$\ddot{x}$ = relative acceleration of mass

$b$ = acceleration of system at point N (driving acceleration)

$D$ = damping coefficient

$\omega$ = angular frequency.

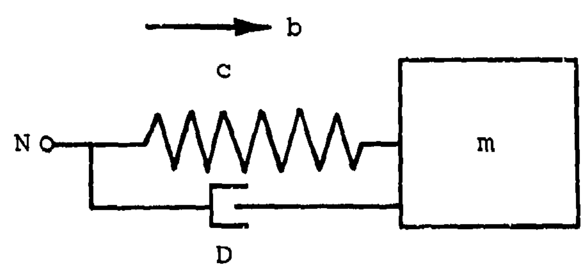



Figure 17. Damped, spring-mass system used in computing J-tolerance. (From Reference 27)

The angular frequency $\omega$ is calculated from the Wayne State Tolerance curve:

$$\omega = \sqrt{k/m} = 635 \text{ rad/sec}$$

where $m$ = mass

$k$ = spring constant

The right side of the differential equation, $-b(t)$, corresponds to its interference function, that is to say, to the excitation caused by the head impact; the solution of the equation is the amplitude

$$x_{max} = f\,[b(t)]$$

Based on the tolerable amplitude $x_{tolr}$, the tolerance value J is given by

$$J = \frac{x_{max}}{x_{tolr}} \qquad (7)$$

where J = 1 is the tolerable limit.

### 4.4.3 Effective Displacement Index

The Effective Displacement Index (EDI), reported by Brinn and Staffeld, is derived from a mathematical spring-mass model based on the work described in Section 4.4.2 (Reference 28). The peak deflection of the model, in inches, is taken as the index of damage. Using a natural frequency of 77 Hz and a damping value of 70.7 percent, a tolerable EDI of 0.15 in. was obtained by fitting the Wayne State Tolerance Curve.

It should be noted here that, as in the case of the Gadd Severity Index, a tolerable EDI is based on anterior-posterior impact only, due to the unavailability of data for other directions.

### 4.4.4 Strain Energy Considerations

Melvin and Evans considered the basic types of skull fracture and investigated the effects of impactor size and shape, skull geometry, and soft tissue (Reference 29).

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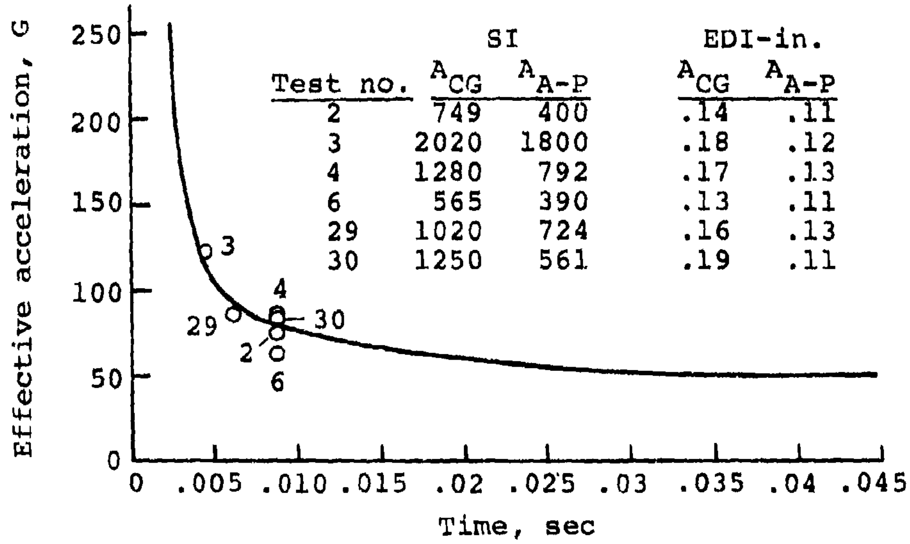
28. Brinn, J., and Staffeld, S. E., THE EFFECTIVE DISPLACEMENT INDEX - AN ANALYSIS TECHNIQUE FOR CRASH IMPACTS OF ANTHROPOMETRIC DUMMIES, Proceedings, Fifteenth Stapp Car Conference, Society of Automotive Engineers, New York, 1971, pp. 817-824.

29. Melvin, J. W., and Evans, F. G., A STRAIN ENERGY APPROACH TO THE MECHANICS OF SKULL FRACTURE, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 666-685.

### 4.4.5 Comparison of Head Injury Predictors

Hodgson and Thomas investigated skull fractures in 40 cadavers that were dropped with their heads striking rigid, flat, hemispherical, and cylindrical surfaces on the front, side, and rear (Reference 30). The Severity Index and the Effective Displacement Index were compared at fracture level for all frontal impacts, and their average values at fracture were found to agree closely with the critical values predicted by the authors of the methods. Results for the frontal flat-plate impacts are shown in Figure 18, where $A_{CG}$ refers to calculation of the indices from resultant acceleration at the head center of gravity. $A_{A-P}$ refers to calculation from anterior-posterior acceleration measured at the point on the skull most distant from the impact site, the condition used in the derivation of the Wayne State Tolerance Curve.




| | SI | | EDI-in. | |
|---|---|---|---|---|
| Test no. | $A_{CG}$ | $A_{A-P}$ | $A_{CG}$ | $A_{A-P}$ |
| 2 | 749 | 400 | .14 | .11 |
| 3 | 2020 | 1800 | .18 | .12 |
| 4 | 1280 | 792 | .17 | .13 |
| 6 | 565 | 390 | .13 | .11 |
| 29 | 1020 | 724 | .16 | .13 |
| 30 | 1250 | 561 | .19 | .11 |

Figure 18. Comparison of SI, EDI, and kinematics of six frontal impacts producing linear fracture. (From Reference 30)

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30. Hodgson, V. R., and Thomas, L. M., COMPARISON OF HEAD ACCELERATION INJURY INDICES IN CADAVER SKULL FRACTURE, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 190-206.

For the drop height range that produced linear fracture in the cadaver's skull due to frontal impact against a rigid, flat plate, the SI for the Alderson 50th-percentile dummy head acceleration response was significantly higher than for the cadaver. The higher acceleration measured for the dummy head would, of course, be expected in such an impact because of the greater rigidity of its metal skull. The EDI, on the other hand, was essentially the same for both cadaver and dummy.

Fan analyzed predictions of the Gadd Severity Index and the Vienna Institute of Technology Brain Model (J-tolerance) and concluded that improvements in both techniques could be made (Reference 31). He concluded that the SI puts too much weight on the acceleration but ignores the time factors of an impulse. Fan's revised approach to SI calculation involves a successive approximation method where variable weighting factors are applied to both acceleration and time. A revised brain model also is presented, based on the Vienna model, with additional information included on dynamic properties of the human skull-brain system. The revised brain model utilized an equivalent viscous damping of 40 percent of critical damping and, on comparison with the Wayne State Tolerance Curve, yielded tolerable values of brain deformation, velocity, and angular velocity of 1.25 in., 135.5 in./sec, and 175 rad/sec, respectively. The maximum deviation from the Wayne State curve was reported to be within 5 percent.

## 4.5 NECK IMPACT TOLERANCE

Tolerance of the human neck to rotation, as experienced in whiplash, and to localized impact loading has been investigated. Mertz and Patrick determined the moment about the occipital condyles (considered to be the center for rotation of the head with respect to the neck) at the threshold of pain for volunteer subjects. On the basis of their investigations, tolerable levels for neck flexion (forward rotation) and neck extension (backward rotation) of a 50th-percentile adult male are proposed in Reference 32.

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31. Fan, W. R. S., INTERNAL HEAD INJURY ASSESSMENT, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 645-665.

32. Mertz, H. J., and Patrick, L. M., STRENGTH AND RESPONSE OF THE HUMAN NECK, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 207-255.

Gadd, Culver, and Nahum, using unembalmed, elderly cadavers, investigated the relationship between rotation and resisting moment in hyperextension and lateral flexion (Reference 33).

## 4.6 CHEST IMPACT TOLERANCE

An extensive research program on impact response of the human thorax has been reported by Kroell, Schneider, and Nahum (Reference 34). A 6-in.-diameter rigid impactor of varying mass and moving at a range of speeds was used to strike unembalmed, seated cadavers. Deflection and force were measured as functions of time. Figure 19 (from Reference 34) shows the Abbreviated Injury Scale (AIS) (Reference 35) plotted against normalized chest deflection (deflection divided by chest anterior-posterior diameter). The least-squares fit shown has an associated correlation coefficient of 0.772. However, as seen in Figure 19, the restrained back data appears to follow a different trend. Scaling of the relationships between force and penetration has been reported in Reference 36.

The Effective Displacement Index (EDI), which was discussed in Section 4.4.3 in reference to head injury, also has been applied to chest injury by Brinn and Staffeld (Reference 28). In agreement with the discussion in the preceding paragraphs, they point out that there is evidence that rib cage deflection may be the governing criterion for chest injury and that the measurement of this deflection might be the basis for a chest

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33. Gadd, C. W., Culver, C. C., and Nahum, A. M., A STUDY OF RESPONSES AND TOLERANCES OF THE NECK, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 256-268.

34. Kroell, C. K., Schneider, D. C., and Nahum, A. M., IMPACT TOLERANCE AND RESPONSE OF THE HUMAN THORAX II, Proceedings, Eighteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1974, pp. 383-457.

35. THE ABBREVIATED INJURY SCALE (AIS), Joint Committee of the American Medical Association, American Association for Automotive Medicine, and the Society of Automotive Engineers; American Association for Automotive Medicine, Morton Grove, Illinois, 1976 Revision.

36. Neathery, R. F., ANALYSIS OF CHEST IMPACT RESPONSE DATA AND SCALED PERFORMANCE RECOMMENDATIONS, Proceedings, Eighteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1974, pp. 459-493.

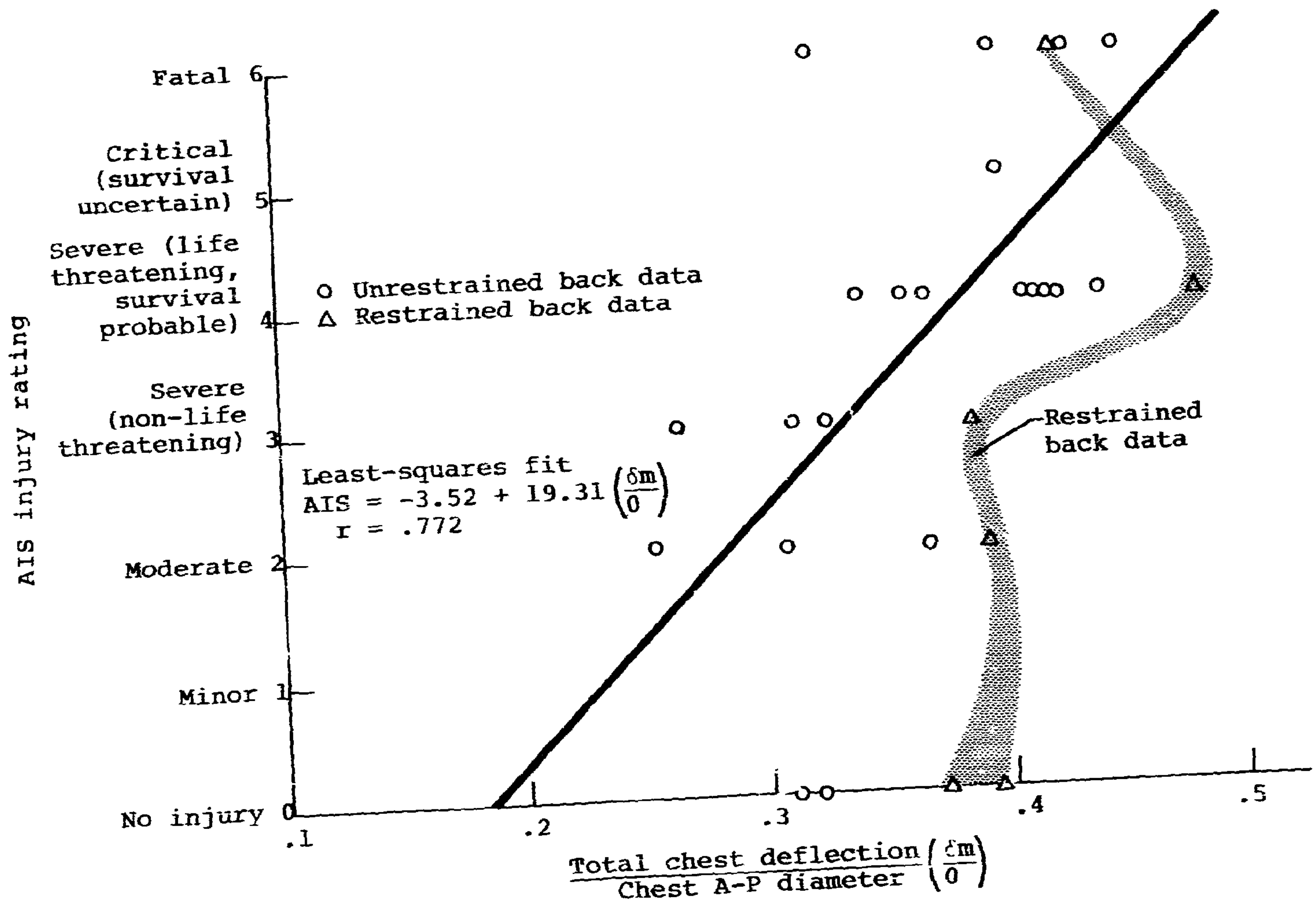


Figure 19. AIS injury rating versus normalized chest deflection. (From Reference 34)

survivability index. However, the acute distress experienced by human volunteers subjected to whole-body deceleration has been considered a justification for an independent hazard index based on acceleration. Using a natural frequency of 15 Hz and a damping value of 25 percent, a maximum EDI of 2.2 in. was obtained for voluntary human exposure. Based on experience with energy-absorbing steering columns, an EDI of 2.8 in. was suggested as a test limit for current-design anthropomorphic dummies.

On the basis of 16 dives performed by an instrumented professional high diver onto a mattress, combined with the results of earlier studies, a long-duration acceleration tolerance level of 60 G with a pulse duration of 100 msec has been recommended for the thorax in the anterior-posterior direction (Reference 37).

## 4.7 ABDOMINAL IMPACT TOLERANCE

Relative to other body regions, little information is available concerning abdominal tolerance to blunt impact trauma. Snyder has reported that one of the major reasons is a "marked disagreement between medical investigators both as to the frequency with which various abdominal organs are involved, as well as the significance of the trauma" (Reference 6). Although a number of experimental animals, mainly hogs, have been utilized in abdominal impact studies, the large number of variables precludes either generalization or, certainly, extrapolation of the results to humans. The large number of organs, their complexity, and their lack of symmetry make the exact location and direction of the impact critical. Also, there are additional considerations, such as whether or not the bladder is full, or whether the stomach is full or empty. Summaries of research in abdominal impact may be found in References 38 and 39.

37. Mertz, H. J., and Gadd, C. W., THORACIC TOLERANCE TO WHOLE-BODY DECELERATION, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 135-157.

38. Frey, C. G., INJURIES TO THE THORAX AND ABDOMEN, Paper 700195, Human Anatomy, Impact Injuries, and Human Tolerances, Society of Automotive Engineers, New York, 1970, pp. 69-76.

39. Gogler, E., et al., BIOMECHANICAL EXPERIMENTS WITH ANIMALS ON ABDOMINAL TOLERANCE LEVELS, Proceedings, Twenty-First Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1977, pp. 713-751.

