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Chapter 6

IMPACT ACCELERATIONS[1]

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IMPACT HAZARDS

Impact acceleration may be encountered during normal as well as emergency phases of spacecraft operations. The impact loads experienced during normal flight phases occur primarily on spacecraft landing upon its return to Earth. Recovery systems used or considered for spacecraft designs have included conventional single and multiple parachute canopies, retrorockets, inflated fabric spheres, and parawings or parasails. Within existing technology and primarily weight limitations, it has not been practical to allow descent and horizontal drift rates to be adequately controlled within the range of impact velocities that would not be hazardous to the crew under all adverse circumstances. Exact knowledge of the physical environment to which the astronauts might be exposed with a particular spacecraft and its recovery system for all potential environmental variables, that is, impact surface, wind, impact angle, and others, is absolutely essential for realistic risk analysis and evaluation of protective requirements.

The severity of the impact experienced during spacecraft landing can be reduced considerably by controlling the landing site. With such control, the impact surface and wind conditions that are most favorable may be selected. Water, or flat, soft terrain have generally proven to produce less severe impacts. Data are available on the dynamics of water impact and, more specifically, on the water and land impact characteristics of the Apollo spacecraft [5, 98]. The descent rate at impact may range up to 8.5 m/s if the recovery system deploys properly. In a design such as the Apollo spacecraft where three recovery parachutes were used, the descent velocity could be as high as 15.2 m/s. The resulting impact pulses under even nominal conditions are typically high amplitude, short rise time accelerations, which are shown in Figure 1.

If a catastrophic failure occurs on the launch pad during final portions of the preflight preparations, short-duration, high-amplitude acceleration may be required to catapult the space vehicle crewman safely away from the launch vehicle. This same emergency escape system may be required during the initial phase of launch vehicle acceleration if there is failure of the propulsion or guidance systems. The acceleration environment associated with use of the escape system is more complex as the launch vehicle achieves higher velocities while it is still within the Earth's atmosphere. This more complex environment is due to interaction with the windstream and rapid deceleration of the escape system immediately after separation from the launch vehicle. In addition, the

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impact of the opening of the recovery parachute may be quite severe at these higher airspeeds.

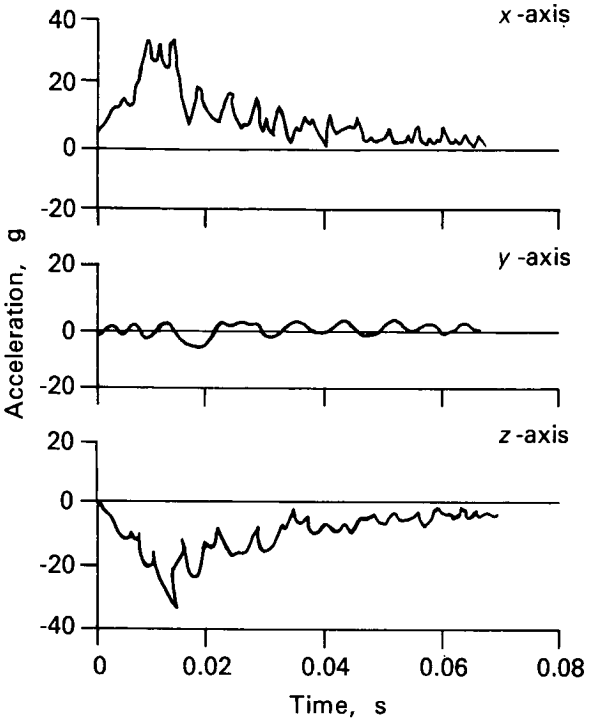

FIGURE 1.—Accelerations recorded during impact tests of the Apollo Crew Module [98]. Impact occurred at a pitch attitude of −27.5°, a roll attitude of 0°, a horizontal velocity of 11.4 m/s, and a vertical velocity of 10.5 m/s.

Two basic types of emergency escape systems have been used to assure spacecraft crew safety. The impact environments associated with each type are different in many aspects. The first type, the individual ejection seat which is used in high-speed aircraft, generates short-duration acceleration pulses throughout its entire sequence. These pulses are created during ignition of the ejection catapult, firing of the sustainer rocket, impact with a high-velocity airstream, parachute opening shock, and landing impact. The second type of escape technique involves propulsion of the entire spacecraft away from the launch vehicle. The catapult acceleration required with this type of escape system is generally of lower magnitude, usually no greater than 8 to 12 g, and longer duration than that required if an ejection seat is used. The alinement of the propulsion system thrust vector and center of gravity of the vehicle are more easily controlled than when the conventional ejection seat is used. A large portion of the ejection velocity must be imparted to the ejection seat while it is still stabilized by ejection rails, because of this problem. Therefore, the ejection acceleration may be as high as 18 to 20 g.

When the entire spacecraft is used as an escape system, it causes two other notable differences in the impact environments. The first, a beneficial difference, is elimination of the problem of impact with the windstream and rapid deceleration. The spacecraft is generally optimally designed for aerodynamic deceleration upon reentry into the Earth's atmosphere, and thus, the deceleration forces tend to be low. The second difference occurs at landing. Landing without the spacecraft is usually accomplished without incident by a properly trained crewman. A crewman descending under a personal parachute may judge his drift rate and even control his direction of drift, whereby he can position himself and use his legs to minimize landing impact effects. A difficult design problem is assuring an equally safe landing of the spacecraft under emergency and even normal conditions.

The relatively complex tasks performed by an individual prior to a parachute landing (which are not easily accomplished without adding undesirable weight and complexity to the spacecraft) are, sensing drift rate and direction and alining himself to obtain the best use of his legs to attenuate the impact. Impact accelerations that are experienced during capsule landing impact are quite variable due to lack of control of these factors. Furthermore, variabilities of the spacecraft structural rigidity, stiffness and contour of impact surface, and oscillation induced by the recovery parachute, coupled with the possibility of multiple impacts in different directions, add to the difficulty of providing a safe landing.

An escape system composed of several of the most desirable attributes of each of the basic escape system approaches represents another alternative. This approach uses the spacecraft to achieve separation from the launch vehicle

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but individual ejection seats are also used after the required separation distance is achieved, and spacecraft velocity has decayed to an acceptable level. This approach avoids ground-landing impact problems associated with crew recovery within the spacecraft; however, it may not be the most effective approach in terms of spacecraft weight and complexity except in cases where there is no requirement to recover the spacecraft.

Impact environments may also be encountered during other portions of the space mission. For example, acceleration associated with spacecraft docking operations, that is, coupling the spacecraft to another spacecraft or propulsion unit, will result in transient acceleration. The ground-landing problem in spacecraft recovery after mission completion or emergency escape is also present during extraterrestrial landing. The impact environments of docking and extraterrestrial landing must necessarily be mild, to prevent injury to the crew or spacecraft equipment that might compromise success of the venture.

Each potential or actual impact hazard associated with the mission must be assessed to determine the degree of risk that may be allowed for injury or equipment failure. A mission risk analysis of this type cannot be carried out without a relatively detailed understanding of the human response to each level of impact stress. A primary objective of research in this technology area has, thus, been the development of human exposure limits in terms suitable for such a risk analysis.

Definition of Impact

Impact is generally defined as an acceleration with a pulse duration of not more than 1 s. The acceleration-time history is defined in terms of its magnitude in m/s^2 or usually in g units and its time parameters. Included in time parameters are rise time (duration from start of acceleration to peak acceleration time), and pulse duration (total time of the individual pulse). Acceleration derivatives such as rate of onset of acceleration (g/s) and rate of offset of acceleration are also commonly used as descriptors. However, it must be kept in mind that these descriptors give approximations only to the true acceleration-time history and that the limits within which they are meaningful must be examined.

For purposes of frequency domain analysis, an impact pulse is composed of energy density distributed over a spectrum of frequencies. Thus, a particular acceleration-time history may be reduced to terms of the power spectral density.

Impact accelerations might occur as linear or rotational accelerations, all together in 6 degrees of freedom.

Terminology used in the study of the human response to impact is varied [19, 48, 49, 87, 88, 96]; however, terminology that is generally understood has been selected for this discussion. Terms such as "overload" used in USSR literature and "dynamic overshoot" used in US literature are not used, in order to permit a more universal understanding of the text. The direction of linear and rotational acceleration vectors is defined with respect to the human body by use of the coordinate system shown in Figure 2, which is standardized for biomechanics.

Physiologic and Pathologic Effects of Linear Impact

Most human impact research has been conducted in connection with general automotive or aviation crash research, not in support of specific space requirements. Impact exposures experienced during emergency escape maneuvers have been studied during the last 30 years in connection with emergency escape from aircraft. Impact situations similar to space capsule landing impacts have been of interest for the last 15 years in the development of aircraft capsule escape systems [12].

Primary physiologic and pathologic effects of impact are caused by localized pressures and resulting relative displacements of body tissue. Massive stimulation of the entire nervous system in an extremely short time results in various sensations and reactions immediately after impact due to activation of pressure and stretch receptors. These sensations will vary in magni-

tude depending on the magnitude of the insult and will vary in seriousness from momentary stunning and mild cardiovascular reactions to cardiovascular shock, unconsciousness, and concussion—the latter probably always connected with pathologic injury. Direct injuries to body tissues result when relative displacements of body tissue exceed mechanical stress limits of the particular tissue involved.