## 4.8 SPINAL INJURY TOLERANCE

Damage to the vertebral column, particularly the upper lumbar and lower thoracic regions, occurs frequently in $+G_z$ impact, where the force is directed parallel to the spine. A $+G_z$ component can be expected in the use of aircraft ejection seats and in aircraft accidents, particularly those involving helicopters, where a vertical component of impact force is usually present. A recent summary of research on $+G_z$ impact exposure limits is contained in Reference 40.

Various mathematical models have been developed for prediction of spinal response to $+G_z$ loading. An obvious injury mechanism is the inertial loading sustained by the vertebrae, resulting in compression fractures. Therefore, the earliest models have been one-dimensional spring-mass systems that assume all the load to be borne by the vertebral body. One such model that has been used extensively in ejection seat evaluation is discussed in Section 4.8.1. However, this simplified approach cannot predict all types of spinal injury and cannot assess the significance of spinal curvature. More comprehensive approaches have included the flexural beam model of Soechting (Reference 41) and the discrete parameter model of Orne and Liu (Reference 42) that accounts for the effects of eccentric loading as well as spinal curvature. Sections 4.8.2 and 4.8.3 describe two more recently developed models that appear promising for use in assessment of spinal injury potential in aircraft crashes.

### 4.8.1 Dynamic Response Index

The human response to short-duration accelerations applied in the upward vertical direction parallel to the spine ($+G_z$) has been modeled by a single lumped-mass, damped-spring system as

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40. Von Gierke, H. E., and Brinkley, J. W., IMPACT ACCELERATIONS, In Foundation of Space Biology and Medicine, Volume II, Book One, Scientific and Technical Information Office, National Aeronautics and Space Administration, Washington, D. C., 1975, pp. 214-246.

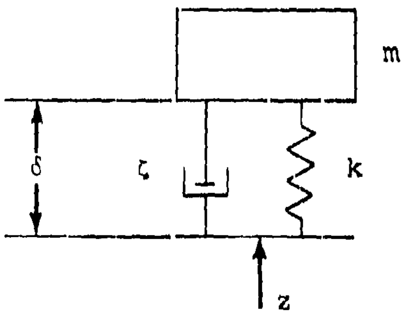
41. Soechting, J. F., RESPONSE OF THE HUMAN SPINAL COLUMN TO LATERAL DECELERATION, Journal of Applied Mechanics, Vol. 40, 1973, pp. 643-649.

42. Orne, David, and Liu, Y. K., A MATHEMATICAL MODEL OF SPINAL RESPONSE TO IMPACT, Journal of Biomechanics, Vol. 4, 1971, pp. 49-71.

shown in Figure 20 (Reference 43). In this model, it has been assumed that the total body mass that acts upon the vertebrae to cause deformation can be represented by the single mass. In use, the relationship

$$\frac{d^2\delta}{dt^2} + 2\zeta\omega_n \frac{d\delta}{dt} + \omega_n^2\delta = z \qquad (8)$$

is solved through the use of a computer. The third term, which includes the deformation of the spine, $\delta$, divided by the gravitational acceleration, g, is referred to as the Dynamic Response Index (DRI). The model is used to predict the maximum deformation of the spine and associated force within the vertebral column for various short-duration acceleration inputs. The spring stiffness for the model was determined from tests of human cadaver vertebral segments; damping ratios were determined from measurements of mechanical impedance of human subjects during vibration and impact.




$m$ = mass (lb-sec$^2$/in.)

$\delta$ = deflection (in.)

$\zeta$ = damping ratio

$k$ = stiffness (lb/in.)

$z$ = acceleration input (in./sec$^2$)

$$*\text{DRI} = \frac{\omega_n^2 \, \delta_{max}}{g}$$

$\omega_n$ = natural frequency of the analog = $\sqrt{k/m}$ (rad/sec)

$g$ = 386 in./sec$^2$

*Dynamic Response Index

Figure 20. Spinal-injury model. (From Reference 43)

43. Stech, E. L., and Payne, P. R., DYNAMIC MODELS OF THE HUMAN BODY, Frost Engineering Development Corp.; AMRL Technical Report 66-157, Aerospace Medical Research Lab, Wright-Patterson Air Force Base, Ohio, November 1969, AD 701383.

An analytical effort was conducted to determine the degree of correlation between the spinal injury (DRI) model and injuries experienced in operational aircraft ejection seats (Reference 44). Figure 21 shows the relationship between operational acceleration environments and actual spinal injury rates. The response of the model is expressed in DRI values. It can be seen that the injury probability does vary with the DRI but that the cadaver data show a higher probability of injury than do the operational data. It would be expected that the intact, living vertebral column imbedded in the torso would be stronger than cadaver segments; consequently, this result might be predicted.

To establish acceptable ejection seat acceleration environments, the Air Force has adopted a system using a combination of acceleration components and the DRI.

In Specification MIL-S-9479 (Reference 45), the acceleration levels to be imposed on the seat occupant are controlled by acceleration, time, and DRI as shown in the following relationship:

$$\sqrt{\left(\frac{DRI}{DRI_L}\right)^2 + \left(\frac{G_x}{G_{xL}}\right)^2 + \left(\frac{G_y}{G_{yL}}\right)^2} \leq 1.0 \qquad (9)$$

Here $G_x$ and $G_y$ are measured acceleration magnitudes in the x and y directions and $G_{xL}$ and $G_{yL}$ are the limit acceleration parameters as read from acceleration versus time curves included in the specification. DRI is the DRI value computed from Equation (7) for the positive z direction. $DRI_L$ is the limit value of the DRI. The value of $DRI_L$ is 18 unless the resultant acceleration vector is more than 5 conical degrees off the z axis and aft of the plane of the seat back, in which case, the value of $DRI_L$ is 16. The computed value for the left-hand term of Equation (9) may not exceed one.

The DRI is calculated from Equation (8) with model coefficients for the positive spinal case (eyeballs down) defined for the

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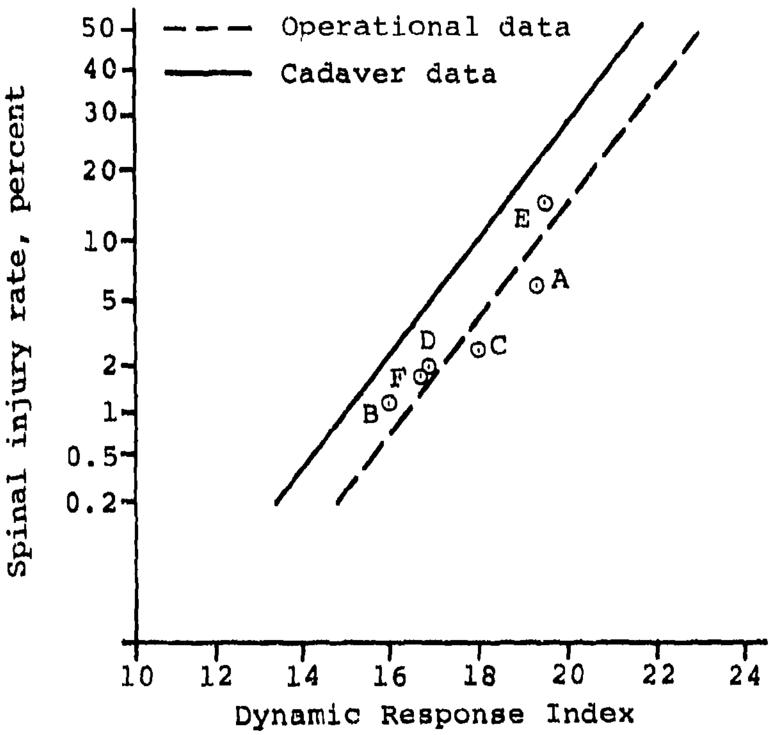
44. Brinkley, J. W., and Shaffer, J. T., DYNAMIC SIMULATION TECHNIQUES FOR THE DESIGN OF ESCAPE SYSTEMS: CURRENT APPLICATIONS AND FUTURE AIR FORCE REQUIREMENTS, Aerospace Medical Research Lab; AMRL Technical Report 71-29-2, Wright-Patterson Air Force Base, Ohio, December 1971, AD 740439.

45. Military Specification, MIL-S-9479, SEAT SYSTEM, UPWARD EJECTION, AIRCRAFT, GENERAL SPECIFICATION FOR, Department of Defense, Washington, D. C., March 1971.




| Aircraft type | Nonfatal ejections |
|---|---|
| A* | 64 |
| B* | 62 |
| C | 65 |
| D* | 89 |
| E | 33 |
| F | 48 |

*Denotes rocket catapult

Figure 21. Probability of spinal injury estimated from laboratory data compared to operational experience. (From Peference 44)

mean age of the Air Force flying population (age 27.9 years). The model coefficients are as follows:

$$\omega_n = 52.9 \text{ rad/sec}$$

$$\zeta = 0.224$$

The DRI has been shown to be effective in predicting spinal injury potential for $+G_z$ acceleration environments in ejection seats. However, it should be remembered that it is a simple model of a complex dynamic system and that the correlations made are for ejection seat acceleration-time pulses that can vary widely from crash pulses. In particular, the rate of onset can be an order of magnitude greater than for ejection seat pulses. Also, the position of the spine at the time of impact can have a significant influence on the susceptibility to vertebral damage. Therefore, a helicopter pilot leaning forward in his seat might be expected to respond differently from an upright, well-restrained ejection seat occupant, and, thus, have lower tolerance to impact.

### 4.8.2 Wayne State University Two-Dimensional Model

A two-dimensional spinal model that considers the details of load transmission among individual vertebrae has been developed by King and Prasad (Reference 46). The model considers the natural spinal curvatures and the effects of flexion and eccentric inertial loading on the spine. Head and neck motions are simulated, and their effects on the forces and moments in the thoracic and lumbar spine can be studied for off-axis impacts in the midsagittal plane. The input acceleration pulse can be an arbitrary function of time. The restraint and support systems have been included to properly simulate a seated vehicle occupant. The experimental data for validation of the model were obtained from cadaveric runs with the spine in the erect and hyperextended mode so that the model incorporates the ability to simulate both spinal configurations.

The following assumptions were made in the mathematical development:

- The 24 vertebral bodies, the head, and the pelvis are rigid bodies constrained to move in the midsagittal plane.

- Each rigid body has three degrees of freedom in the midsagittal plane--two translational and one rotational.

- The intervertebral discs are massless, and deformation of the spine takes place at the discs.

46. King, A. I., and Prasad, P., AN EXPERIMENTALLY VALIDATED MODEL OF THE SPINE, Journal of Applied Mechanics, Vol. 41, No. 3, September 1974, pp. 546-550.

- The discs are replaced by a system of springs and dampers--one spring and damper for axial forces and another spring and damper arrangement for restoring torques due to relative angular motion between adjacent vertebral bodies.

- The facets and laminae are springs connected to the vertebral body by a massless rigid rod.

- Each rigid body is assumed to carry a portion of the torso weight that is eccentric with respect to the center line of the spine.

- The rigid bodies are arranged to simulate the spinal curvatures as closely as possible.

Equations of motion are derived for each vertebra, resulting in a set of 78 second-order differential equations that are solved numerically on a digital computer.

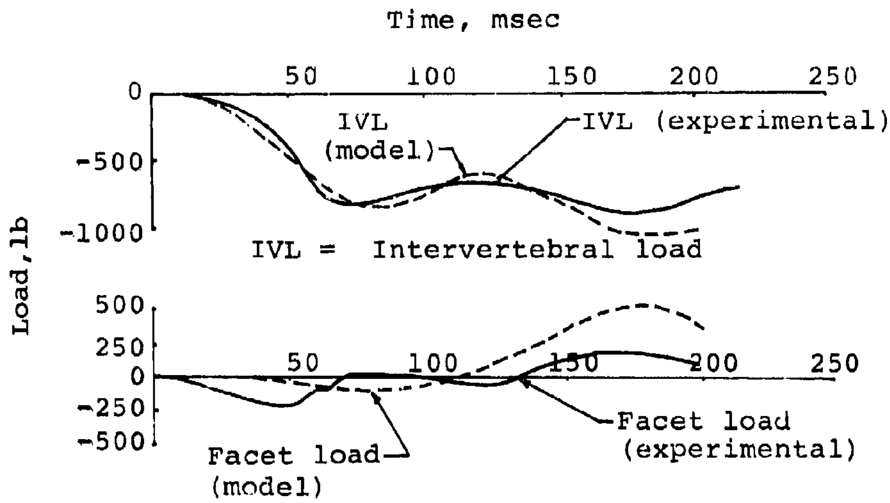
Experiments involving the use of human cadavers were carried out for model validation. An acceleration input was applied at the pelvis while the top of the head was allowed to remain stress free. The parameter used for validation was the force between two adjacent vertebrae, and an intervertebral load cell was developed to provide the magnitude and line of action of the force. Comparisons of model predictions and experimental data are shown in Figure 22 for two 10 G runs with the spine in different positions, where the loads were measured between the second and third lumbar vertebrae. Shown are both the loads between vertebral bodies (IVL) and those in the facets, which limit relative rotation of the vertebrae. The significance of the initial curvature of the spine is evidenced by the difference in response between the erect and hyperextended (backward rotation of the torso) modes.

This model appears to be potentially useful in spinal injury prediction, provided that dynamic fracture loads for vertebrae are known, as discussed in Reference 47.

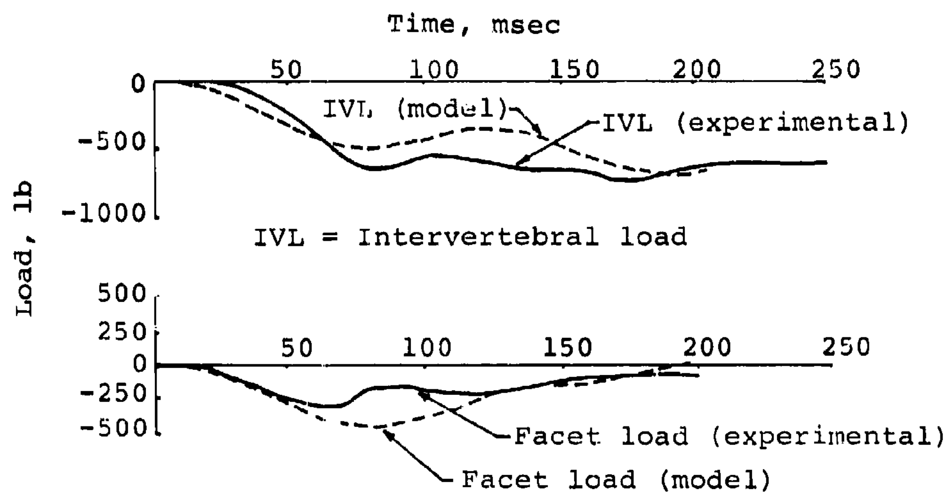
### 4.8.3 Air Force Head-Spine Model

Under the sponsorship of the U. S. Air Force Aeromedical Research Laboratory, a three-dimensional, discrete model of the human spine, torso, and head was developed for the purpose of

47. Hakim, N. S., and King, A. I., STATIC AND DYNAMIC ARTICULAR FACET LOADS, Proceedings, Twentieth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1976, pp. 607-639.




(a) Erect




(b) Hyperextended

Figure 22. Comparison of model output and experimental data for 10 G runs with the spine in the (a) erect and (b) hyperextended modes. (From Reference 46)

evaluating mechanical response in pilot ejection. It was developed in sufficient generality to be applicable to other body response problems, such as occupant response in aircraft crash and head-spinal system response to arbitrary loads. There are no restrictions on the distribution of direction of applied loads, therefore, a wide variety of situations can be treated. The model has been described in Reference 48.

The anatomy is modeled by a collection of rigid bodies that represent skeletal segments, such as the vertebrae, pelvis, head, and ribs, interconnected by deformable elements that represent ligaments, cartilaginous joints, viscera, and connective tissues. Techniques for representing other aspects of the ejection environment, such as harnesses and the seat geometry, are included also. The model is valid for large displacements of the spine and treats material nonlinearities. The elements of the model are illustrated in Figure 23.

The basic model is modular in format so that various components can be omitted or replaced by simplified representations. Thus, while the complete model is rather complex and involves substantial computational effort, various simplified models, which are quite effective in duplicating the response of the complete model within a range of conditions, are available. Three methods of solution are available for the analysis: direct integration in time either by an explicit, central difference method or by an implicit, trapezoidal method, and a frequency analysis method.

A variety of conditions have been simulated, including different rates of onset, ejection at angles, effects of lumbar curvature, and eccentric head loadings. It has been shown that large initial curvatures and perfectly vertical acceleration loadings, which cause large bending moments, result in substantial flexural response of the spine. It has been further shown that the combination of the spine's low flexural stiffness, initial curvature, and mass eccentricity are such that stability cannot be maintained in a 10 G ejection without restraints or spine-torso-musculature interaction.

The complete models were used mainly to study the effects of the rib cage and viscera on spinal response. The flexural stiffness of the torso is increased substantially by a visceral model, even though it has no inherent flexural stiffness. In addition, the viscera provide significant reductions in the axial loads.

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48. Belytschko, T., Schwer, L., and Schultz, A., A MODEL FOR ANALYTIC INVESTIGATION OF THREE-DIMENSIONAL HEAD-SPINE DYNAMICS - FINAL REPORT, University of Illinois at Chicago Circle; AMRL Technical Report 76-10, Aerospace Medical Research Lab, Wright-Patterson Air Force Base, Ohio, April 1976, AD A025911.

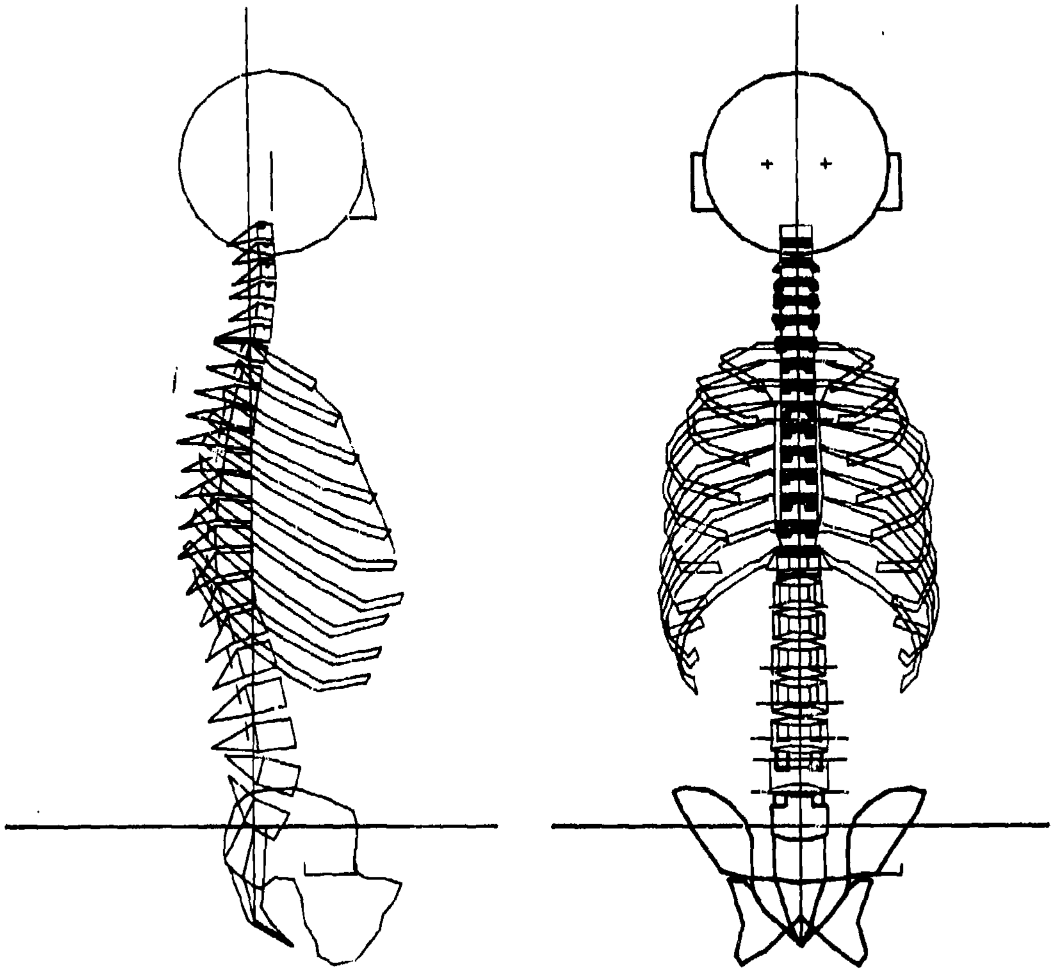

Figure 23. Three-dimensional head-spine model. (From Reference 48)

#### 4.8.4 Vertebral Properties

Accurate strength and deflection properties of the seated human torso when exposed to inertial loads are urgently needed. Reference 49 includes a consolidation of the available

49. MODELS AND ANALOGUES FOR THE EVALUATION OF HUMAN BIODYNAMIC RESPONSE, PERFORMANCE, AND PROTECTION, AGARD Conference Proceedings No. 253, NATO Advisory Group for Aerospace Research and Development, Neuilly sur Seine, France, 1968.

data from King, Kazarian, and Hodgson (References 46, 50, and 51) for the head and spinal column exposed to $+G_z$ loading. They present a set of recommended properties that may find use in a more sophisticated model for spinal injury than the DRI.

## 4.9 LEG INJURY TOLERANCE

Femoral fracture due to longitudinal impact on the knee has been studied extensively, probably because of the frequency of this type of injury in automobile accidents.

Based on cadaver data obtained by Patrick, et al. (Reference 52), King, et al., recommended a peak fracture load of 1700 lb as a realistic criterion (Reference 53). This value has since been adopted for use in Federal Motor Vehicle Safety Standard 208. Recent experiments reported by Powell, et al., point to this value as being conservative for impacts of less than 20 msec (Reference 54).

Viano has presented a criterion that assesses the dependence of the permissible human knee load on the duration of the primary

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50. Kazarian, L., and Graves, G. A., COMPRESSIVE STRENGTH CHARACTERISTICS OF THE HUMAN VERTEBRAL CENTRUM, AMRL Technical Report 77-14, Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio, 1977.

51. Hodgson, V. R., and Thomas, L. M., HEAD IMPACT RESPONSE - PROPOSAL NUMBER VRI 7.2, Society of Automotive Engineers, Inc., New York, 1975.

52. Patrick, L. M., Kroell, C. K., and Mertz, H. G., FORCES ON THE HUMAN BODY IN SIMULATED CRASHES, Proceedings, Ninth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1965, pp. 237-259.

53. King, J. J., Fan, W. R. S., and Vargovick, R. J., FEMUR LOAD INJURY CRITERIA - A REALISTIC APPROACH, Proceedings, Seventeenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1973, pp. 509-524.

54. Powell, W. R., et al., INVESTIGATION OF FEMUR RESPONSE TO LONGITUDINAL IMPACT, Proceedings, Eighteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1974, pp. 539-556.

force exposure (Reference 55). Based on the knee impact data from previous experiments with both fresh and embalmed cadavers, Viano suggests the following femur injury criterion (FIC) to define a permissible peak knee load:

$$F(kN) = 23.14 - 0.71\ T(msec),\ T < 20\ msec$$

$$F(kN) = 8.90,\ T \geq 20\ msec \qquad (10)$$

or, in English units,

$$F(lb) = 5200 - 160\ T(msec),\ T < 20\ msec$$

$$F(lb) = 2000,\ T \geq 20\ msec \qquad (10a)$$

The relationship of Equation (10) is illustrated in Figure 24.

## 4.10 ABBREVIATED INJURY SCALE

The Abbreviated Injury Scale (AIS), first reported in 1971, was developed as a comprehensive system for rating tissue damage that would be acceptable to physicians, engineers, and researchers working in automotive crash investigation. An engineer concerned with interpretation or use of crash injury data can find information on the AIS and its use in References 35 and 56.

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55. Viano, C. C., CONSIDERATIONS FOR A FEMUR INJURY CRITERION, PROCEEDINGS, Twenty-First Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1977, pp. 445-473.

56. States, M. M., et al., FIELD APPLICATION AND DEVELOPMENT OF THE ABBREVIATED INJURY SCALE, Proceedings, Fifteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1971, pp. 710-738.

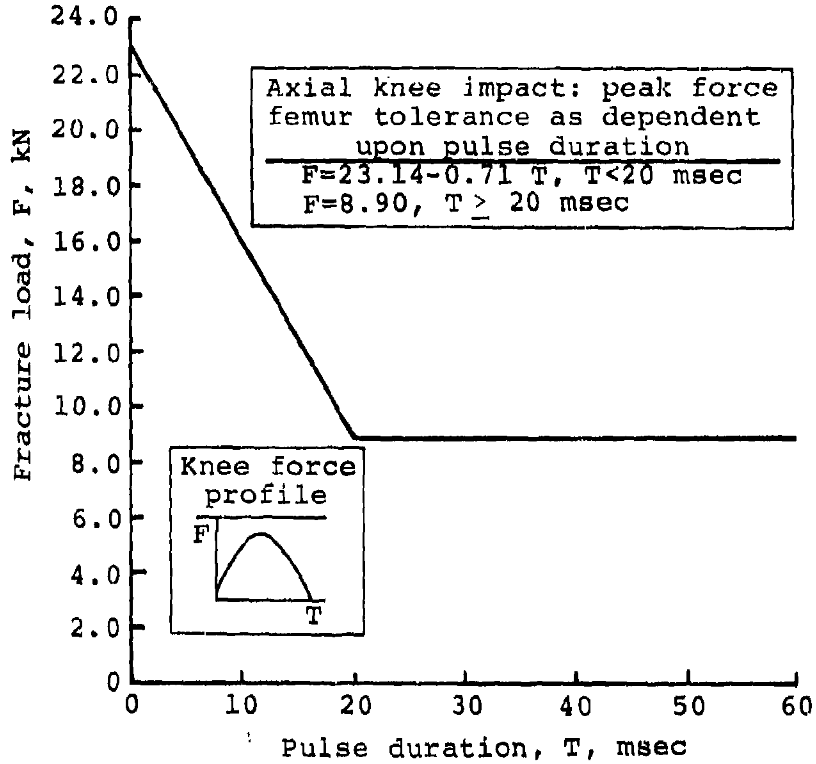


Figure 24. Femur injury criterion. (From Reference 55)

# 5. OCCUPANT MOTION ENVELOPES

## 5.1 INTRODUCTION

The purpose of this chapter is to acquaint the reader with the extent of an aircraft occupant's motion in a crash environment. This knowledge is vital in designing for occupant protection from injury due to impact with the aircraft interior, a topic discussed in Volume IV.

The kinematics of body action associated with aircraft crash impacts are quite violent, even in accidents of moderate severity. The flailing of body parts is much more pronounced when the aircraft occupant is restrained in a seat with a lap belt only. However, even with a lap belt and a shoulder harness that are drawn up tightly, multidirectional flailing of the head, arms, and legs, and to a lesser extent, the lateral displacement of the upper torso within its restraint harnessing, is extensive. If it were possible to provide adequate space within the occupant's immediate environment, this flailing action of a fully restrained occupant would not be a particular problem. Since space for occupants is usually at a premium in aircraft, especially in cockpit areas, it is not feasible to remove structural parts of the aircraft sufficiently to keep the occupant from striking them. The only alternative is to design the occupant's immediate environment so that, when the body parts do flail and contact rigid and semirigid structures, injury potential is minimized.

An occupant who is even momentarily debilitated by having his head strike a sharp, unyielding structural object or by a leg injury can easily be prevented from rapidly evacuating the aircraft and may not survive a postcrash fire or a water landing. The importance of occupant environment designed for injury prevention, therefore, should be emphasized if crash protection is to be ensured.

Several approaches are available to alleviate potential secondary impact problems. The most direct approach, which should be taken if practical, is to relocate the hazardous structure or object out of the occupant's reach. Such action is normally subject to tradeoffs between safety and operational or human engineering considerations. If relocation is not a viable alternative, the hazard might be reduced by mounting the offending structure on frangible or energy-absorbing supports and applying a padding material to distribute the contact force over a larger area. Application of protective padding for both energy absorption and load distribution is discussed in Volume IV.

## 5.2 FULL RESTRAINT

Body extremity strike envelopes are presented in Figures 25 through 27 for a 95th-percentile Army aviator wearing a restraint system that meets the requirements of MIL-S-58095(AV) (Reference 8). The restraint system consists of a lap belt, lap belt tiedown strap, and two shoulder straps. The forward motion shown in Figures 25 and 26 was obtained from a test utilizing a 95th-percentile anthropomorphic dummy subjected to a spineward ($-G_x$) acceleration of 30 G. The lateral motion is based on expected restraint system deflections in a 30 G lateral environment.

## 5.3 LAP-BELT-ONLY RESTRAINT

Although upper torso restraint is required in new Army aircraft, strike envelopes for a 95th-percentile aviator wearing lap-belt-only restraint are presented for possible use in Figures 28 through 30. They are based on 4 G accelerations and 4 in. of torso movement away from the seat laterally and forward.

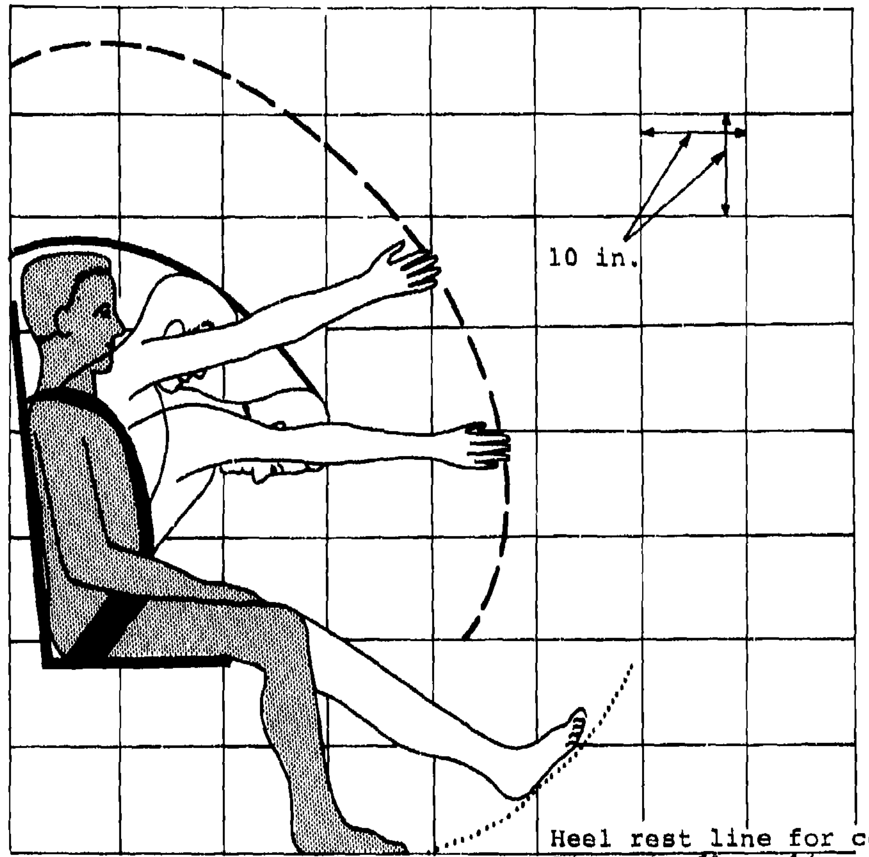


Figure 25. Full-restraint extremity strike envelope - side view.