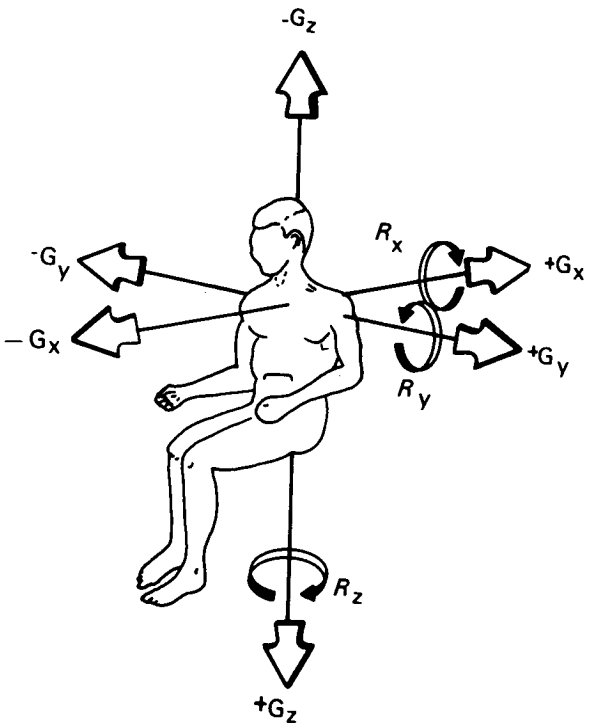


FIGURE 2.—Coordinate system for description of impact inertial response vector direction.

Such injuries may be at a cellular or subcellular level with no gross evidence of shear, tensile, and/or compressive stresses. Damage due to blood movement has not, in general, been observed, although conjunctivitis and retinal symptoms observed in $-G_x$ impact may be related to this phenomenon. This type of injury would not be expected for very short duration impact since the duration of exposure to acceleration is too brief to allow significant shifting of blood volumes. Injuries may also occur with more acute pathologic effects such as: abrasions, contusions, hematomas and laceration of soft tissues; strains, sprains, failure of cartilaginous structure and joint derangement; and various fractures or subluxations within the skeletal system. The seriousness of these injuries will vary from simple, reversible disabilities to chronic, irreversible impairment of anatomic structures or physiologic functioning of the body, and major trauma which may be either immediately or eventually fatal.

Physical response of the body and its organs, i.e., stress distribution along the body and stress severity, is dependent upon the acceleration-time history of the impact environment. Other major factors influencing the response include acceleration direction [52, 66], restraint degree [68], and body condition, that is, age, physical state, and others [44, 82]. The pathologic manifestations described rely heavily on analysis and interpretation of aviation, automotive, sport, and home accident data as well as data collected from suicides [91, 92]. Causes and mechanisms leading to these effects are derived from low-level noninjurious human tests or animal experiments.

Research conducted so far has shown different mechanisms of injury and symptoms for each impact direction studied. Much information that is available on these injury mechanisms has been collected from studies of accidents as well as laboratory experiments. In accident situations, head injury is the most frequent and most severe manifestation [74].

Head Injuries

More than 75 percent of aircraft crash fatalities result from head injuries. These injuries usually result from heavy blows to the head rather than from acceleration of the head structure as a whole [2, 3]. Neck injury from indirect acceleration of the head is not as well understood; these injuries may include ligamentous, disk, and vertebral damage as well as involvement of the spinal cord. Concussion may result from either neck hyperflexion or hyperextension if the head is not supported during impact [18, 43, 70]. Other concussion types are observed after concentrated

blows to the head that deform or fracture the skull [31, 32, 60] and cause deformations throughout the brain tissue [10, 33, 55, 67].

Head impact studies have been conducted with anesthetized monkeys and dogs to relate the severity and duration of concussion to intracranial pressure change and its duration [34]. Pressure changes were recorded with intracranial pressure transducers. Moderate to severe concussion effects were observed in the range of 2.1–6.3 kg/cm^2 intracranial pressure change concurrent with head impact. Observations of concussion effects in humans are limited to clinical investigations. Accuracy of such studies is greatly limited by the ability to estimate impact conditions associated with trauma.

Case histories of 317 patients with head injury have been studied and related to impact velocity, impact direction, and rigidity of the impact surface [75]. Severity of the trauma was evaluated on the basis of the patient's condition immediately after impact and during the affliction. Severity of the trauma resulting from impact velocities in the range of 3.0 to 10.5 m/s depended, to a considerable degree, on occurrence of fractures in the cranial base and degree of pathogenic involvement of the intracranial structures adjacent to the cranial base. Frontal region impacts resulted in a less severe clinical picture than impacts in the temporal and occipital regions. Cranial base fractures and damage to adjacent cerebral tissues almost always resulted if the impact velocity was greater than 5.0 m/s.

Spinal Impact

Damage to the vertebral column is a common mechanism of injury where the impact is applied parallel to the spine in the $+G_z$ direction such as in seat ejection maneuvers [17, 102]. Compression fracture of individual vertebral bodies is frequently observed in radiographic examination of individuals who have used aircraft ejection seats [38]. These fractures are usually confined to the upper lumbar and lower thoracic areas of the vertebral column. Although such injuries to the upper thoracic and cervical spine are relatively uncommon, they are observed when the ejecting crewman is poorly positioned prior to ejection. Immediate symptoms of this injury may range from slight, to severe, incapacitating pain. Ileus, persistent neuralgic and sciaticlike pains are common lingering symptoms. Compression fractures or fractures of spinal processes may, in extreme cases, be sufficiently extensive to result in intrusion of bone fragments or the disk into the spinal cord canal. Such instances may result in paralysis or other neurologic symptoms.

Physiologic and pathologic effects of impact in the $-G_z$ direction have not been identified in humans [17]. Investigators have speculated that intracranial hemorrhage would be the limiting factor, on the basis of results of longer duration acceleration experiments conducted on centrifuge facilities. However, impact tests with animal subjects have not supported this theory. Dogs exposed to accelerations up to 15 g from a duration of 0.05 s and 7 g for up to 1 s showed only petechial hemorrhages, generally in the mucous sinus membranes. Autopsy of dogs revealed no indication of intracranial hemorrhages. Experiments with volunteers have been limited to tests required to support development of the downward ejection seat and evaluation of Project Apollo crew protection designs [9, 40].

Transverse Impact

When impact is transverse to the longitudinal axis of sitting, in a well-supported and restrained body, the first signs of limiting human tolerance [94] have been various degrees of shock, i.e., pallor, perspiration, and transient elevation and subsequent drop of blood pressure. In one test, brief attacks of low blood pressure and albuminuria were observed for about 6 h after impact. More severe impacts will result in unconsciousness. Effects of maximum voluntarily tolerated impact levels were at times not pronounced, but delayed effects occurred with gradual onset in the following 24-h period. Subtolerance impact exposures in this axis normally cause elevation of pulse rate to approximately 150–170 pulses/min with respiration rate of 30–40 breaths/min followed by a rapid drop in these rates. Upon repeated exposures, the degree of these functional changes before and immediately after impact is decreased [24, 67, 68].

Bradycardia and extrasystoles in the first seconds after impact may be indicative of traumatic effects. Disturbance of cardiac rhythm in white rats, as a rule, accompanied damage to internal organs [26, 67]. However, bradycardia has been observed immediately after exposures of human subjects to $-G_x$ and $+G_x$ impact levels as low as 15 g [101]. This response was related to activity of the vagus nerve, since atropine blocks bradycardia. Test subjects also exhibited transient neurologic symptoms for brief periods after exposure to impacts in the 15 to 25 g range in the $\pm G_x$ direction.

Although physiologic stimulation may be of hormonal or neural origin, immediate onset of bradycardia in response to impact is consistent with neural stimulation. Cardio-inhibitory body reflexes can be initiated from baroreceptors in the aortic arch and carotid sinus. Visceral afferent nerves originating in nearly all tissues and organs except the skin may produce bradycardia [83]. Stretch receptors in the lung can initiate reflex cardiac slowing [16]. Stretching or distortion of lung tissue can occur during $-G_z$ impact and may be the cause of bradycardia observed in tests in this axis. Vascular fluid shifts are an unlikely source of stimulation to the cardio-inhibitory reflex areas because of the brief duration of impact. However, it is apparent that the inertial effects of $-G_z$ impact would produce a transient increase in the hydrostatic pressure sensed by the baroreceptors, which in turn respond to this pressure increase by reflex slowing of heart rate.

Evidence of damage to the respiratory system is also evident in impact studies. Injury ranges from minor functional changes in maximum ventilation of human subjects within voluntary exposure levels [35] to contusion and hemorrhage in animal subjects at near-lethal levels [6]. Restraint straps and structures may be responsible for lung damage noted in some of these experiments [6, 22, 84].

Biochemical Changes

Biochemical changes following impact have been studied in an effort to develop indices that would correlate with stress imposed by the acceleration environment, and forewarn and/or refine the definition of the injury threshold [4]. Transient hematuria and reduction in the number of circulating blood platelets has been observed after exposure of human subjects to impact [37, 101]. Urinary excretion of vanillylmandelic acid has been measured prior to and after volunteers were impacted in the $+G_x$ direction at a level of 25 g with a rate of onset of 1000 g/s and an impact velocity of 10 m/s [36]. Sham tests of each subject served as a control. Average urinary excretion of vanillylmandelic acid increased in both instances, with the greatest increase after true impact.

Impact experiments with white rats have been conducted to study the activity of aspartic aminotransferase, alanine aminotransferase, aldolase, and lactate dehydrogenase in blood serum [90]. Statistically, significant changes, rapidly appearing and prolonged in duration of activity, were noted in test groups where specific organ damage occurred. An increase in aspartic aminotransferase activity in blood serum of volunteer subjects has been found following both $+G_z$ and $+G_x$ impact tests with acceleration profiles of very short rise time and magnitudes in the range of 25 to 38 g and 22 to 40 g respectively [80]. Measurements of serum myocardial enzyme levels (creatine phosphokinase, hydroxybutyric dehydrogenase, lactic dehydrogenase, glutamic oxalacetic transaminase, and glutamic pyruvic transaminase) after $+G_x$ impact tests with human subjects at magnitudes ranging from 11.7 to 24 g have been accomplished without detecting levels outside normal ranges [43].