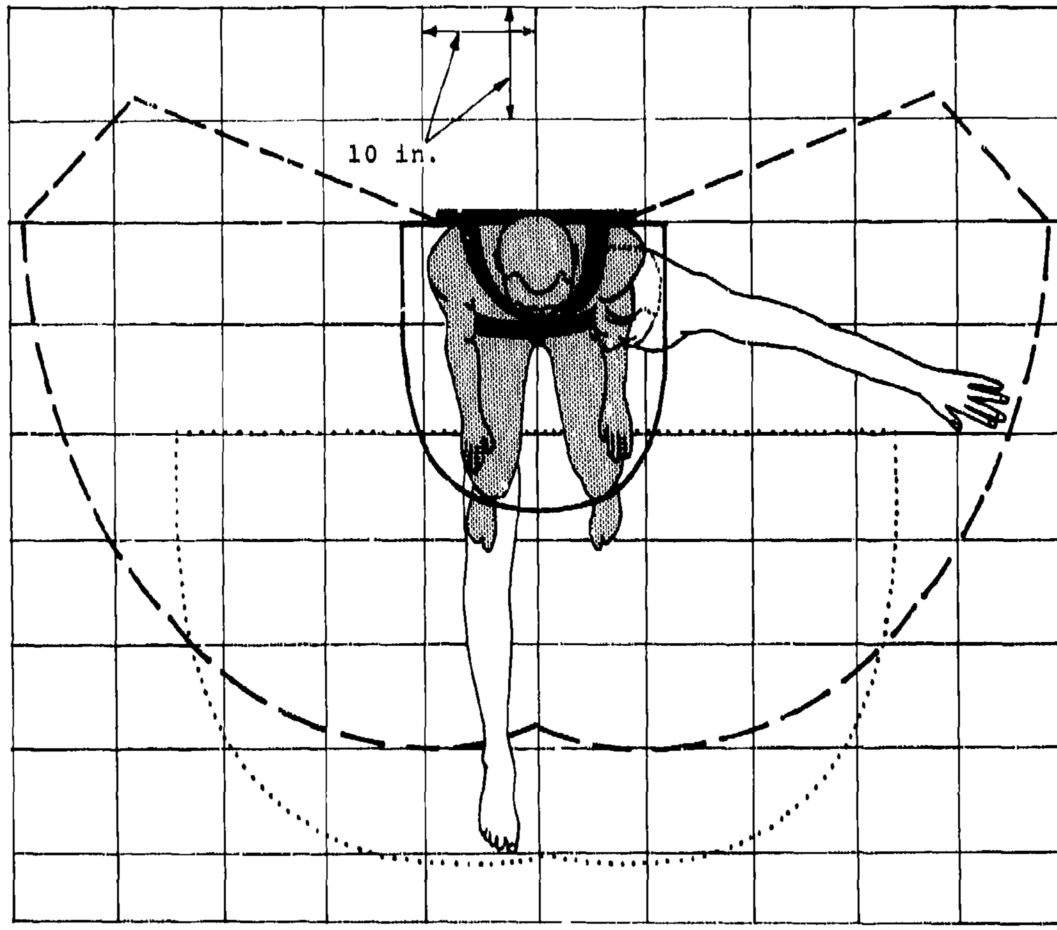


Figure 26. Full-restraint extremity strike envelope - top view.

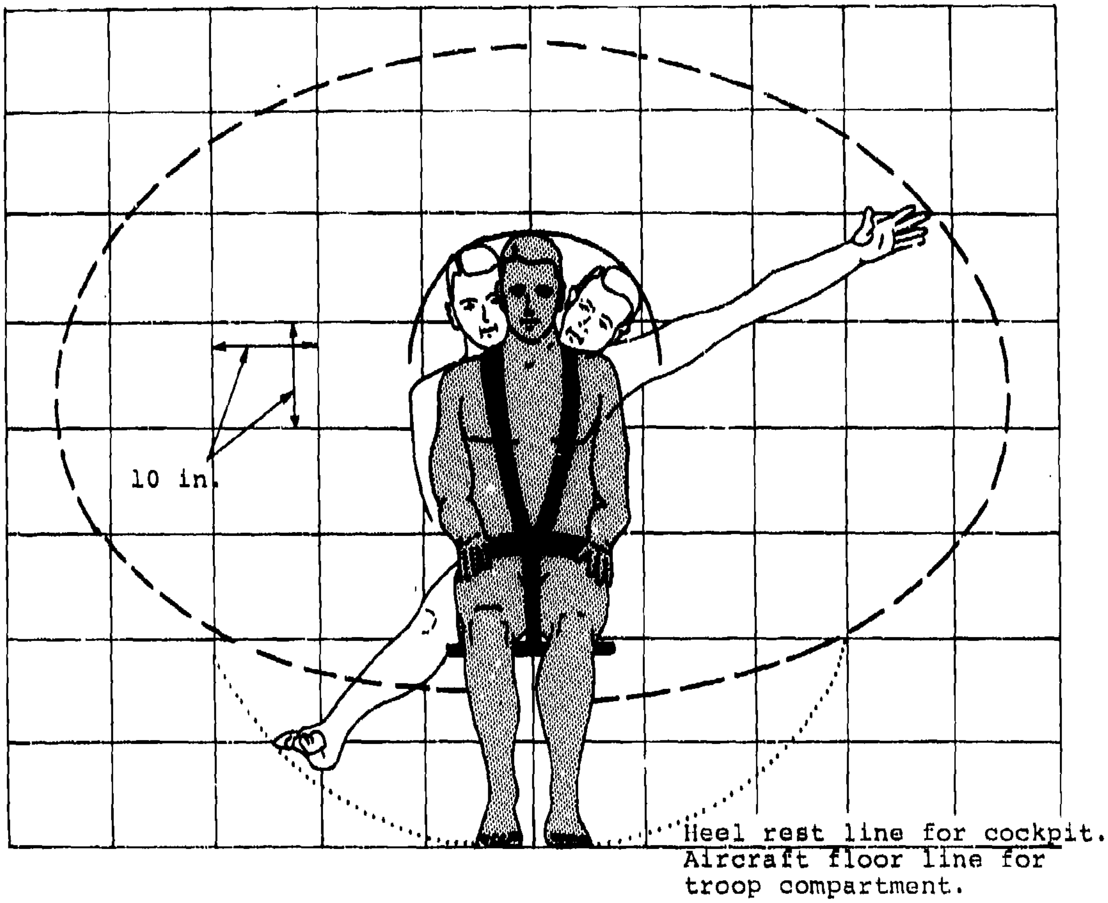


Figure 27. Full-restraint extremity strike envelope - front view.

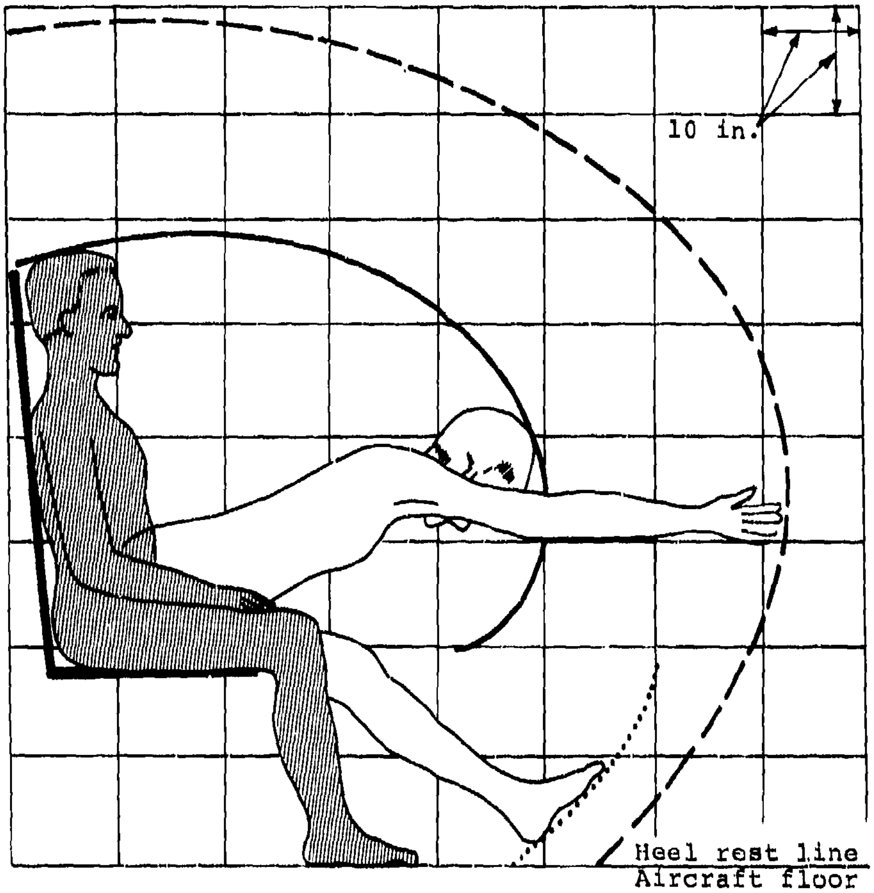


Figure 28. Lap-belt-only extremity strike envelope - side view.

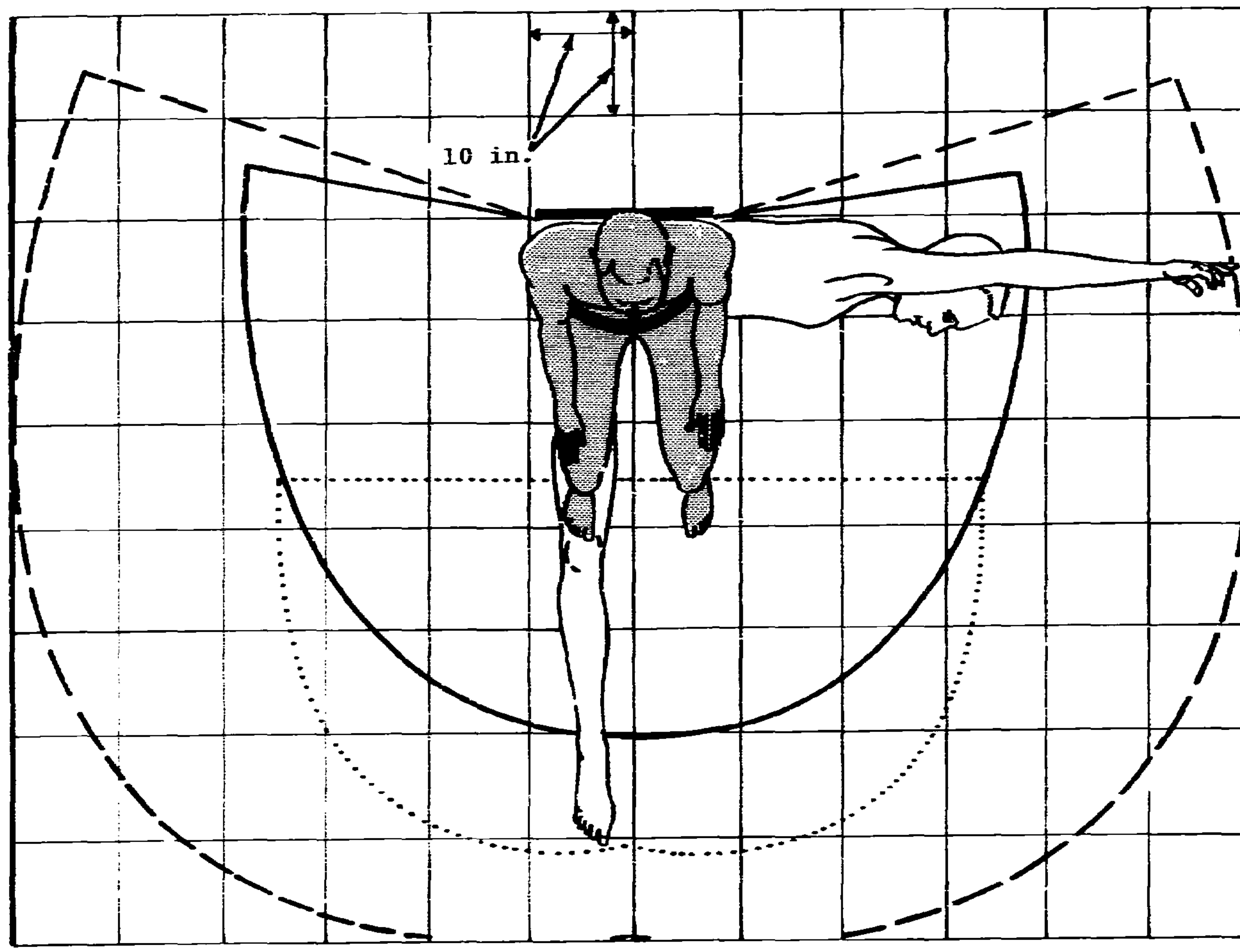

Figure 29. Lap-belt-only extremity strike envelope - top view.

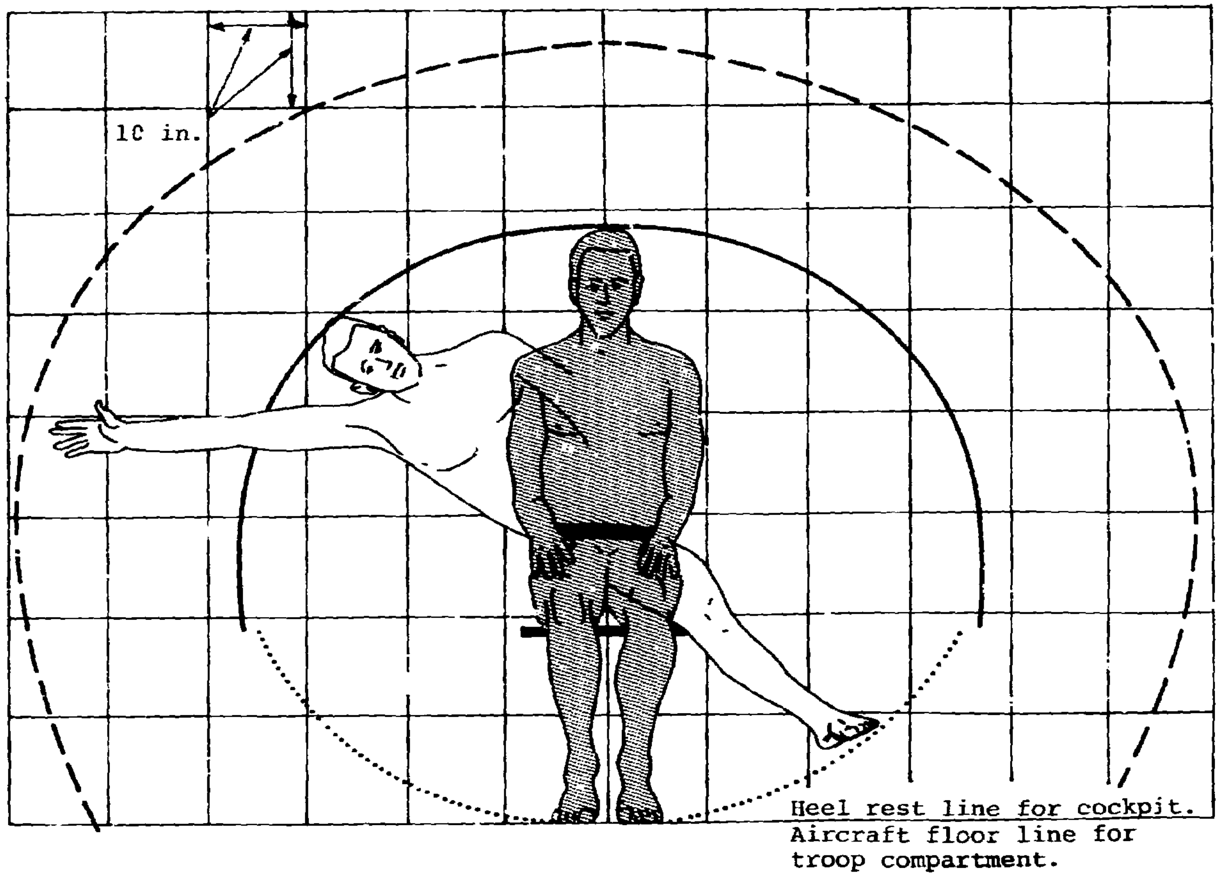


Figure 30. Lap-belt-only extremity strike envelope - front view.

# 6. HUMAN BODY DIMENSIONS AND MASS DISTRIBUTION

## 6.1 INTRODUCTION

This chapter presents information on the dimensions and properties of the human body. Anthropometric measurements are external dimensions of the human body that can be used to define aircraft requirements such as seat height and width, eye height, or cabin height. A specialized type of anthropometric measurement is the "link length," or distance between joint centers, which can be used in locating control positions and is essential for the design of mathematical or physical simulators of the human body. Finally, the inertial properties of the body and parts of the body also are required in the design of human simulators.

## 6.2 ANTHROPOMETRY

Anthropometry is a specialized area of physical anthropology that is concerned with the measurement of the human body and its parts. Two types of anthropometric measurements have been recorded, and the use of both types in vehicle design has been summarized in Reference 57. Conventional dimensions of the body obtained with subjects in rigid, standardized positions are easily obtained. Extensive collections of such data are used in clothing design and may determine certain vehicle design parameters including seat height and eye height. A second class of data, which may be referred to as workspace dimensions, is more difficult to obtain and can be applied only to the specific workspace studied. However, these workspace dimensions are essential in designing aircraft interiors for maximum occupant protection.

### 6.2.1 Conventional Anthropometric Measurements

Conventional anthropometric measurements of greatest interest in aircraft interior design include those dimensions illustrated in Figure 31, as well as standing height and body weight. The most recent anthropometric survey of U. S. Army aviators is contained in Reference 58, and the dimensions of greatest potential usefulness are presented in Table 4. Corresponding

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57. Roe, R. W., and Kyropoulos, P., THE APPLICATION OF ANTHROPOMETRY TO AUTOMOTIVE DESIGN, Paper 700553, Society of Automotive Engineers, New York, 1970.

58. Churchill, E., et al., ANTHROPOMETRY OF U. S. ARMY AVIATORS - 1970, Anthropology Research Project, Yellow Springs, Ohio; Technical Report 72-52-CE, U. S. Army Natick Laboratories, Natick, Massachusetts, December 1971, AD 743528.

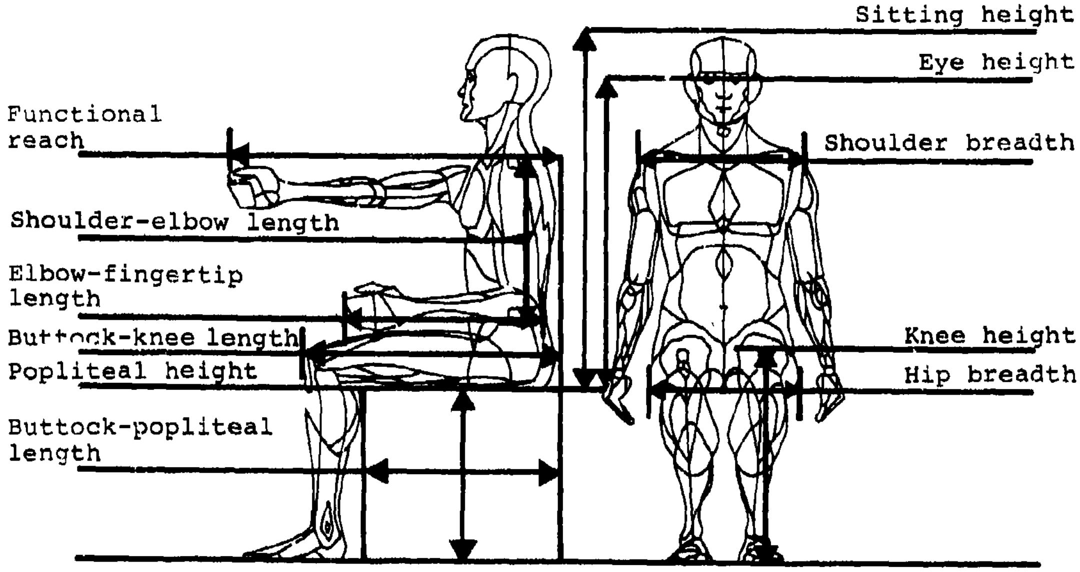


Figure 31. Conventional seated anthropometric dimensions.

dimensions for nonaviators, taken from Reference 59, are listed in Table 5. These dimensions are nude measurements; the dimensions of bulky clothing and helmets must be considered for specific applications.

Because anthropometric surveys involve a large population sample, they generally follow a normal (bell-shaped) distribution. A normal distribution, as shown in Figure 32, can be described in terms of its mean or average value and its dispersion about the mean, often expressed as standard deviation.

The percentile value, which corresponds to a rank order, is a useful statistic for designers. If a group of subjects were ordered from least to greatest for any given measurement, such as standing height, the first percentile would be that part exceeded by 99 percent of the group; the 5th percentile would be that exceeded by 95 percent of the group. The 50th percentile, or median, would be that half of the group exceeded by the other half. For a normally distributed sample, the median value is the same as the mean, or average.

59. THE BODY SIZE OF SOLDIERS - U. S. ARMY ANTHROPOMETRY - 1966, USANL Technical Report 72-51-CE, U. S. Army Natick Laboratories, Natick, Massachusetts, December 1971, AD 743465.

TABLE 4. SUMMARY OF ANTHROPOMETRIC DATA FOR U. S. ARMY AVIATORS (REFERENCE 58)

| Measurement | Percentiles (in.) 5th | 50th | 95th |
|---|---|---|---|
| Weight (lb) | 133.0 | 171.0 | 212.0 |
| Stature | 64.6 | 68.7 | 72.8 |
| Seated height | 33.7 | 35.8 | 37.9 |
| Shoulder breadth | 17.0 | 18.7 | 20.3 |
| Functional reach | 28.8 | 31.1 | 34.2 |
| Hip breadth, sitting | 13.2 | 14.8 | 16.7 |
| Eye height, sitting | 29.0 | 31.0 | 33.1 |
| Knee height, sitting | 19.3 | 20.8 | 22.6 |
| Popliteal height | 15.1 | 16.6 | 18.3 |
| Shoulder-elbow length | 13.3 | 14.4 | 15.6 |
| Elbow-fingertip length | 17.6 | 19.0 | 20.3 |
| Buttock-popliteal length | 17.7 | 19.3 | 21.0 |
| Buttock-knee length | 22.0 | 23.7 | 25.4 |

For a normal distribution, as shown in Figure 32, 68 percent of the sample is included within plus or minus one standard deviation of the mean, and 95 percent within plus or minus two standard deviations.

An example of the use of the statistics of anthropometric data in setting design limits for a vehicle dimension (seat height adjustment range) was presented in Reference 57.

### 6.2.2 Body Joints and Ranges of Motion

Few body joints involve rotation about fixed axes or pivot points. Rather, the instantaneous center of rotation may depend on position, as illustrated in Figure 33 for the shoulder

TABLE 5. SUMMARY OF ANTHROPOMETRIC DATA FOR SOLDIERS (REFERENCE 59)

| Measurement | Percentiles (in.) | | |
|---|---|---|---|
| | 5th | 50th | 95th |
| Weight (lb) | 126.0 | 156.0 | 202.0 |
| Stature | 64.5 | 68.7 | 73.1 |
| Seated height | 33.3 | 35.7 | 38.1 |
| Shoulder breadth | 16.3 | 17.8 | 19.6 |
| Hip breadth, sitting | 11.9 | 13.0 | 14.5 |
| Eye height, sitting | 28.6 | 31.0 | 33.3 |
| Knee height, sitting | 19.6 | 21.3 | 23.1 |
| Popliteal height | 16.0 | 17.5 | 19.2 |
| Shoulder-elbow length | 13.3 | 14.5 | 15.7 |
| Elbow-fingertip length | 17.4 | 18.8 | 20.4 |
| Buttock-popliteal length | 18.0 | 19.6 | 21.3 |
| Buttock-knee length | 21.6 | 23.4 | 25.3 |

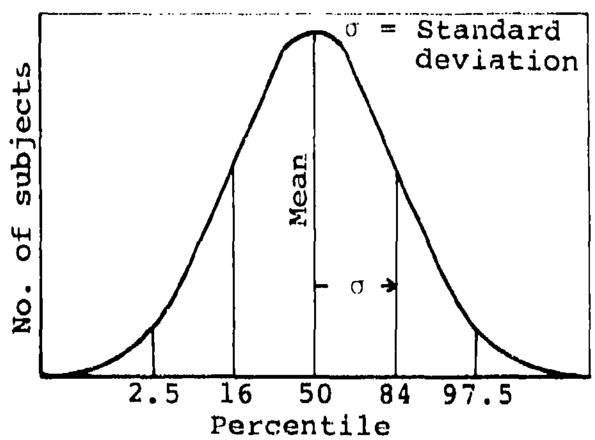


Figure 32. Normal distribution curve.

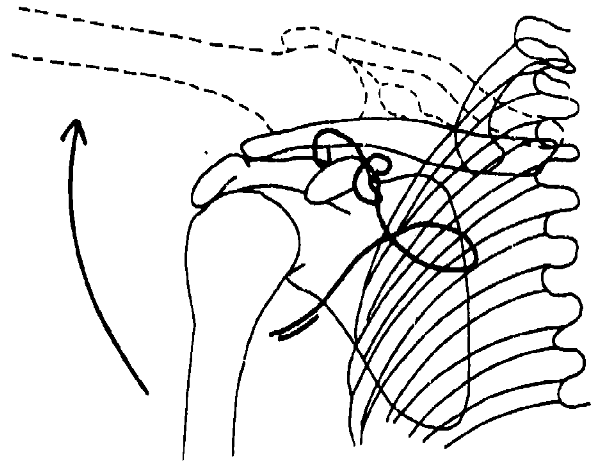

Figure 33. Path of instantaneous center of rotation during shoulder abduction. (From Reference 60)

joint. Dempster reported on an extensive study of workspace requirements for seated operators, in which he determined "link lengths" between effective joint centers for major body parts (References 60 and 61). These link lengths have a number of crashworthiness-related applications: first, in developing or expanding the strike envelopes shown in Chapter 5; second, in designing crash test dummies; and third, in providing numbers for mathematical simulators. From the data of several investigations, the skeletal joint locations for a 50th-percentile Army aviator, illustrated in Figure 34, were developed (Reference 49).

Joint ranges of motion are required in the same areas of application listed above the link lengths. These movements, illustrated in Figure 35, are measured from a standard anatomical position defined as an erect standing posture with the palm surfaces of the hands positioned anteriorly. Various studies have determined the ranges of motion that may be attained voluntarily and under external force; Table 6 lists angles

60. Dempster, W. T., SPACE REQUIREMENTS FOR THE SEATED OPERATOR, Wright Air Development Center; WADC Technical Report 55-159, Wright-Patterson Air Force Base, Ohio, 1955, AO 087892.

61. Dempster, W. T., and Gaughran, G. R. L., PROPERTIES OF BODY SEGMENTS BASED ON SIZE AND WEIGHT, American Journal of Anatomy, Vol. 120, 1967, pp. 33-54.

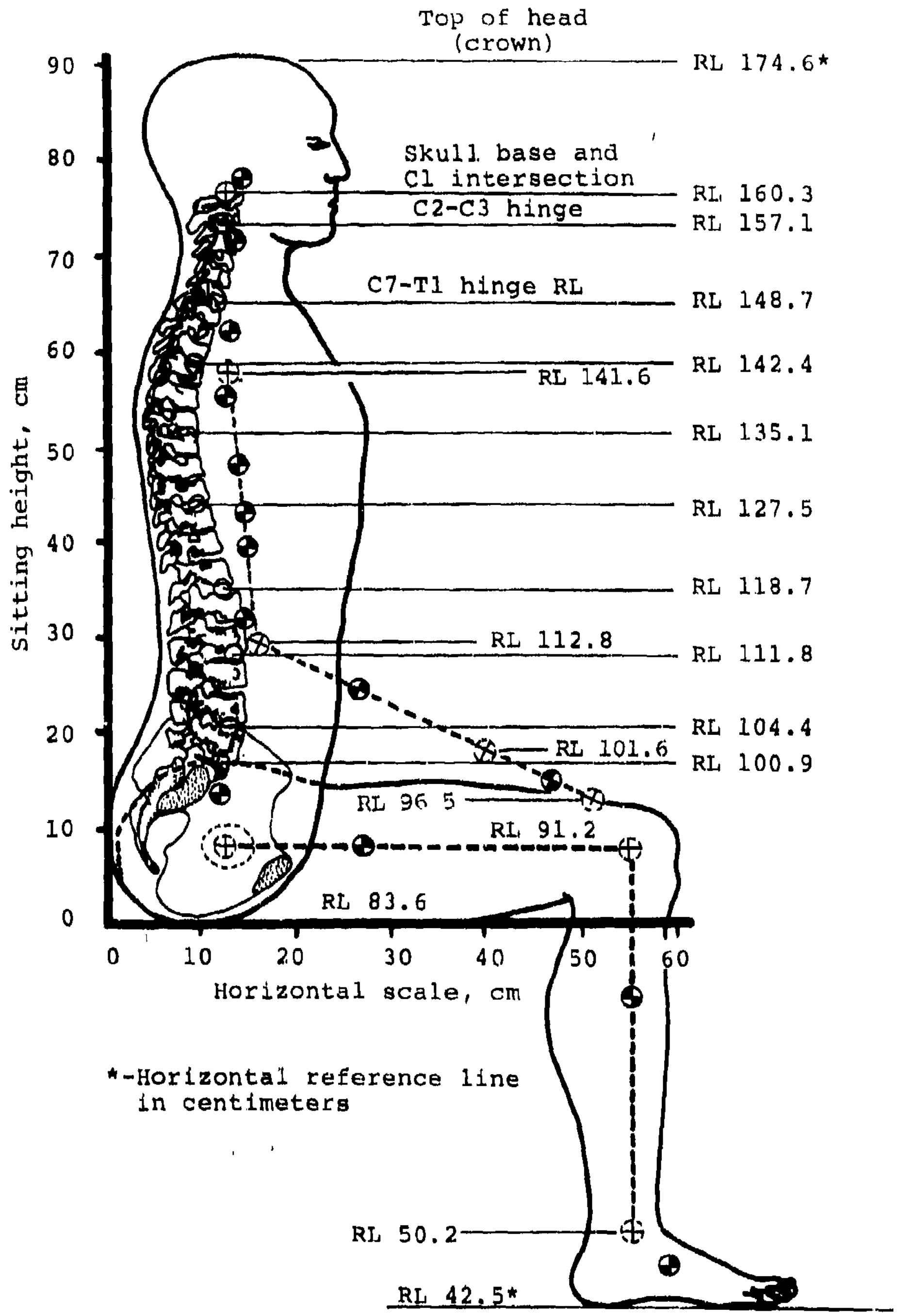


Figure 34. Sitting skeletal joint locations based on a 50th-percentile Army aviator. (From Reference 49)

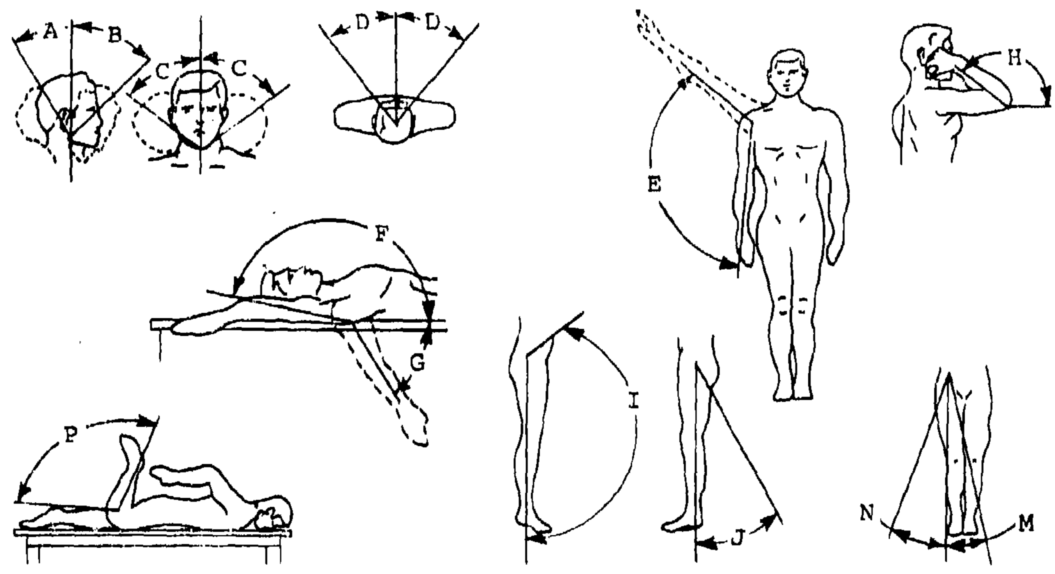


Figure 35. Joint ranges of motion.