In this same work, a study was included of 40 accident victims, with the conclusion that serum myocardial enzymes are of no value as an index of cardiovascular injury in accident victims with mixed bodily injuries. Use of biochemical indices has proved useful in detecting the presence of general tissue damage; however, considerable research remains to provide methods to indicate specific tissue damage. Such specificity is required before a truly practical tool becomes available for clinical and impact injury research applications.

There is a general lack of controlled experimental data on physiologic and pathologic effects

of lateral ($+G_y$) impact. Prior to the emphasis placed on this particular problem by space vehicle designers, knowledge of lateral impact effects had been limited to data from accidents and from centrifuge experiments where long-duration acceleration up to 10 g was shown to be tolerable [39]. Radiographs collected during these experiments showed extensive displacement of thoracic and abdominal viscera at acceleration levels as low as 6 g.

In support of specific spaceflight requirements, rhesus monkeys were subjected to impacts up to 75 g at velocities up to 9.8 m/s, with and without contoured lateral support, and with no observation of postmortem evidence of injury [78]. Electrocardiographic evidence of transient changes in both conduction and rhythm was noted at higher accelerations and impact velocities. Comparison of radiographs taken before and after impact revealed a heart displacement in the direction of the inertial response; however, sequential radiographic observation indicated that the heart returned to a normal position within about 3 h after impact.

Response to Angular Acceleration

Angular impact acceleration may occur during the initial phase of ejection when the escape system separates from the ejection rails or during landing impact of the spacecraft [12, 98, 102]. Studies of physiologic and subjective response of volunteer subjects have been limited to the environmental ranges that have been explored with motion simulation devices. One study, conducted with acceleration durations of 0.2 to 0.22 s and braking durations of 0.25 to 0.26 s, explored acceleration levels up to 534 rad/s^2 with rotation about a "side to side" axis close to the seat-man center of gravity [100]. Limiting symptoms were manifested as hyperemia, indicating that the limiting factor for the range of acceleration amplitudes and durations explored thus far is the inertial force within the cardiovascular system acting within the head. Angular accelerations up to 1089 rad/s^2 with a duration of 0.2 s (braking deceleration was 816 rad/s^2 for 0.25 s) were well-tolerated when the rotation was about the longitudinal axis of the body.

The effects of angular velocities up to 13.1 rad/s have been studied with exposure times of several seconds [104]. These velocities were tolerated when the axis of rotation was through the center of gravity of man, i.e., through the abdomen at the level of the iliac crest. Symptoms in the head approached subjective tolerance at 8.8 to 9.4 rad/s. The development of conjunctival petechiae was found to be a reliable measure of the stress imposed on the unsupported peripheral vasculature. The curve for conjunctival petechiae, when the center of rotation was at the iliac crest, varied from 3 s at 9.4 rad/s to 2 min at 5.2 rad/s. With the center of rotation at the heart, petechiae appeared only at velocities of 2.7 to 3.1 rad/s higher for the same durations.

Cumulative Effects of Omnidirectional and Repetitive Impact

The unpredictability of the impact vector and the possibility of repetitive impacts during capsule landing in rough terrain or severe sea conditions necessitated various studies with oblique impact vectors. Although these results are by no means conclusive or exhaustive, they proved the safety of limited, anticipated impact profiles and precluded unexpected biological effects [66, 95, 103]. These studies are discussed in more detail later in the chapter.

Evidence of cumulative effects of several successive impact exposures of human subjects in the same or different directions close to voluntary limits has not been reported so far. The number of subjects and exposures are too limited, and physiologic and psychologic tests are too crude to permit valid differentiation of subtle effects of such stress from the changes with time in individuals not exposed to impact.

Experiments designed to study the pathology associated with repeated impacts have been carried out with white rats [23, 29]. This study was performed with impacts up to approximately 600 g at 1.2 to 0.8 ms durations. Accelerations of 450 to 600 g were applied at 2 to 3 min intervals in one series of experiments and in 1 and 24 h intervals in a second series. The animals were impacted 2 to 14 times. Impact velocities were

varied from 4 to 7 m/s. Cumulative lesions resulting from repeated exposures at 1-h intervals were detected as primary lesion of the lungs. Lesions developed after a comparatively small number of repeated exposures.

Another study was carried out with white rats and dogs exposed to repeated impacts at lower levels [26]. The rats were exposed to 300–350 g three times at 10-min intervals. Respiratory and heart rate changes intensified with repetition of the impact. Disturbances of cardiac rhythm (extrasystoles, atrioventricular block) became most marked. Dogs were exposed to 4 to 5 impacts at levels less than 200 g of 0.01 to 0.015-s duration at 2 to 5 d intervals without marked increase in functional disorders which had been observed in white rats [26]. The investigators concluded that the length of the impact interval reduced the degree of functional deviations and apparently caused some adaptation.

RESEARCH STUDIES OF TOLERANCE LIMITS

Numerous approaches have been used in research to determine the physiologic effects of impact and to quantify impact exposure limits. Early studies of man's reaction to impact, conducted during and immediately after World War II, were directed toward answering questions concerning the safety of ejection seat catapults [1, 17, 81, 86]. Extensive experimentation was also undertaken to study effects of aviation crash landings and the short-duration $-G_x$ deceleration encountered during ejection from a high-speed aircraft [94]. Most of this early experimentation was with human subjects—often the investigators themselves. Anthropomorphic dummies were used to evaluate adequacy of the experimental apparatus prior to tests with volunteer subjects, but the usefulness of data collected with dummy subjects was very limited. Animal tests were also performed but their value was minimal and at best, qualitative, due to the paucity of information that might be used to relate the relative impact tolerances of man and animals.

The significant work accomplished at this stage in the development of aviation medicine was, for the most part, based on subjective comments of volunteers, symptoms that were usually mild and often vague, and judgments of investigators. This approach continues to be used to define voluntary tolerance limits and evaluate the relative merits of protection systems, but refinements of methodology and more substantial scientific literature have reduced somewhat the risk associated with this approach.

Impact testing with animals has become a more meaningful approach to assess the effects of specific impact environments and to recognize and analyze specific injury patterns as the volume of data collected with each species has increased. Experiments with animals provide a basis for estimating injury types that might be expected for different acceleration directions and variations in protective equipment configurations [26, 45, 56, 59, 65, 93]. Animal tests to determine frequency of lethal injury have served to substantiate theories of the biomechanical effects of impact, that is, deformation of load-bearing tissues and effects of impact-time parameters on the attainment of injurious levels [50, 51]. While animal data originally were only of qualitative use in identifying injury patterns and mechanisms, their quantitative usefulness had to wait for the establishment and verification of dimensional scaling laws based on broad progress in biomechanics. The validity of these scaling relationships is supported by tests with various types of mechanical stimuli such as airblast, vibration, and sustained acceleration [6, 105, 106].

Despite advancements made in this aspect of impact research, data collected from animal experiments must be approached with more than an ordinary degree of caution. Basic differences in anatomical geometry on both a macro- as well as a microscopic level undermine the fundamental scaling requirements for similitude of structural geometry and material properties. Furthermore, not only may the dimensional proportions of the animal be significantly different from those of man, but also, perhaps more importantly, physiological responses may be manifestations of other dissimilarities.

The use of human cadavers or tests on their tissue or organs constitute another approach used to determine impact limits without actually

endangering living subjects. This approach has been more successful in studies of the breaking strength of bone since postmortem changes in bone are less pronounced than in soft tissue. Impact exposure limits for the $+G_z$ direction have been developed, partially based on tests conducted on cadaver vertebral segments [27, 81, 97]. A great deal of the findings available on head injury [31, 33, 53, 55, 85] has been obtained from tests with cadaver skulls.

Biomechanics Research

Contemporary biomechanics research has become progressively directed more toward establishment of impact exposure limits in terms of probabilities of injury and/or fatality instead of the oversimplified concepts used earlier of "limit of tolerance" or "zone of injury." Such relationships can only be obtained by the integration and correlation of all six basic approaches:

1. Experimentation at low-impact levels using volunteer subjects to establish basic kinematics of the living body and its relationship to kinematics of animal and cadaver bodies;
2. Discovery of areas of injury, mechanisms of injury, and severity of local impact using cadavers at high-impact levels;
3. Experimentation with animals to study the full range of physiologic and pathologic responses in various species;
4. Analysis of human accident data to verify laboratory research and clinically evaluate severity of the injury and the longer term outcome of these injuries;
5. Testing of isolated components of the human body such as vertebral segments or skulls to determine mechanical properties, i.e., breaking strength, stiffness, and others.
6. Integration of results from approaches (1) to (5) into a theoretical framework or mathematical model, which allows prediction of response dynamics and injury probability for exposure parameters not yet experimentally tested.

One major difficulty in determining useful impact exposure-limit criteria is that impact levels are not determined by the biological system alone, but are strongly influenced by, and coupled to, the body support or restraint system used in applying mechanical force to man. A definition of impact-exposure limits without definition and accurate description of this support and restraint is meaningless. The physical dimensions and mechanical properties of all contact areas, that is, seat, backrest, restraints, head support, and others, must be controlled and described with test data. With animal experiments, these "mechanical components" must also be scaled dimensionally, dynamically, and in strength to allow meaningful extrapolation to the human case.