TABLE 6. RANGE OF JOINT ROTATION (REFERENCE 62)

| Body component motion | Symbol | Motion description | Measured rotation (deg) Voluntary | Forced |
|---|---|---|---|---|
| Head - with respect to torso | A | Dorsiflexion | 61 | 77 |
| | B | Ventriflexion | 60 | 76 |
| | C | Lateral flexion | 41 | 63 |
| | D | Rotation | 78 | 83 |
| Upper arm - at shoulder | E | Abduction (coronal plane) | 130 | 137 |
| | F | Flexion | 180 | 185 |
| | G | Hyperextension | 58 | 69 |
| Forearm - at elbow | H | Flexion | 141 | 146 |
| Thigh - at hip | I | Flexion | 102 | 112 |
| | J | Hyperextension | 45 | 54 |
| | K | Medial rotation | - | - |
| | L | Lateral rotation | - | - |
| | M | Adduction | - | - |
| | N | Abduction | 71 | 79 |
| Lower leg - at knee | P | Flexion | 125 | 138 |

obtained by Glanville and Kreezer (Reference 62) for the movements defined in Figure 35.

## 6.3 INERTIAL PROPERTIES

Inertial properties of the human body have been used in design of escape systems, and moments of inertia of live subjects in a seated position were determined by Santischi, DuBois, and Omoto (Reference 63). However, anthropomorphic dummies and mathematical simulations require inertial properties of body segments, specifically moments of inertia, mass, and center-of-mass locations. Dempster reported on a study in which center-of-mass locations and moments of inertia with respect to transverse (y) axes were measured on segmented cadavers (References 60 and 61). Clauser, McConville, and Young determined center-of-mass locations and developed regression equations for cadaver segments (Reference 64). From data of various sources, the study presented in Reference 49 determined the segment masses and center-of-mass locations presented in Table 7 and Figure 36 for a 50th-percentile Army aviator. Chandler, et al., measured moments of inertia with respect to six axes for fourteen segments of six cadavers and, from them, calculated the principal moments of inertia for the segments (Reference 65). Their results are presented in Table 8.

62. Glanville, A. D., and Kreezer, G., THE MAXIMUM AMPLITUDE AND VELOCITY OF JOINT MOVEMENTS IN NORMAL MALE HUMAN ADULTS, Human Biology, Vol. 9, 1937, pp. 197-211.

63. Santischi, W. R., DuBois, J., and Omoto, C., MOMENTS OF INERTIA AND CENTERS OF GRAVITY OF THE LIVING HUMAN BODY, AMRL Technical Data Report 63-36, Aerospace Medical Research Lab, Wright-Patterson Air Force Base, Ohio, 1963.

64. Clauser, C. E., McConville, J. T., and Young, J. W., WEIGHT, VOLUME, AND CENTER OF MASS OF SEGMENTS OF THE HUMAN BODY, Antioch College; AMRL Technical Report 69-70, Aerospace Medical Research Lab, Wright-Patterson Air Force Base, Ohio, August 1969, AD 710622.

65. Chandler, R. F., et al., INVESTIGATION OF INERTIAL PROPERTIES OF THE HUMAN BODY, Report No. DOT-HS-801-430, U. S. Department of Transportation, Washington, D. C., March 1975.

TABLE 7. CENTER-OF-MASS DISTRIBUTION OF SEATED TORSO - 50TH-PERCENTILE ARMY AVIATOR (REFERENCE 49)

| Body segment identity | Segment mass (kg) | Z-axis location (cm)* | X-axis location (cm)** |
|---|---|---|---|
| Head | 4.74 | 77.6 | 10.1 |
| Neck (C1-C7) | 1.63 | 71.3 | 9.7 |
| Upper thoracic (T1-T3) | 4.07 | 62.4 | 10.0 |
| Upper mid thoracic (T4-T6) | 4.07 | 55.6 | 9.7 |
| Lower mid thoracic (T7-T9) | 4.66 | 48.1 | 11.2 |
| Upper arm | 4.44 | 43.5 | 12.2 |
| Lower thoracic (T10-T12) | 5.29 | 40.2 | 13.0 |
| Lumbar (L1 and L2) | 4.48 | 31.8 | 13.2 |
| Lumbar (L3 and L4) | 4.87 | 24.5 | 13.1 |
| Forearm | 2.62 | 24.6 | 23.7 |
| Lumbar (L5) | 2.52 | 19.0 | 12.2 |
| Hand | 0.92 | 15.3 | 45.1 |
| Pelvis | 8.89 | 13.0 | 11.2 |
| Thigh (hip) | 15.83 | 7.6 | 27.2 |
| Lower leg | 6.38 | -8.6 | 55.0 |
| Foot | 1.99 | -37.1 | 59.0 |
| TOTAL | 77.40 | | |

*Location is based on floor level of zero with 50th-percentile male head crown equal to reference line of 174.6 cm.
**Location is based on seat back with reference line equal to zero. Seat back is perpendicular to seat bottom, and torso touches seat back at head, shoulders, and buttocks.

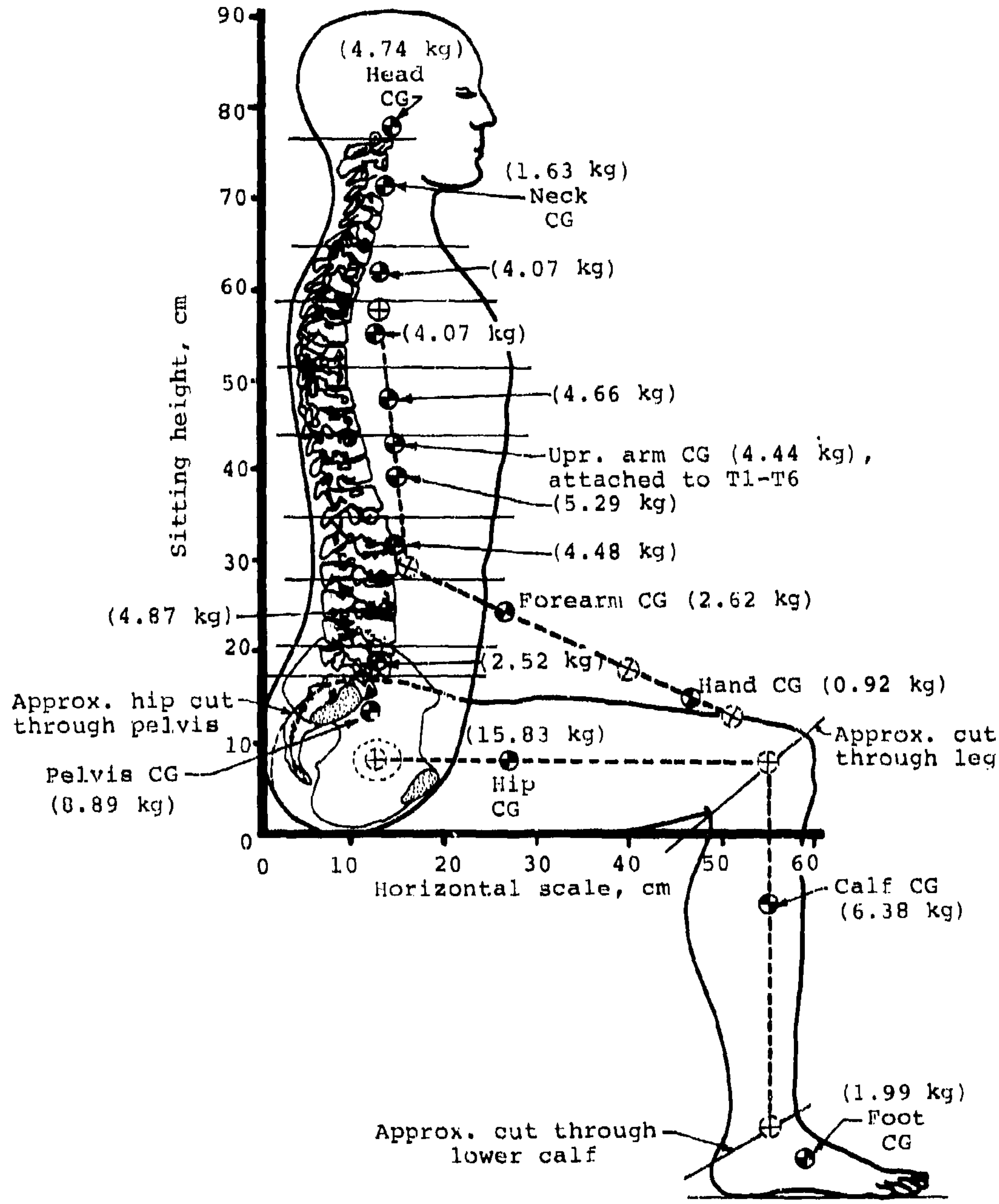


Figure 36. Mass distribution of seated torso referenced to the skeletal structure for a 50th-percentile Army aviator. (From Reference 49)

TABLE 8. SEGMENT MOMENTS OF INERTIA ABOUT THE CENTER OF MASS (REFERENCE 65)

| Body segment | Moment of inertia ($10^3$ gm-cm$^2$)* $I_{xx}$ | $I_{yy}$ | $I_{zz}$ |
|---|---|---|---|
| Head | 174.00 | 164.40 | 202.90 |
| Torso | 16,194.00 | 10,876.00 | 3,785.00 |
| Upper arm | 143.60 | 135.20 | 22.00 |
| Forearm | 65.80 | 63.80 | 8.70 |
| Hand | 7.19 | 5.86 | 1.97 |
| Thigh | 1,144.00 | 1,190.00 | 218.70 |
| Calf | 393.10 | 391.20 | 28.90 |
| Foot | 32.62 | 30.76 | 7.29 |

*Mean values of stature and weight reported to be 172.2 cm and 69.6 kg, respectively, for sample of six cadavers.

## 6.4 SCALING OF MEASUREMENTS

References 58 and 59 contain a significant volume of anthropometric data, whose statistics have been completely analyzed. In other words, the mean, standard deviation, and percentiles are listed for all measurements. However, the link lengths presented in Section 6.2.2 and the inertial properties presented in Section 6.3 are based on rather small samples. If the user of this guide wishes to scale the dimensions of Figures 34 or 36 to an occupant size other than the 50th-percentile Army aviator, it is recommended that the scaling be based on the most similar anthropometric dimension. For example, the lower leg length shown in Figure 34 is 41.0 cm. In order to convert this dimension to a 95th-percentile value, it would be multiplied by the ratio of popliteal heights from Table 4. The 5th-percentile lower leg length is then calculated as

$$L_{95} = \left(L_{50}\right)\left(\frac{PH_{95}}{PH_{50}}\right)$$

$$= \left(41.0\right)\left(\frac{18.3}{16.6}\right)$$

$$= 45.2 \text{ cm}$$

The center-of-mass location then can be determined for the new length using the same percentage of total segment length as shown in Figure 34. For example, according to Figure 34, the center of mass of the lower leg is approximately 38.8 percent of the distance down from the knee joint to the ankle joint. Retaining this percentage for the 95th-percentile occupant, the distance down from the knee joint to the lower leg center of mass is 0.388 (45.2) = 17.5 cm. Scaling segment masses presents a greater problem as there is no complete set of segment data for all size occupants. The simplest scaling approach would be to multiply the masses of Figure 36 by the ratio of total body masses (or weights) from References 58 or 59. For a more rigorous approach, the regression equations of Reference 64 might be used.

# 7. CRASH TEST DUMMIES

## 7.1 INTRODUCTION

The technology of crash test dummies has advanced significantly since 1968, when the first standard dummy was defined. Several designs currently are available, and many one-of-a-kind systems have been developed by various laboratories for their own use. However, for use in aircraft system evaluation, special consideration must be given to the effects of the vertical component of impact force, which make the aircraft crash environment quite different from that of an automobile, for which most dummies have been developed.

This chapter briefly outlines the evolution of current dummy technology, indicates the design features that are desirable for aircraft system testing, and summarizes research and comparative performance of dummies and humans.

## 7.2 DUMMY TECHNOLOGY

### 7.2.1 History of Dummy Development

An early dummy design by one of today's principal manufacturers was a rugged ejection seat dummy built by Sierra Engineering Company for the Air Force in 1949. According to Reference 66, this dummy had limited articulation and poor biomechanical fidelity, but it filled an important need, not only for aircraft system manufacturers but for the automobile industry as well.

A significant step toward the present anthropomorphic dummies was made by Swearingen, who, in 1949, needed a dummy better than the rigid test articles then available in order to evaluate explosive decompression for an aircraft cabin as a result of window failure. He designed a 120-lb dummy with articulated principal joints, realistic distribution of body weight, and centers of gravity approximating the human body. More than 500 blast tests were made to determine the hazard of explosive decompression and to provide means for preventing or reducing passenger injury. Placed in hundreds of positions, the dummy was blown against, into, and through simulated aircraft cabin windows. Most of these would have been fatal to a human and each test resulted in rather gross disintegration of the dummy.

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66. LeFevre, R. L., and Silver, J. N., DUMMIES - THEIR FEATURES AND USE, Proceedings, Automotive Safety Engineering Seminar, Society of Automotive Engineers, New York, June 20-21, 1973.

In 1951, Swearingen completed an improved dummy, capable of withstanding 35 to 50 G, which was used in evaluating a new safety harness for general aviation (Reference 67).

In 1954, Alderson Research Laboratories, Inc. created the first mass production dummy, unique for its modular design. The design permitted new parts to be added as needs changed and as knowledge grew over the subsequent decade. In 1967, both of the major dummy manufacturers marketed new devices that featured increased articulation in the vertebral column and shoulders, as well as increased chest compliance. These changes effected some improvement in biomechanical response but still fell far short of what is available today.