Mathematical Models

The application of models to represent dynamic responses of the human body and support and restraint systems can be of great value in determining relative effects of specific characteristics of the human, or his mechanical protection system elements in impact environments [71, 72, 89]. Their use further enables analytic determination of the detailed effects of complex waveforms that could not be obtained using such simple parameters as peak acceleration, rise time or rate of onset—parameters which are meaningful only as descriptors of relatively simple waveforms.

Various models developed have had one or more of these purposes:

1. Understanding the basic pathologic, physiologic or anatomic dysfunctions resulting from impact;
2. Extrapolating from environments evaluated in the laboratory to operational environments not yet tested;
3. Determining optimum protection system designs for a given set of environmental parameters;
4. Using the model to evaluate and interpret tests on human surrogates, i.e., animals or anthropomorphic dummies;
5. Providing a technique to describe human tolerance to impact in a format that can be more easily understood by aerospace equipment design engineers.

General types of biodynamic models may be categorized as models that describe properties of tissue, human body subsystems such as the head and neck, total body response, or kinematic response of the whole body. Models developed to describe experimentally obtained tissue properties provide some understanding of basic physical processes by which mechanical energy is transmitted through the body tissue in various frequency ranges [20, 47]. Subsystem models of the human body such as mathematical representations of the head [10] and spinal column [9, 27] have the greatest degree of practical usefulness. Models of this type account for the statistical variability of failure modes and effects of parameters such as age of the individual [97].

The total body model is composed of several of the subsystem models and allows more complete understanding of interaction of various responses. Kinematic models depict individual segments of the body as a linkage system with individual components having the geometric shape and inertial properties of human body segments and the degree of joint mobility as well as muscle forces derived from experimentation [62]. Such models are useful in determining crewmen's motion of the body segments during specific impact conditions and in predicting interaction of body segments with the restraint system and interior surfaces of the spacecraft.

Model Response to Impact Forces

Most of the total body and subsystems models used to describe human response to impact forces are of the lumped parameter type, presenting the body or body segment as a mechanical system composed of masses, springs, and dampers. These models assume a simple stress-strain relationship. More complex models have been suggested and mathematically described; however, available data on mechanical properties of the body are not yet sufficient to justify their use in evaluation of practical operational problems. Such models can be used successfully to describe main tissue motions such as head, upper torso, or abdominal viscera motions. At higher frequencies, lumped parameter representation becomes increasingly less valid when wave phenomena (transverse shear waves as well as compression waves) become apparent. However, gross body deformations and organ motions leading to major injury patterns observed under impact accelerations occur in time periods corresponding to frequencies below several hundred Hz and are well-described by lumped parameter representation.

The model shown (Fig. 3) is an example of a total body model developed to combine the body's response characteristics in the G_z direction as measured in both vibration and impact exposures. Only the airways are represented by their fluid dynamic properties and not by lumped parameters. Spinal compression, interthoracic pressure, and chest and abdominal motions can be calculated for this model and exhibit typical resonance phenomena observed on these systems under impact or steady-state vibration. For example, the upper torso mass combined with the spinal spring has a resonance of 5.6–8.4 Hz and the abdominal mass undergoes maximum displacement, i.e., is most sensitive, in the 4–6 Hz region. For a more detailed analysis of specific injury modes, it is often preferable to use subsystem models where further refinements and nonlinearities can be investigated more easily. An example of the application of such a simple lumped parameter model to describe spinal injury under $+G_z$ impact loads will be discussed later.

Total body models usually are a complex coupling of simple second-order subsystems, each representing individual dynamics of a body segment or organ system. Although the acceleration transmitted to a specific subsystem may be modified by the dynamics of intervening and surrounding subsystems, the tolerable (noninjury producing) acceleration level is determined primarily by the individual response of each subsystem. Thus, the dynamic response characteristics of the system, i.e., natural frequency, damping properties, and the like, determine sensitivity to impact.

A complete discussion of how differing impact environments produce different maximum strain or peak force level in a second-order system is beyond the scope of this chapter. Detailed discussions of these effects are available in the

technical literature [51, 71, 97]. However, certain basic principles of dynamic systems should be understood. First, the maximum strain or peak force in a dynamic system is related to velocity change associated with impact acceleration-time history when the acceleration time duration is less than the natural period of the dynamic system. The force that will cause equal strain increases as the acceleration pulse duration decreases. Second, for a given pulse shape, the maximum strain or force level in the dynamic system is primarily related to acceleration magnitude when acceleration duration is greater than the natural period of the system. If the

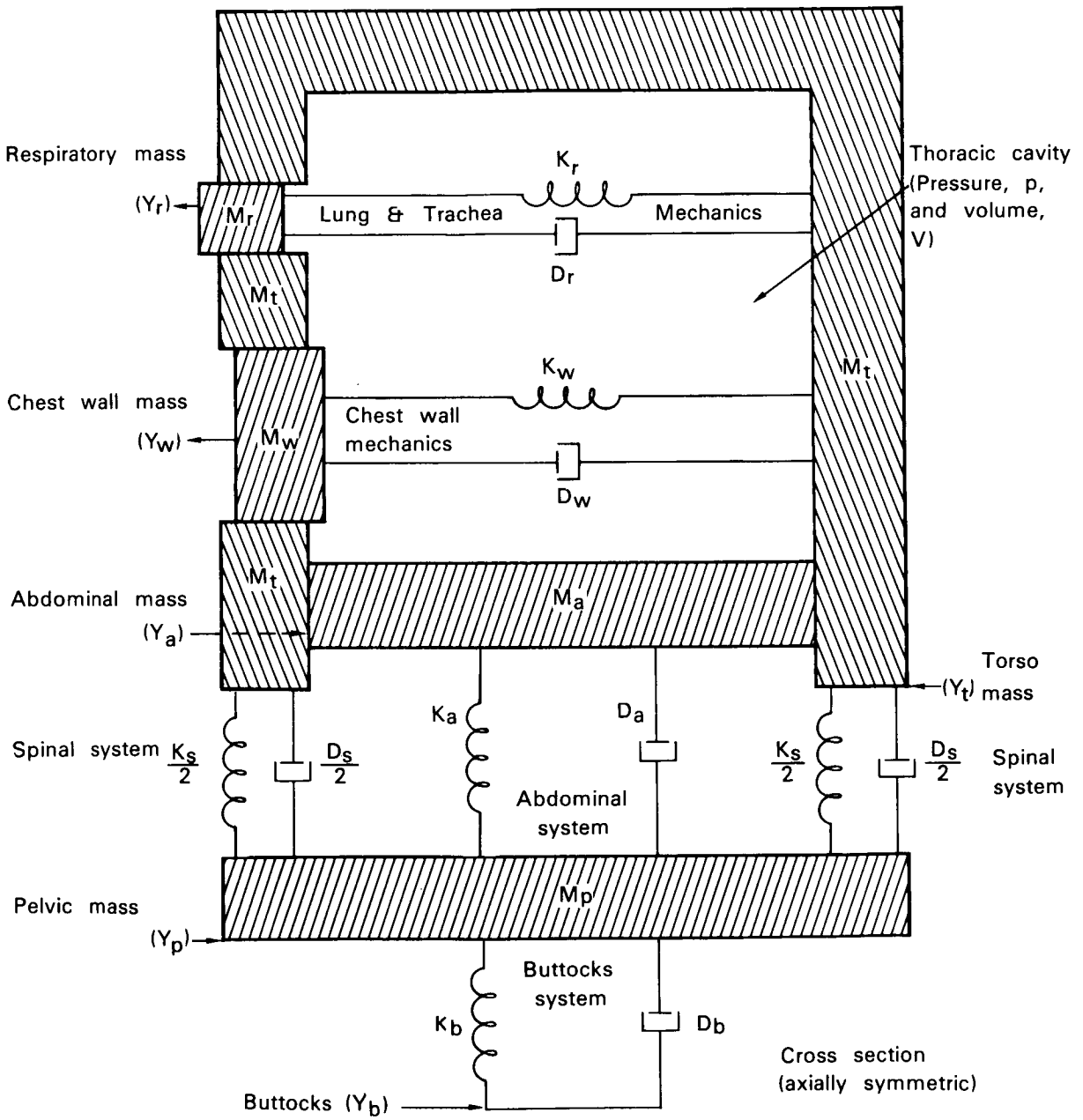


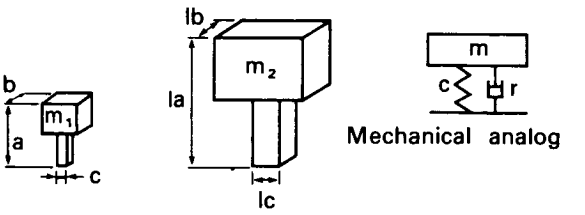
FIGURE 3.—Multidegree of freedom model to depict whole body response to impact [21].

impact pulse duration is comparable with the natural period of the system, the peak strain or force is a complex function of velocity change and acceleration level. (The natural period of the dynamic system depends upon its natural frequency and damping coefficient.) Therefore, in later sections of this chapter dealing with these parameters, velocity change and g level are used as descriptors of human exposure limits.

Subhuman Primates Response to Impact Forces

Interpretation and application of the relatively large amount of available data on the effects of impact on subhuman primates and other mammals is vitally dependent upon the use of model scaling techniques. The basic assumption of this approach is that an impact environment will lead to similar injury mechanisms in animal and man when dynamic similarity or scaling laws are applied. This assumption must be continually verified with efforts to use this approach, in light of the geometric dissimilarities between species. Methods commonly employed in such verification include evaluation of the similarity of the mechanical properties of tissue; steady-state vibration response analysis of various species of different size; kinematic response to impact; and evaluation of injury mechanisms observed in clinical investigations of humans involved in accidents where the impact environment can be reasonably estimated.

The anatomic and physiologic differences between various species and assumption of similarity of injury mechanisms may present sizable obstacles; however, valuable first approximation results can be obtained by using scaling laws. By applying the scaling laws (in Fig. 4), approximate resonant frequencies may be obtained for chest, spinal, and abdominal systems for various animal species (shown in Fig. 5). Smaller species generally have higher natural frequencies for the same organ, which involve two important consequences; in a somewhat oversimplified statement, these are: (1) equivalent injury patterns in smaller animals are produced by correspondingly shorter duration impact patterns, which leads to the requirement for "scaling" the impact pattern for experiments with small animals in order to make results interpretable in terms of human injury; and (2) smaller animals in general have lower impact sensitivity, i.e., they can stand higher G-loads.



	Tissue	Airspace
$\frac{m_2}{m_1}$	l^3	l^3
$\frac{c_2}{c_1}$	l	l
$\frac{r_2}{r_1}$	l	l^2
w_0	$\frac{1}{l}$	$\frac{1}{l}$
Damping $\frac{r}{w_0 m}$	$\frac{1}{l}$	Constant

FIGURE 4.—Scaling laws for geometrically similar structures such as mammals of different size [21].

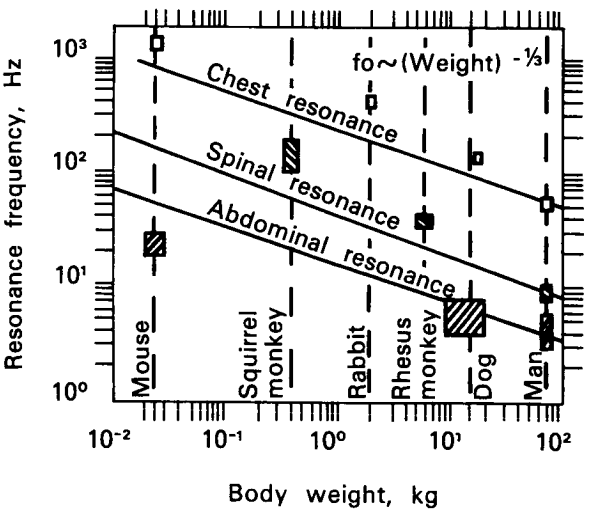


FIGURE 5.—Approximate resonance frequencies of total body response models as a function of body size (weight) [21].

Impact Simulation Techniques

Mechanical facilities in a wide variety have been used to simulate the impact environments anticipated in normal and emergency spaceflight operations. To assure broad usage of test data, their mathematical interpretation, and easy application to biodynamic models, most work has not been conducted with the complex acceleration waveforms encountered in actual operational situations, but with simple approximations to these patterns such as rectangular, triangular, and half sine pulses. The simplest of the facilities are the vertical deceleration towers—devices which use gravity to assure the reproducibility of the impact velocity. The impact-time history may be controlled by using hydraulic decelerators [11, 103], crushable materials such as paper or aluminum honeycomb [37, 46], or energy storage devices such as elastomeric materials or liquid springs.

Ejection towers, which have been used since immediately after World War II to study man's response to $+G_z$ acceleration, evaluate personnel protective equipment, and provide crew training [1, 17], have incorporated both pyrotechnic and pneumatic devices to accelerate ejection seats and subjects. Rocket-powered sleds, propelled along horizontal tracks into water brakes, have been used to study combined effects of short-duration decleration and windblast encountered during emergency escape from high-speed aircraft [1, 94]. More precise studies have used a pneumatically propelled sled and water brake decelerator, designed for conducting human tests [11]. This facility is shown in Figure 6.

Other impact simulators include simple pendulums and pneumatically powered strikers. Pendulum impact devices have been used to study impact protection systems [94], head impact tolerance, and to evaluate protective headgear. Special small-scale pneumatic strikers have been developed to study head and thoracic trauma [70].

Impact simulators must be designed to provide precise control of the impact environment parameters, if human subjects are to be used at impact levels approaching tolerance. Reproducibility of the test environment is especially critical in experimentation where impact stress is increased in small increments until voluntary tolerance is reached. Furthermore, the test apparatus used with the simulator must be given extraordinary care in design and in understanding its contribution to test results. Where prototype hardware, such as an astronaut ejection seat, is used, it must be recognized that the design of the structure of the seat may include only a small margin of safety, for example, a factor of 1.25, since the impact environment under study would be encountered only under emergency conditions. This margin of safety, while suitable for a low occurrence probability such as emergency escape, is normally not considered adequate for experimentation with volunteer subjects.

Rigidity of the structure, or lack of it, is important not only in considering the safety of the apparatus, but also in the fidelity with which it transmits the simulator impact to the subject. The acceleration transmission characteristics of the apparatus and component articles such as seat cushions and padding are, unfortunately, often ignored. Under these conditions, it is usually difficult, if not impossible, to draw any general conclusions about the work or to extrapolate to other equipment configurations. Where determination of human tolerance is the primary objective of the experimentation, it is often simpler to assure that the structure is rigid and to eliminate elastic padding. Furthermore, the rigid structure lends itself to repetitious use common for impact testing.

Beyond the more straightforward considerations of experimental procedures and apparatus design are the fundamental ethical questions surrounding impact experimentation. Perhaps the most basic question: "Is the information value resulting from the test commensurate with the risk to the subject?", should be answered not only in the initial planning stages of the research program but also immediately before initiation of testing when the scope and adequacy of data to be collected are more completely defined. In any case, investigators are ethically bound to minimize risk to the subject. Actions which can achieve this end include thorough

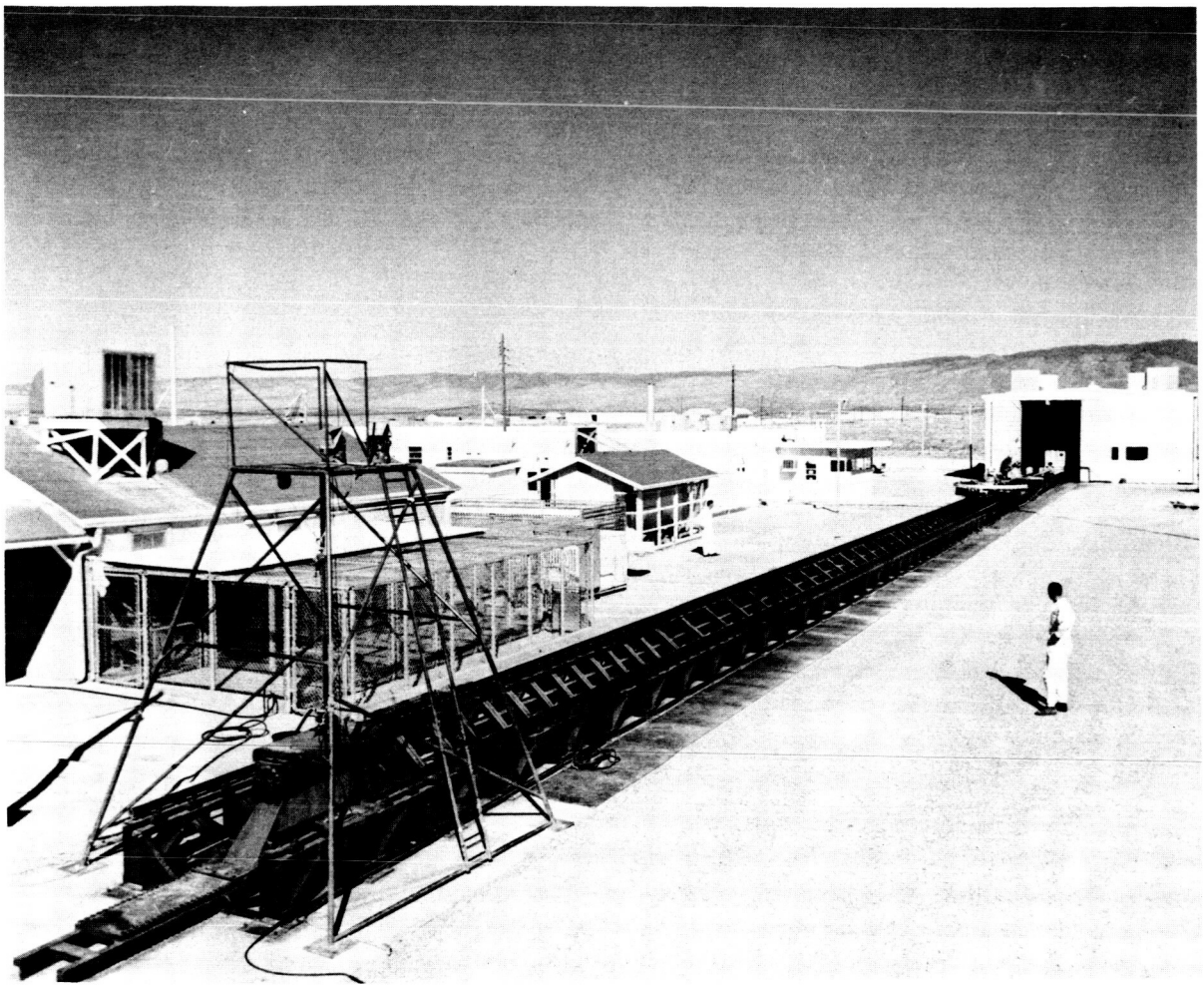


FIGURE 6.—The daisy decelerator.

physical examinations prior to, and after testing, and careful medical monitoring throughout the experimentation and posttest period as well as meticulous attention to operation of impact simulation equipment and emergency procedures. Posttest examination and followup of subjects depends on specific test goals, subject symptoms reported, and the medical investigator's report.