In 1968, SAE Recommended Practice J963 was published as a partial definition of a standard 50th-percentile male anthropometric test device (Reference 68). J963 recommends weights, center-of-gravity locations, dimensions for body segments, and the ranges of motion for body joints. Although moments of inertia and many design details were left unspecified, this was a first step toward a standard test device. Alderson upgraded its design to meet J963 in 1968 and 1971, while the Sierra counterpart appeared in 1970.

### 7.2.2 Part 572 Dummy

The role of the anthropomorphic dummy in automobile safety testing was formally changed in 1971 by the National Highway Traffic Safety Administration. Prior to that time, dummies had been used for determining relative performance of similar safety systems. The new law carried the implications that dummies must determine the absolute potential for injury to human occupants in an automobile crash and that different testing organizations should obtain the same results. The transition from relative to absolute measuring instrument forced the requirement for the dummy to be a standardized test instrument, as well as a reasonable simulation of a human being, since the legal performance limits are based on human tolerance data.

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67. Swearingen, J. J., DESIGN AND CONSTRUCTION OF A CRASH TEST DUMMY FOR TESTING SHOULDER HARNESS AND SAFETY BELTS, Civil Aeronautics Administration, Civil Aeronautics Medical Research Laboratory, (now FAA Civil Aeromedical Institute), Oklahoma City, Oklahoma, April 1951.

68. SAE Recommended Practice, ANTHROPOMORPHIC TEST DEVICE FOR USE IN DYNAMIC TESTING OF MOTOR VEHICLES, SAE J963, SAE Handbook, Society of Automotive Engineers, Warrendale, Pennsylvania, 1978, pp. 34.107-34.110.

In 1972, General Motors Corporation produced the Hybrid II dummy, a 50th-percentile male anthropomorphic test device. This dummy utilizes torso and limbs from the Alderson VIP-50A dummy with modifications made to the chest to allow increased deflection and damping. The head assembly was adapted from the Sierra 292-1050 design with several anatomical modifications. Both the neck and lumbar spine consist of a butyl rubber cylinder, the latter being reinforced by an internal steel cable.

Along with a number of other modifications, the design of the Hybrid II formed the basis for the Code of Federal Regulations, Title 49 (49CFR) Part 572 specification for dummies (Reference 69). Its specified dimensions and inertial properties are displayed in Figure 37 and Tables 9 through 11. Segment moments of inertia reported in Reference 70 for a Hybrid II dummy are listed in Table 12.

### 7.2.3 Other Recent Dummy Designs

Various organizations involved in safety research and engineering have produced their own dummy designs through modification of production devices, as was done in the Hybrid II development discussed in Section 7.2.2. These dummies have generally been designed to yield improved simulation of human response, but none have been officially standardized for manufacturing as in the case of the Part 572 system. They are briefly summarized here only to provide background information regarding the state of the art.

"Repeatable Pete" was developed by the Highway Safety Research Institute to accurately match the dynamic response of unembalmed human cadavers, particularly the head acceleration, for

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69. Code of Federal Regulations, ANTHROPOMORPHIC TEST DUMMY, Title 49, Chapter 5, Part 572, Federal Register, Vol. 38, No. 62, April 2, 1973, pp. 8455-8458.

70. Massing, D. E., Naab, K. N., and Yates, P. E., PERFORMANCE EVALUATION OF NEW GENERATION OF 50TH-PERCENTILE ANTHROPOMORPHIC TEST DEVICES: VOLUME I - TECHNICAL REPORT, Calspan Corporation; DOT-HS Technical Report 801-431, U. S. Department of Transportation, National Highway Traffic Safety Administration, Washington, D. C., March 1975, PB 240-920.

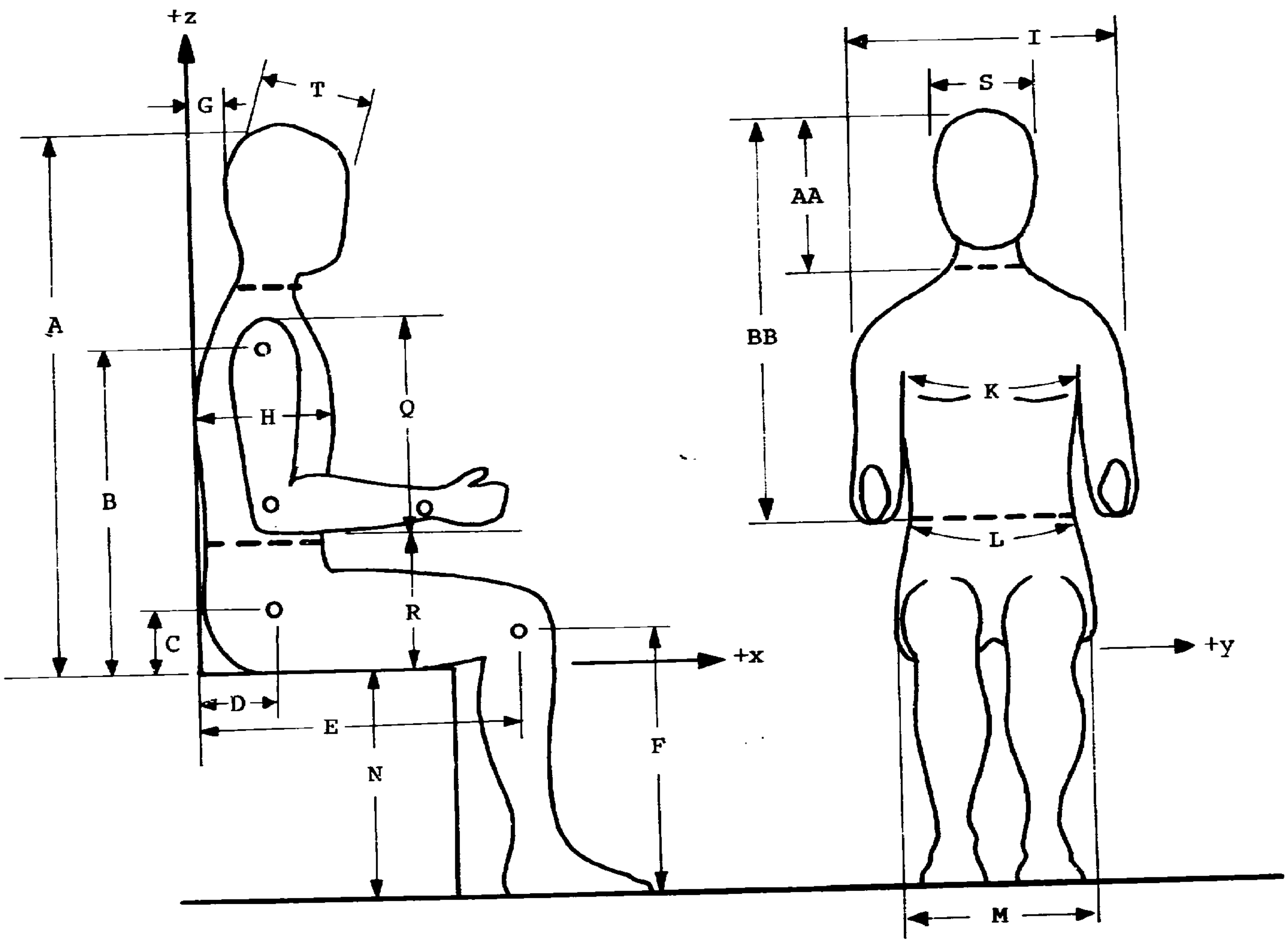


Figure 37. Dummy external dimensions.

TABLE 9. DUMMY EXTERNAL DIMENSIONS (PART 572)

| Designation | Figure 37 code | Part 572 specification (in.) |
|---|---|---|
| Seated height | A | 35.7 ± 0.1 |
| Shoulder pivot height | B | 22.1 ± 0.3 |
| Hip pivot height | C | 3.9 |
| Hip pivot from back line | D | 4.8 |
| Knee pivot from back line | E | 20.4 ± 0.3 |
| Knee pivot from floor | F | 19.6 ± 0.3 |
| Head back from back line | G | 1.7 |
| Chest depth | H | 9.3 ± 0.2 |
| Shoulder width | I | 18.1 ± 0.3 |
| Chest circumference over nipples | K | 37.4 ± 0.6 |
| Waist circumference at minimum girth | L | 32.0 ± 0.6 |
| Hip width | M | 14.7 ± 0.7 |
| Popliteal height | N* | (17.3 ± 0.2) |
| Shoulder-elbow length | Q* | (14.1 ± 0.3) |
| Elbow rest height | R* | (9.5 ± 0.5) |
| Head width | S* | (6.1 ± 0.2) |
| Head length | T* | (7.7 ± 0.2) |
| Head segment line | AA | 9.3 |
| Shoulder-thorax segment line | BB | 25.1 |

*Added to Part 572 data, SAE specification value in parentheses.

TABLE 10. DUMMY COMPONENT WEIGHTS (PART 572)

| Segment | Part 572 specification (lb) |
|---|---|
| Head | 11.2 ± 0.1 |
| Upper torso (including lumbar spine) | 41.5 ± 1.6 |
| Lower torso (including visceral sac and upper thighs) | 37.5 ± 1.5 |
| Upper arm | 4.8 ± 0.2 |
| Lower arm | 3.4 ± 0.1 |
| Hand | 1.4 ± 0.1 |
| Upper leg | 17.6 ± 0.7 |
| Lower leg | 6.9 ± 0.3 |
| Foot | 2.8 ± 0.1 |
| Total dummy (including instrumentation in head, torso, and femurs) | 164.0 ± 3.0 |

TABLE 11. CENTER-OF-GRAVITY LOCATIONS (PART 572)

| Segment | x and z reference origin | Part 572 specification x (in.) | Part 572 specification z (in.) |
|---|---|---|---|
| Head | Back and top of head | +4.0 ± 0.2 | -4.7 ± 0.1 |
| Upper torso | Backline and top of head | +4.1 ± 0.3 | -17.2 ± 0.3 |
| Lower torso and upper thigh | Backline and top of head | +4.9 ± 0.5 | -31.0 ± 0.5 |
| Upper arm | Shoulder pivot | 0.0 ± 0.3 | -5.0 ± 0.3 |
| Lower arm | Elbow pivot | +4.2 ± 0.3 | 0.0 ± 0.3 |
| Hand | Wrist pivot | +2.2 ± 0.3 | 0.0 ± 0.3 |
| Upper leg | Knee pivot to upper leg rotation center | -6.7 ± 0.3 | 0.0 ± 0.3 |
| Lower leg | Knee pivot to ankle pivot | 0.0 ± 0.3 | -8.0 ± 0.3 |
| Foot | Ankle pivot | +2.2 ± 0.3 | -1.7 ± 0.3 |

NOTE: Axis system is shown in Figure 37 (drawing of external dummy dimensions), using +x forward and +z up.

TABLE 12. HYBRID II MASS MOMENTS OF INERTIA (REFERENCE 70)

| Body segment | Moment of inertia (in.-lb-sec$^2$) | | |
|---|---|---|---|
| | $I_x$ | $I_y$ | $I_z$ |
| Head | 0.226 | 0.275 | - |
| Head/neck | 0.310 | 0.367 | 0.233 |
| Upper torso (includes lumbar spine) | 2.18 | 1.79 | - |
| Lower abdomen, pelvis, and visceral sac | 2.32* | 1.73* | - |
| Right upper arm | 0.134 | 0.132 | 0.022 |
| Right forearm (no hand) | 0.012 | 0.068 | 0.071 |
| Right upper leg | 0.127 | 0.873 | 0.890 |
| Right lower leg (no foot) | 0.599 | 0.575 | 0.359 |

*Included lumbar spine section.

NOTES: 1. Instrumentation was installed in the head, chest, and femurs during the measurements.
2. Estimated accuracy of measurements: ±3 percent.

more viable use in determination of head injury criteria (Reference 71). General Motors used detailed human anthropometric data in designing the GM ATD 502 dummy (Reference 72).

71. McElhaney, J. H., Mate, P. I., and Roberts, V. L., A NEW CRASH TEST DEVICE - "REPEATABLE PETE," Proceedings, Seventeenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1973, p. 467.

72. Hubbard, R. P., ANTHROPOMETRIC BASIS OF THE GM ATD 502 CRASH TEST DUMMY, Paper 750429, Society of Automotive Engineers, Warrendale, Pennsylvania, 1975.

A distinguishing feature of the ATD 502 design is the location of the vertebral column for an average adult male in an erect seated posture. In addition, General Motors has reported development of the Hybrid III dummy (Reference 73). Based on the ATD 502, this design includes transducers for measurement of neck loads and chest deflections.

The Transport and Road Research Laboratory in England has developed a specialized side impact dummy for use in side impact testing of automobiles (Reference 74). The segment joints are designed for lateral motion, and seven force transducers are used to measure the load distribution between the occupant and the vehicle interior.

## 7.3 COMPARISON OF DUMMY AND HUMAN RESPONSE

There are two basic questions regarding the use of a mechanical system such as a dummy to evaluate the degree of protection a vehicle system would afford its human occupants. First, how closely does the dummy response simulate human response? Secondly, how does performance vary from one dummy to another and from one test laboratory to another?

Finding an answer to the first question presents a problem. The response of live human subjects can, of course, be determined only at safe acceleration levels, substantially below crashworthiness design conditions. The response of human cadavers at higher acceleration levels has been used in dummy design, but questions do exist in the quality of simulation provided by cadavers. Recently, Walsh and Romeo reported on a series of sled tests and full-scale car crash tests wherein fresh, unembalmed cadavers and dummies were exposed to identical environments (Reference 75). Both belt restraints and air

73. Foster, J. K., Kortge, J. O., and Wallanin, M. J., HYBRID III-A BIOMECHANICALLY-BASED CRASH TEST DUMMY, Proceedings, Twenty-First Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1977, pp. 973-1014.

74. Harris, J., THE DESIGN AND USE OF THE TRRL SIDE IMPACT DUMMY, Proceedings, Twentieth Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, Pennsylvania, 1976, pp. 75-106.

75. Walsh, M. M., and Romeo, D. M., RESULTS OF CADAVER AND ANTHROPOMORPHIC DUMMY TESTS IN IDENTICAL CRASH SITUATIONS, Proceedings, Twentieth Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, Pennsylvania, 1976, pp. 108-131.

bags were used. Although the overall kinematic response between cadaver and dummy agreed fairly well, some injuries that could not have been detected with the dummy were observed in the cadaver.

Several test programs have been conducted to compare the dynamic response of different dummy designs. Chandler and Christian demonstrated that, as dummies become more complex, the number of test variables may exceed those that the experimenter can control (Reference 76). The requirement for standard test practices also was noted. Massing, Naab, and Yates compared several dummies in tests with either belt restraints, air bags, or energy-absorbing steering columns (Reference 70). Some of the sled tests were repeated at two different laboratories. As an example of the results, mean head resultant accelerations for ten repeated tests on each of five dummies with belt restraints are shown in Figure 38. Dummies D5 and D6 are both GM-50X designs (Reference 72), and A1 and A2 are Highway Safety Research Institute (HSRI) devices (Reference 71). Figure 39 shows the mean head accelerations for ten repeated tests with the same HSRI dummy conducted at two different facilities, the FAA Civil Aeromedical Institute (CAMI) and Calspan Corporation. Because only one dummy was tested at the two different facilities, data are insufficient to permit generalization with respect to comparative performance. However, Figure 39 indicates that differences in performance did exist. With respect to dummies tested at the same facility, differences in repeatability were noted. The HSRI dummies were found to be less repeatable than the GM 50X dummies, and both of these were found to be less repeatable than the Hybrid II for the air bag restraint. As shown in Figure 38, differences do exist among the performance characteristics of all the dummies. Detailed analyses of the data are presented in Reference 70.