HUMAN IMPACT TOLERANCE AS RELATED TO SPACE MISSIONS

During early work on manned spacecraft designs, there was recognition of the necessity to acquire more complete data on human response to impact. Available literature reflected that the majority of impact research had been directed toward solution of aviation problems. First, acceleration exposure limits for the z-axis had been developed as design criteria for ejection seat catapults, and thus were defined in terms of the acceleration waveforms that are normally obtained from such ballistic devices. Second, x-axis limits were similarly defined for pulse shapes that were anticipated during the deceleration of ejection seats immediately after ejection into high-velocity windstreams. Third, practically no data were available to assess effects of impact vectors acting in the y-axis. Furthermore, information available pertained only to the

cardinal axes and thus, the effects of impact vectors acting in directions other than these axes could not be evaluated.

The work of Eiband summarized data available at that time within the US [17]. These data are summarized graphically in Figures 7 and 8. While these data have been of inestimable value in developing design criteria for manned spacecraft, they were inadequate for evaluation of specific impact problems associated with both normal and emergency astronautic operations. It was mentioned previously that providing escape from the launch pad with an ejection seat requires use of a high-magnitude, short rise time acceleration pulse.

It is also most important that landing impact environments anticipated during the recovery phase of space missions presented a set of potentially severe conditions, characterized by high-magnitude, short rise time impact pulses of varying direction and irregular waveform. The impact exposure environments, unfortunately, are hard to predict as long as the prototype space system is not available for test and always subject to large statistical fluctuations depending on details of landing conditions. Tolerance limits presented in Figures 7 and 8 are only available in terms of idealized trapezoidal waveforms. Deduction of a plateau level and time duration from a complex acceleration-time history encountered in actual practice is not an easy task, and in some instances impossible.

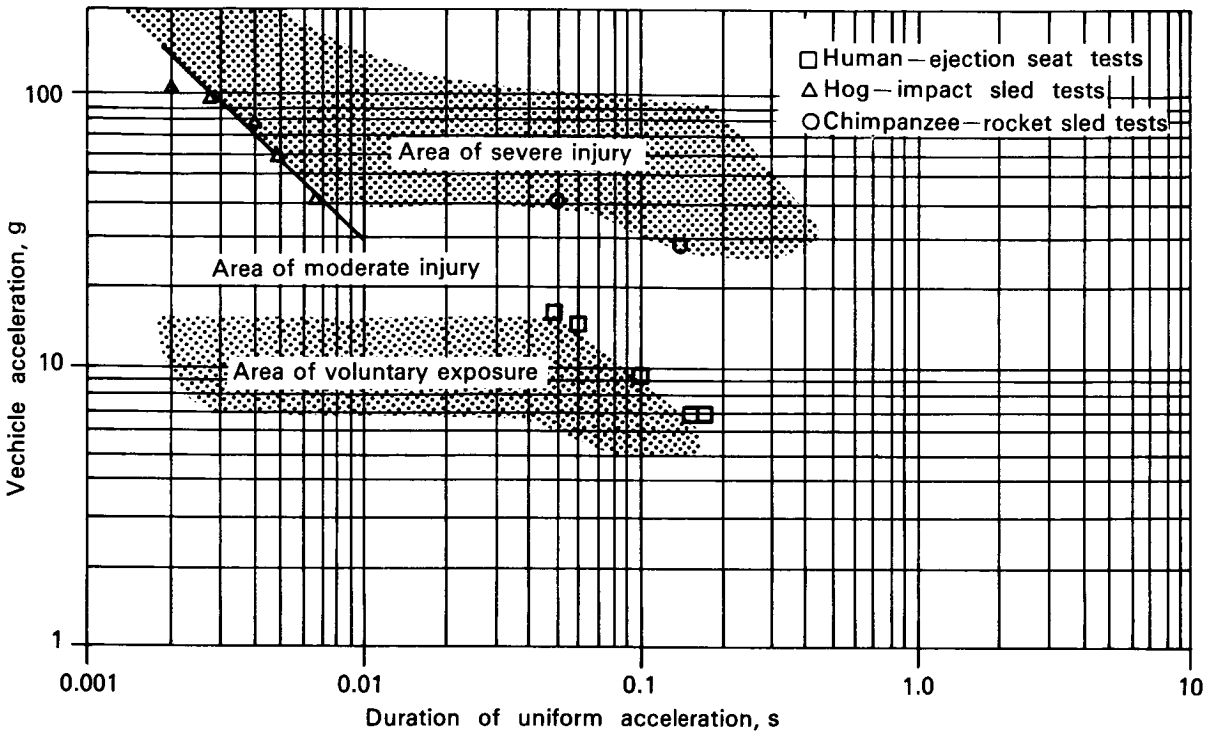
$+G_z$ Impact Exposure Limits

Evaluation of the Eiband summaries shows that there is a considerable unknown region between the areas of voluntary human tolerance and injury. In the $+G_z$ direction (Fig. 7a) the unknown area shown covers over 20 g in the ordinate and does not show human exposures for time durations less than 0.04 s. It is unfortunate that this unknown region includes impact environments of greatest interest in space operations. It is clear that boundaries are not well-defined and a few more data points might change the shape of the curves. Although plotted data are too limited in numbers of tests and control of variables to provide a basis for accurate interpretation [69], the general form of the curve shown in Figure 7a merits some comment to provide insight into the general form of the tolerance curve in the short duration region. It should be noted that for impact plateau durations up to 0.007 s, data points dividing areas of severe and moderate injury decrease in nearly linear fashion on the log-log scale as time duration increases. The relationship of these data points is as it should be, if viewed in terms of the dynamic response of a mechnical system. Use of a mechanical analog seems appropriate here, since the injury mechanism that is operationally important is mechanical in nature, that is, compression fracture within the vertebral column.

The simplest analog developed for the study of impact applied parallel to the vertebral column ($+G_z$) is a mechanical model composed of a mass, a spring, and a viscous damper [97]. The mechanical elements are lumped parameter elements, e.g., all the human body mass that acts upon the vertebrae to cause deformation is represented by the mass element. The model, shown diagrammatically in Figure 9, is used to predict maximum deflection and associated force within the vertebral column for any given impact environment. Compression fracture occurs when the force in the spring exceeds its breaking strength. Properties of model elements have been derived from existing data. Spring stiffness and breaking strength have been determined from cadaver vertebral segments, and damping ratio calculated from measurements of mechanical impedance during vibration tests [15, 97]. Response of the model can be determined for any given acceleration-time history by solution of a second order, differential equation with terms representing the positions of mechanical elements in regard to time.

Injury Prediction

The mechanical model also provides a basis for a probabilistic approach to injury prediction. Since the model reduces the effect of the impact environment to a single parameter, that is, peak deflection or force in the spring element, a correlation can be determined between this parameter and injury. For example, the breaking strength of vertebrae is variable but it can be statistically described in terms of failure prob-