## 7.4 SUITABILITY OF DUMMIES FOR AIRCRAFT SYSTEM EVALUATION

All of the recently developed dummies described in Section 7.2 were designed for automotive testing. All are based on the anthropometry of a 50th-percentile U. S. civilian male. In dynamic testing of an energy-absorbing seat, design for aircraft occupant weight can play a critical role. It would be desirable, although generally not practical, to evaluate a seat for a range of occupant sizes. A 95th-percentile dummy would verify the strength of the seat structure and restraint system

76. Chandler, R. F., and Christian, R. A., COMPARATIVE EVALUATION OF DUMMY PERFORMANCE UNDER $-G_x$ IMPACT, Proceedings, Thirteenth Stapp Car Crash Conference, Society of Automotive Engineers, New York, 1969, pp. 61-75.

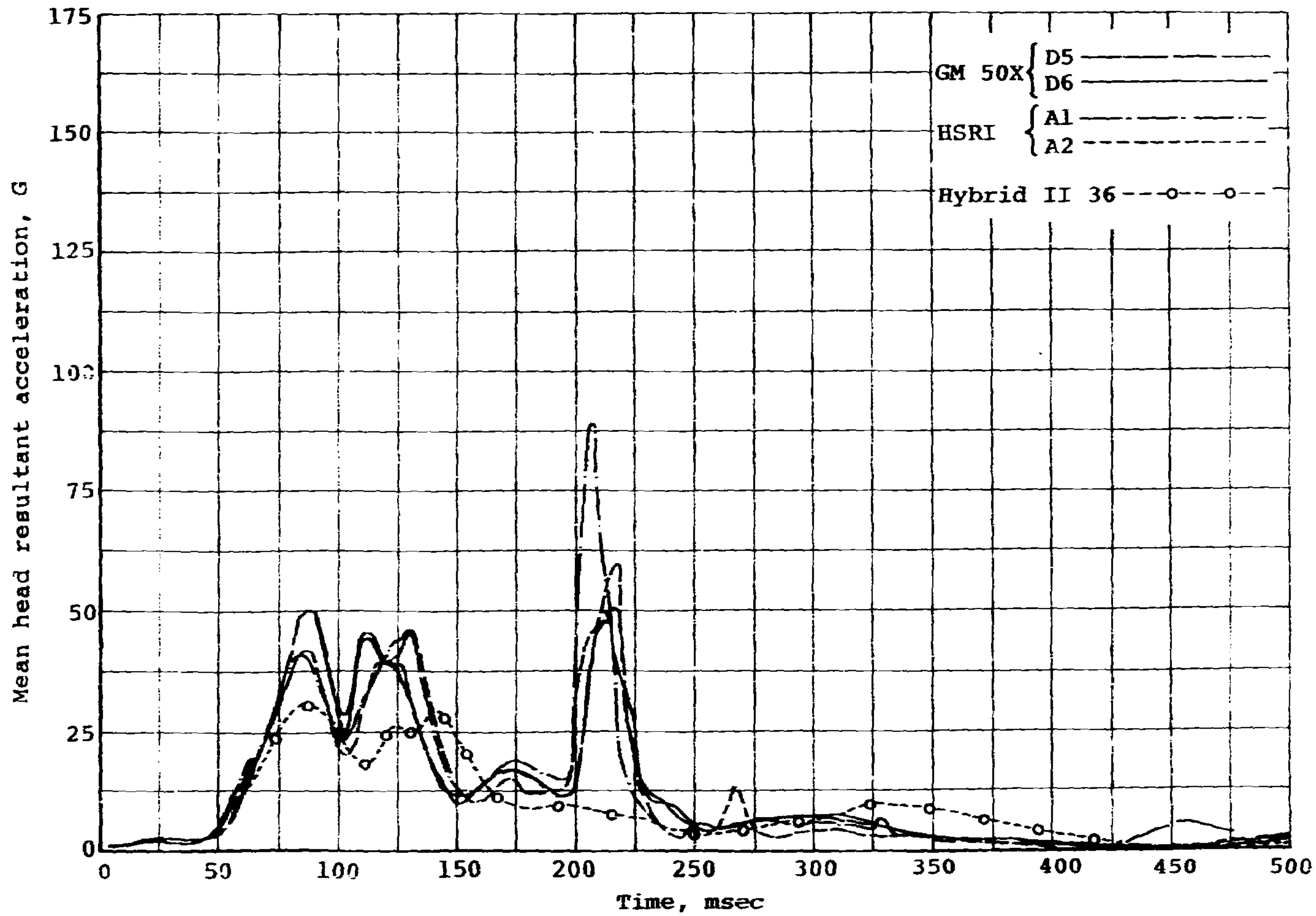


Figure 38. Comparison of mean head resultant acceleration responses for three different dummy designs. (From Reference 70)

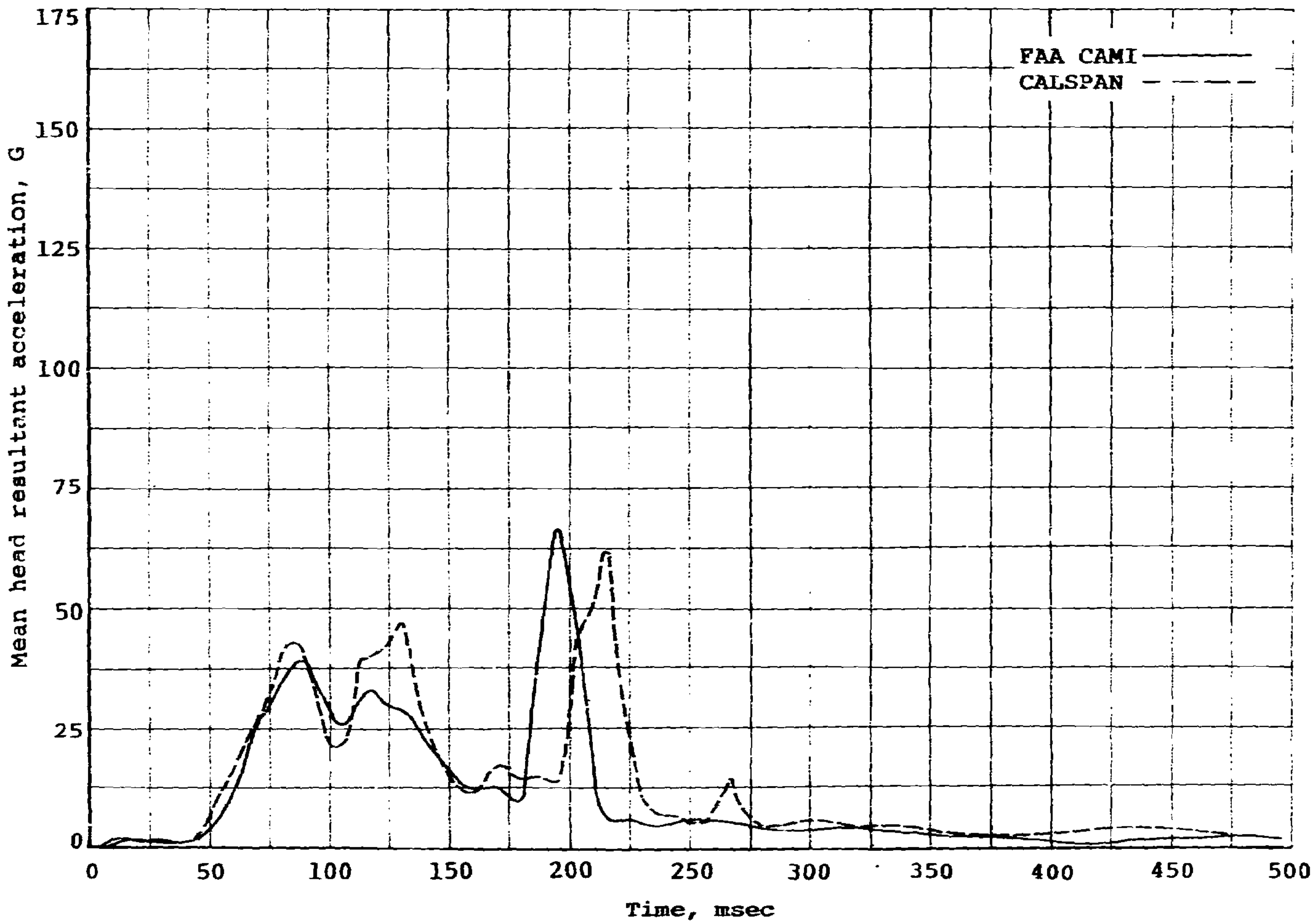


Figure 39. Comparison of mean head resultant acceleration responses for HSRI dummy conducted at two laboratories. (From Reference 70)

as well as the adequacy of the energy-absorbing stroke. Testing with a 50th-percentile dummy would demonstrate the performance of the system for an occupant of average height and weight. A 5th-percentile dummy would probably experience accelerations of higher magnitude and would establish the severity of a given set of impact conditions for the smaller occupant. However, both the expense of dummy purchase and the cost of conducting dynamic tests may make such a test program impractical. An alternative procedure might be to establish the occupant protection capability of a seat design by analysis and to conduct a dynamic test with a 95th-percentile dummy to verify system strength.

There are two additional factors that should be considered in dummy selection for aircraft seat testing. First, some designs are more suitable than others for testing with a headward ($+G_z$) acceleration component. None of the dummies have been designed for accurate response to vertical impact. The spinal column, which is a critical region of human tolerance to aircraft crash loading, has been designed to simulate response to $-G_x$ loading rather than the more critical $+G_z$ direction. However, the reinforced rubber cylinder used as the lumbar spine in the Part 572 dummy permits more consistent positioning than the steel ball-and-socket configuration used in some other dummies. Instability in the latter type could affect response of the upper torso with concomitant penalties on test repeatability. Another advantage of the Part 572 dummy for aircraft seat testing is a humanlike pelvic structure, which should result in load distribution on the cushion close to that for a human. Secondly, if the results of tests conducted at different facilities are to be compared, standardization of dummies and test procedures is mandatory.

At present, it seems that use of the Part 572 dummies, modified to improve their simulation accuracy to impact loading in the $+G_z$ direction and sized to 5th-, 50th-, and 95th-percentile versions of the U. S. Army aviator, provides the recommended approach.

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## INDEX

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US ARMY RESEARCH AND TECHNOLOGY LABORATORIES (AVRADCOM)
FORT EUSTIS, VIRGINIA 23604

## READER'S SERVICE LETTER

(For use in submitting comments, recommendations, corrections, and revisions for *Aircraft Crash Survival Design Guide*)

FROM:

TO: Director, Applied Technology Laboratory, US Army Research and Technology Laboratories (AVRADCOM), ATTN: DAVDL-ATL-ASV, Fort Eustis, Virginia 23604

REMARKS:

SIGNATURE: ____________________ DATE: ________________________

DEPARTMENT OF THE ARMY
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US ARMY RESEARCH AND TECHNOLOGY LABORATORIES (AVRADCOM)
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SUPPLEMENTARY

INFORMATION

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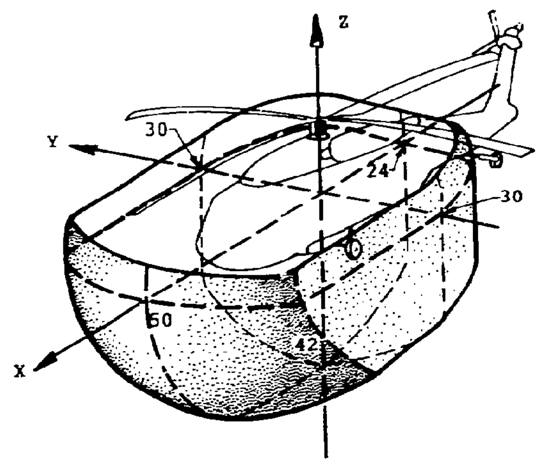
## ERRATUM

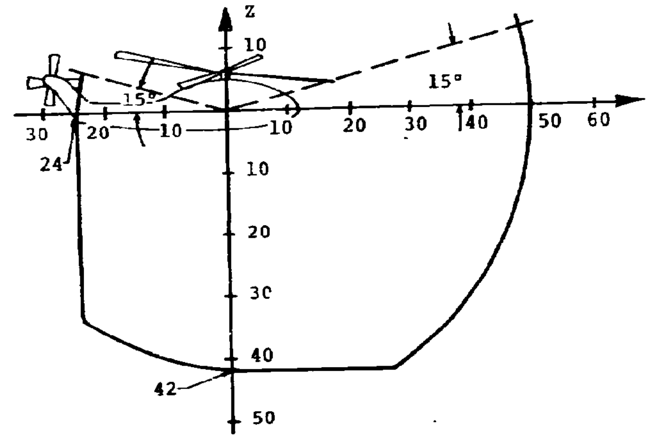
USARTL-TR-79-22B

TITLE: Aircraft Crash Survival Design Guide
Volume II - Aircraft Crash Environment and Human Tolerance

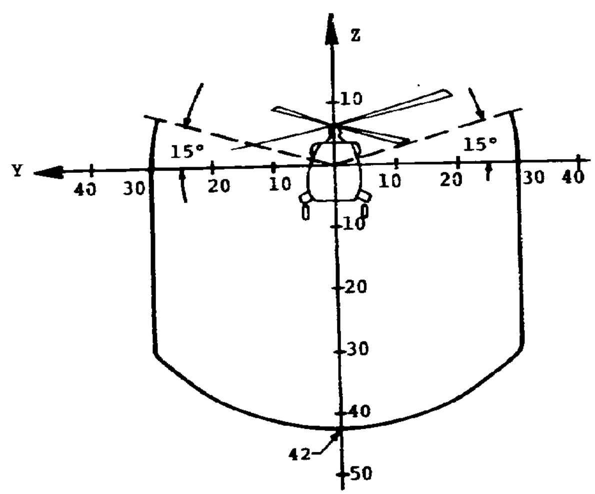
Insert revised page 28.




(a) Design velocity changes (three-dimensional display)




(b) Longitudinal-vertical

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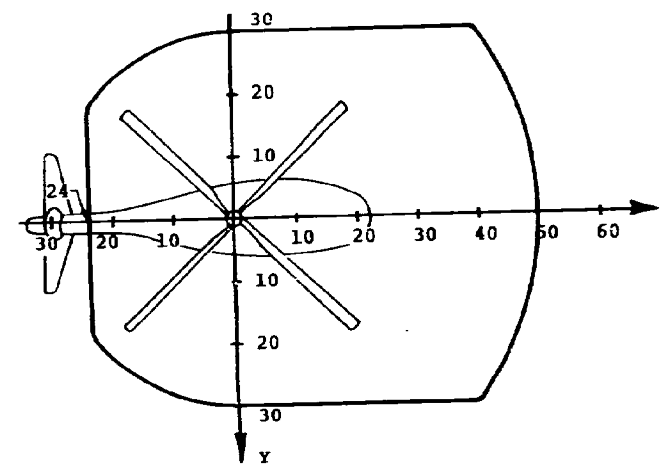


(d) Longitudinal-lateral

Figure 6. Design velocity changes - off-axis requirements.

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# ERRATA

USARTL TR 79-22B

TITLE: Aircraft Crash Survival Design Guide
Volume II - Aircraft Crash Environment and Human Tolerance

Make the following changes:

Page 54, first paragraph --

Change (Reference 24) to (Reference 77)

Page 114 -- Add:

77. Slobodnik, B. A., SPH-4 HELMET DAMAGE AND HEAD INJURY CORRELATION, USAARL Report No. 80-7, U. S. Army Aeromedical Research Laboratory, Fort Rucker, Alabama, September 1980.