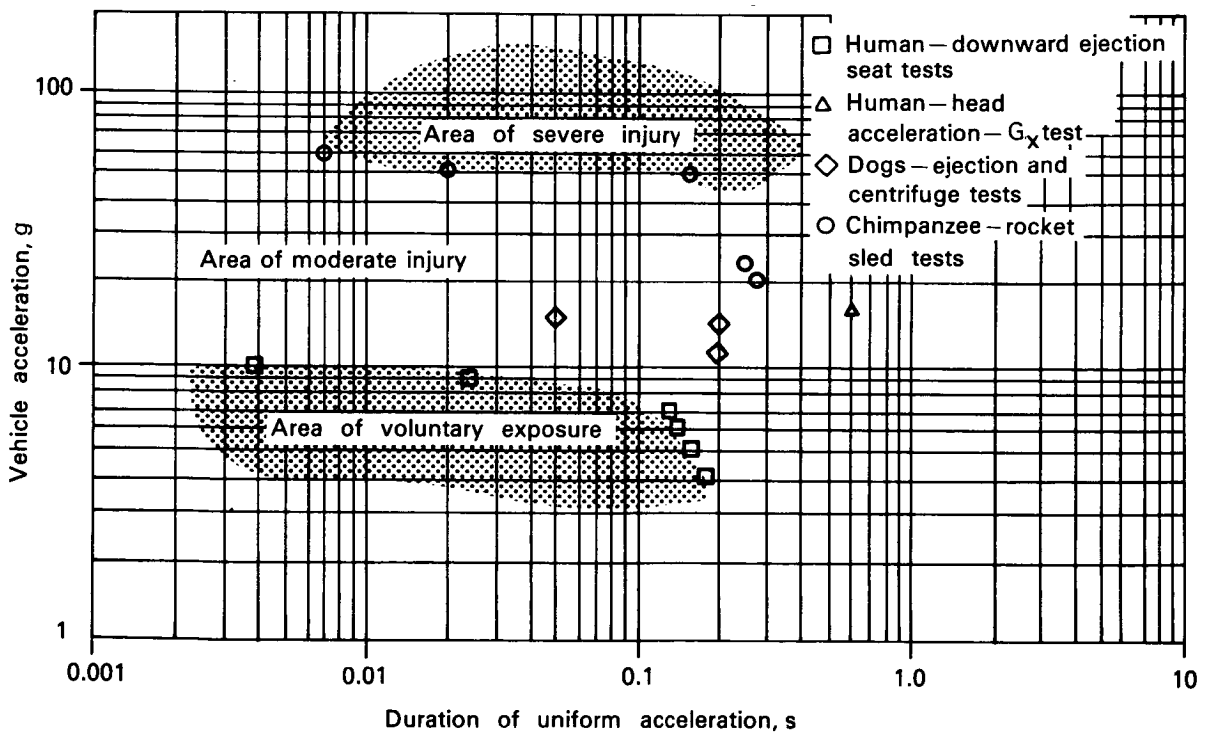
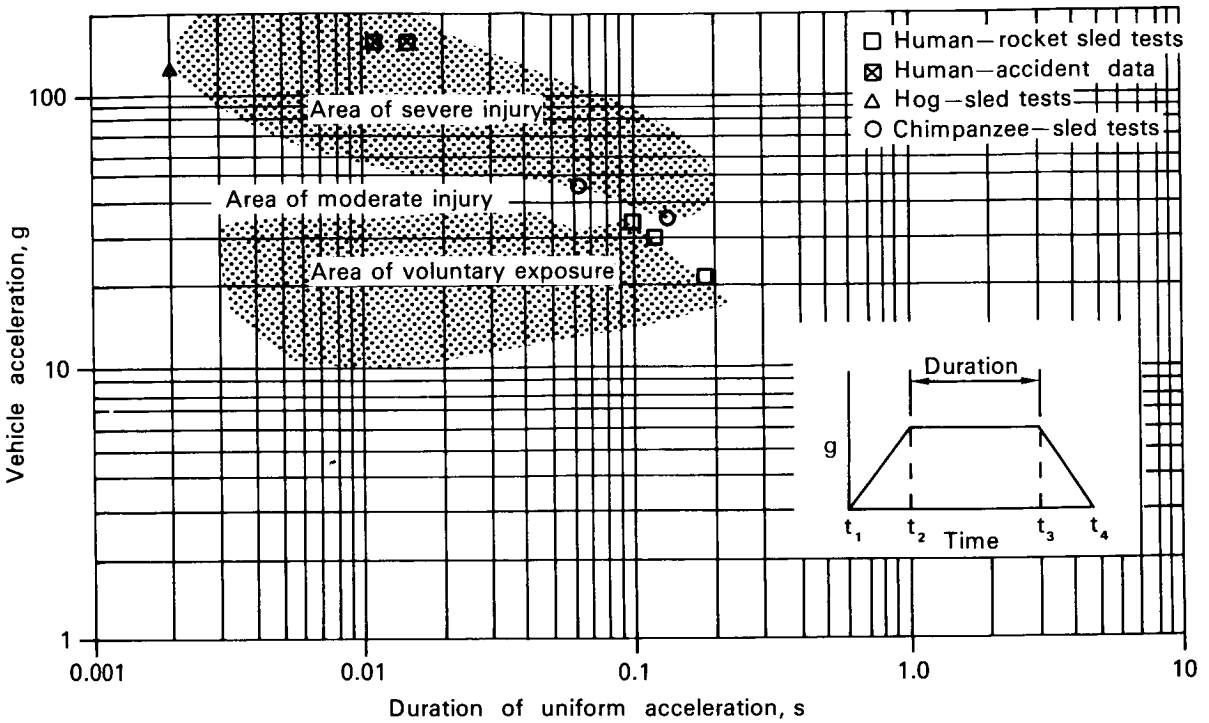


FIGURE 7, a and b.—These two graphs show durations and magnitudes of abrupt decelerations in the G_z direction which have been endured by various animals and man, showing areas of voluntary endurance without injury, moderate injury, and severe injury marked by shading. Graph "a" shows data of $+G_z$ acceleration (headward) and "b" shows data for $-G_z$ acceleration (tailward). (After [17])



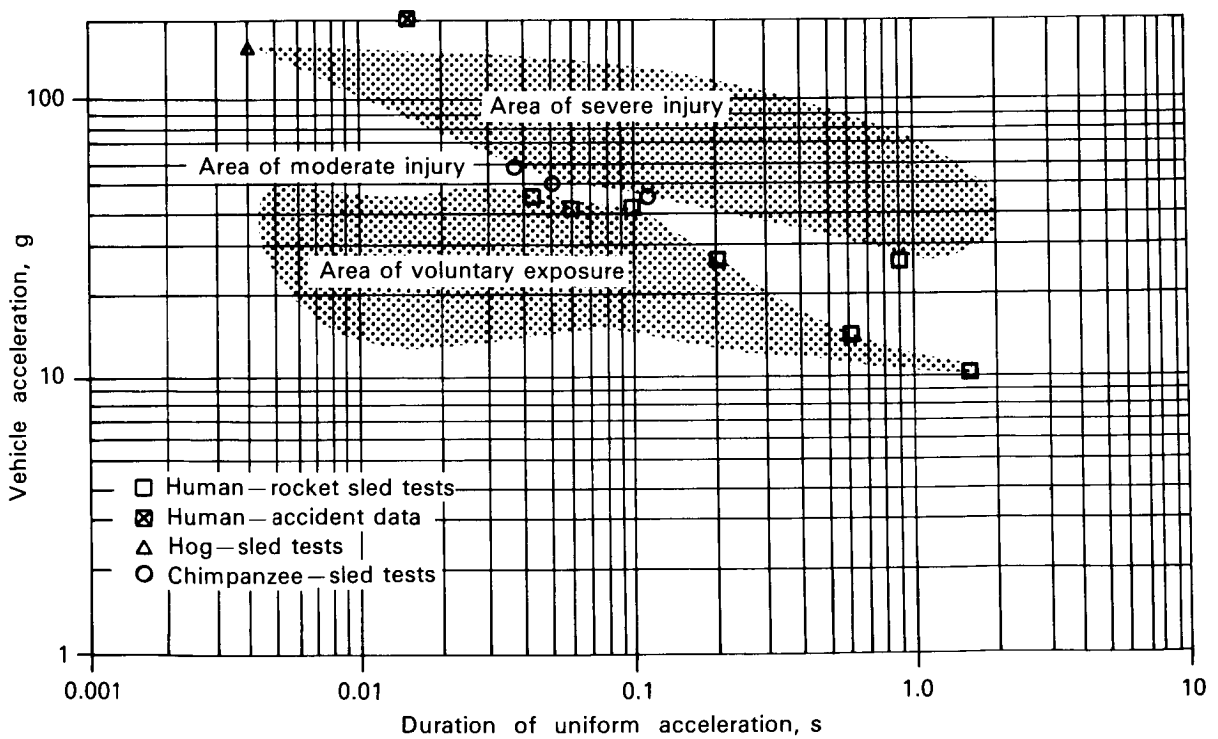


FIGURE 8, a and b.—Two graphs show durations and magnitudes of abrupt transverse decelerations endured by various animals and man, showing areas of voluntary endurance without injury, moderate injury, and severe injury. Graph "a" summarizes ($+G_x$) data (chest to back acceleration) and "b" shows ($-G_x$) data (back to chest acceleration).

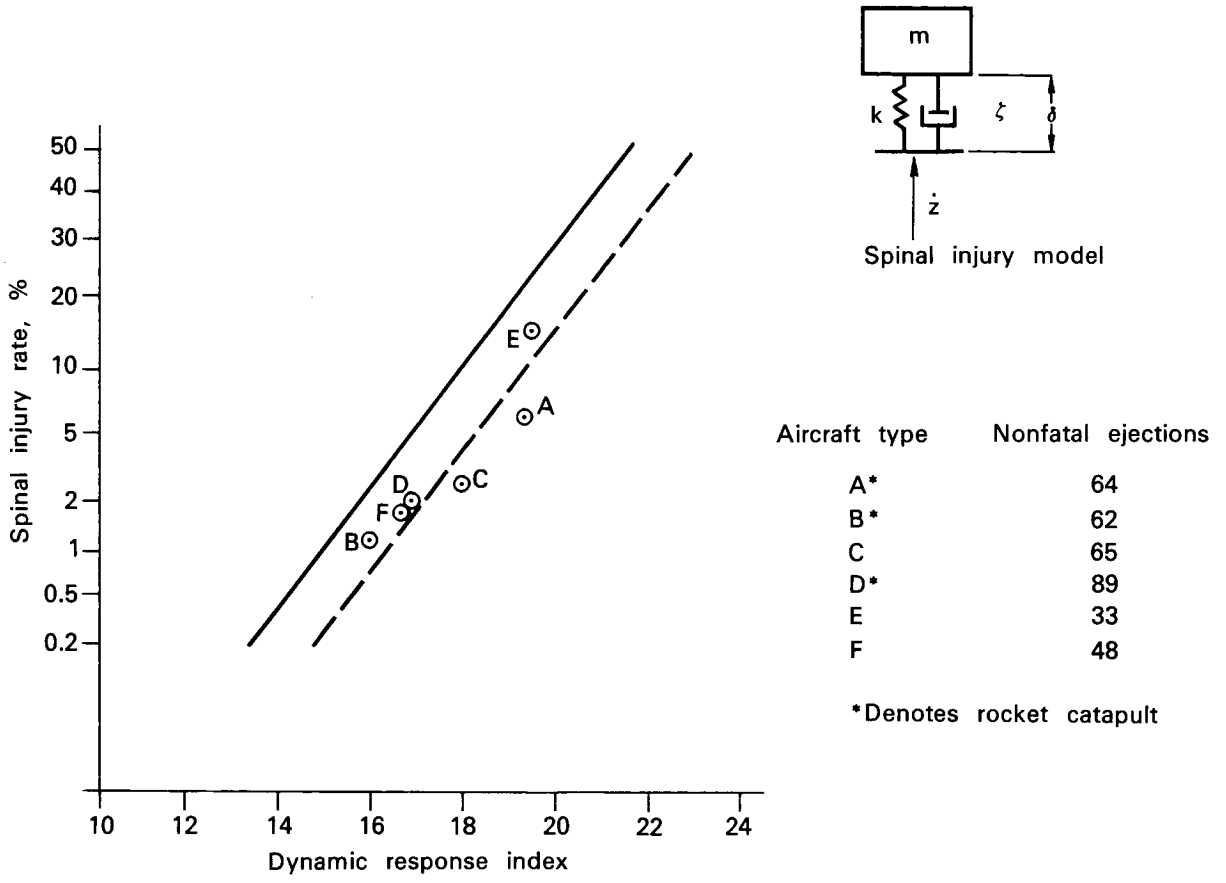


FIGURE 9.—Probability of spinal injury estimated from laboratory data compared to operational experience. (After [8])

ability [97]. This same approach provides estimates of the relationship between age and breaking strength [73, 97].

An analytical effort was made to determine the degree of correlation between the spinal injury model and injuries experienced in operational aircraft ejection seats [8]. The relationship between operational acceleration environments and actual spinal injury rates of the ejection systems included in the study are shown in Figure 9. The response of the model is expressed in terms of dynamic response index (DRI) values. The initial estimate of injury probability as determined from cadaver data is compared to operational data. The slope of the line drawn through operational data points was established on the variance of vertebral strength used to establish the initial estimate. The spinal injury model and this injury probability estimate have been used to assess risk of spinal injury associated with the Project Apollo mission impact environments.

Vertebral and Intervertebral Strengths

The vertebral failure process has best been described by a mechanical deformation and effect sequence, shown in Table 1.

Extensive studies of vertebral and intervertebral disk strength have been conducted to determine more precise estimates of $+G_z$ impact tolerance [27]. This work significantly increases the number of data points, since a total of 530 vertebrae was studied, which included tests of cervical vertebral segments. Only a few data points were available previously to provide

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an estimate of the breaking strength of the cervical spine. The mean ultimate strength of vertebral segments tested in this study are in Table 2. The values indicate the same general change of breaking strength as a function of position of the vertebral segment, as do similar collections [73], but the breaking strength is approximately 18% higher. Data were obtained from vertebral specimens in men ranging in age from 19 to 40. Less than 30 h elapsed after death before the start of the experiment. Data shown were obtained at a deformation rate of 10 mm/min. From 6 to 16 observations were used to compute arithmetic means.

TABLE 1.—*Mechanical Failure Sequence of Vertebral Body Under Axial* ($+G_z$) *Compression* [27]

Deformation, %	Effect
6–10	Within elastic range of deformation No macroscopic structural changes
12–13	First macroscopic irreversible changes Compression of limbic zone
17–18	Cracks and compression in area of wrist of vertebral body
25–26	Fractures within vertebral bodies without displacement of hips
36–37	Fractures with dislocation

Average mechanical characteristics of intervertebral disks of cervical, thoracic, and lumbar sections of the vertebral column are in Table 3. The ultimate strength was identified by rupture of the disk fibrous ring and extrusion of a jellylike substance.

Physical Inactivity/Immobilization and Weightlessness

In connection with longer space missions, potential effects on impact tolerance to prolonged immobilization, physical inactivity, and weightlessness have been of considerable interest and speculation [30, 82]. Cardiovascular and metabolic effects of simulated and actual weightlessness are treated in a separate chapter; it will only be mentioned here that cardiovascular changes observed must have some effect on the cardiovascular impact responses described. Quantitative data on this subject are not available and these changes are not usually considered to limit human tolerance. However, decrease in bone strength from osteoporosis of disuse is established; bone loss has been measured on astronauts after space missions and in simulated weightlessness studies on man and animals [61]. Although bone loss, per se, cannot yet be related directly to bone strength, there is good reason to assume a noticeable reduction in bone strength after prolonged space missions.

In rhesus monkeys immobilized for 240 d by plaster of Paris casts, a reduction of 25% in overall spinal impact tolerance was observed, the main decrease in strength having already occurred after 60 d immobilization, which is shown in Figure 10 [46]. These data cannot yet be applied quantitatively to an estimate of strength reduction in human subjects. However, they obviously call for further studies and conservative application of all bone strength/bone impact limits data obtained on "normal" human subjects adapted to the Earth's gravitational field.

Tolerance to $+G_z$ impact applied to the standing subject has been studied to determine the effects of explosions beneath a vehicle floor [41]. With impact on the sole of the foot with leg extended, fracture of the distal tibia in the human leg resulted at a load of 680 kg applied in axial compression between knee and foot [42]. Limiting velocity change for impact transmitted to a stiff-legged subject is 3 m/s; the resulting impact exposure limit curve is shown in Figure 11. A few empirical studies on cadaver legs are plotted. Such exposure criteria are of value in the design of lunar or planetary landing vehicles where the crew may be standing upright during landing.

After the initial compressive phase of impact motion response of the floor, the unrestrained man will be thrown and propelled off the floor with some velocity that will not cause injury; however, it will have bearing on his velocity at the termination of his motion when injury can occur. The kickoff velocities of men in the standing and seated positions have been measured for various impact pulses [41]. Ratio of peak deck velocity,

TABLE 2.—*Ultimate Strength of Vertebrae Compressed Vertically* [27]

Vertebra segment	Strength, kg	Vertebra segment	Strength, kg	Vertebra segment	Strength, kg
C1	800	T2	436	T10	860
C2	510	T3	467	T11	917
C3	404	T4	522	T12	1054
C4	408	T5	551	L1	1059
C5	453	T6	619	L2	1175
C6	563	T7	681	L3	1269
C7	464	T8	824	L4	1296
T1	475	T9	840	L5	1286

TABLE 3.—*Mechanical Characteristics of Intervertebral Disks Compressed Vertically* [27]

Vertebra section	Ultimate strength, kg	Elastic deformation, mm
Cervical	486	1.2
Thoracic	1270	1.6
Lumbar	1502	2.1

V_d, to kickoff velocity, V_k, was plotted as a function of the ratio of rise time to peak velocity (t_p) to natural period of man (T) (Fig. 12). The curves follow the form:

$$\frac{V_k}{V_d}=2.7\left(\frac{t_p}{T}\right)^{0.44}$$

where T is 0.1 s for the standing man and 0.167 s for the seated man.

Transverse ($\pm G_x$) Impact Exposure Limits

Impact effect in the $-G_x$ direction is critically dependent upon the type of restraint and body posture at impact time. Volunteers have been exposed to impact levels to approximately 45 g for 0.09 s with an onset rate of 413 g's [94]. Subjects were restrained by 7.5 cm-wide shoulder straps, lap belt, and thigh straps, and the subjects' head and neck were preflexed prior to impact. Onset rate or rise time was instrumental in production of perceptible subjective differences and cardiovascular shock symptoms. Under operational conditions where only 5 cm-wide shoulder straps and lap belt are used, and the crewman wears a helmet weighing from 1.5 to 2 kg, moderate injury may be expected at as low as 30 g. In an open ejection seat, even higher acceleration levels can be tolerated because of the counteracting effects of aerodynamic forces. If the crewman is protected only by a lap belt, impact tolerance is reduced further. Volunteers have tolerated short $-G_x$ impacts up to 32 g where the impact velocity was 4.69 m/s and the acceleration duration was 0.001 s with an onset rate of 1600 g/s with no significant injury [94]. In other experiments, the volunteer was restrained only by a lap belt; impact velocities ranging from 5.8 to 8.8 m/s with accelerations from 11.4 to 20.0 g produced more pronounced subjective complaints and minor trauma [94].

In transverse $+G_x$ impact direction, human tolerance is potentially higher than in any other axis, if the crewman is restrained by full body

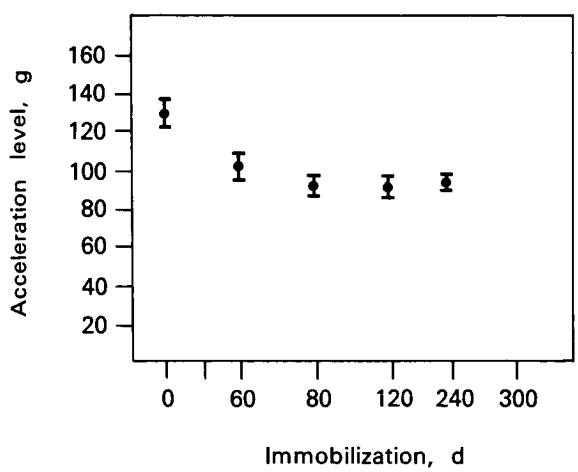


FIGURE 10.—Spinal impact tolerance of normal and osteoporotic primates. (After [46])

support. Impact levels up to 35 g for 0.16 s with an onset rate of 1115 g/s were tolerated by a volunteer subject with only relatively mild symptoms [94]. Shock symptoms—pallor, vertigo, no readable blood pressure, and loss of consciousness—were the results in a volunteer test at 40.4 g from 0.040 s duration with a velocity change of 14.8 m/s and a rise time of 0.083 s, resulting in an onset rate of 2140 g/s [94]. Human subjects exposed to $+G_x$ impact in the range of 35 to 40 g for 0.03 s with onset rates of 4000–5000 g/s complained of pelvic pain and changes in cardiovascular system activity, that is, bradycardia and decreases in systolic and diastolic blood pressures were recorded [4]. The impact exposure limit for the $+G_x$ direction, based on available data, is estimated at 35 g for acceleration durations up to 0.1 s to prevent injury [17]. Higher accelerations have been estimated as tolerable, if moderate injury is acceptable.

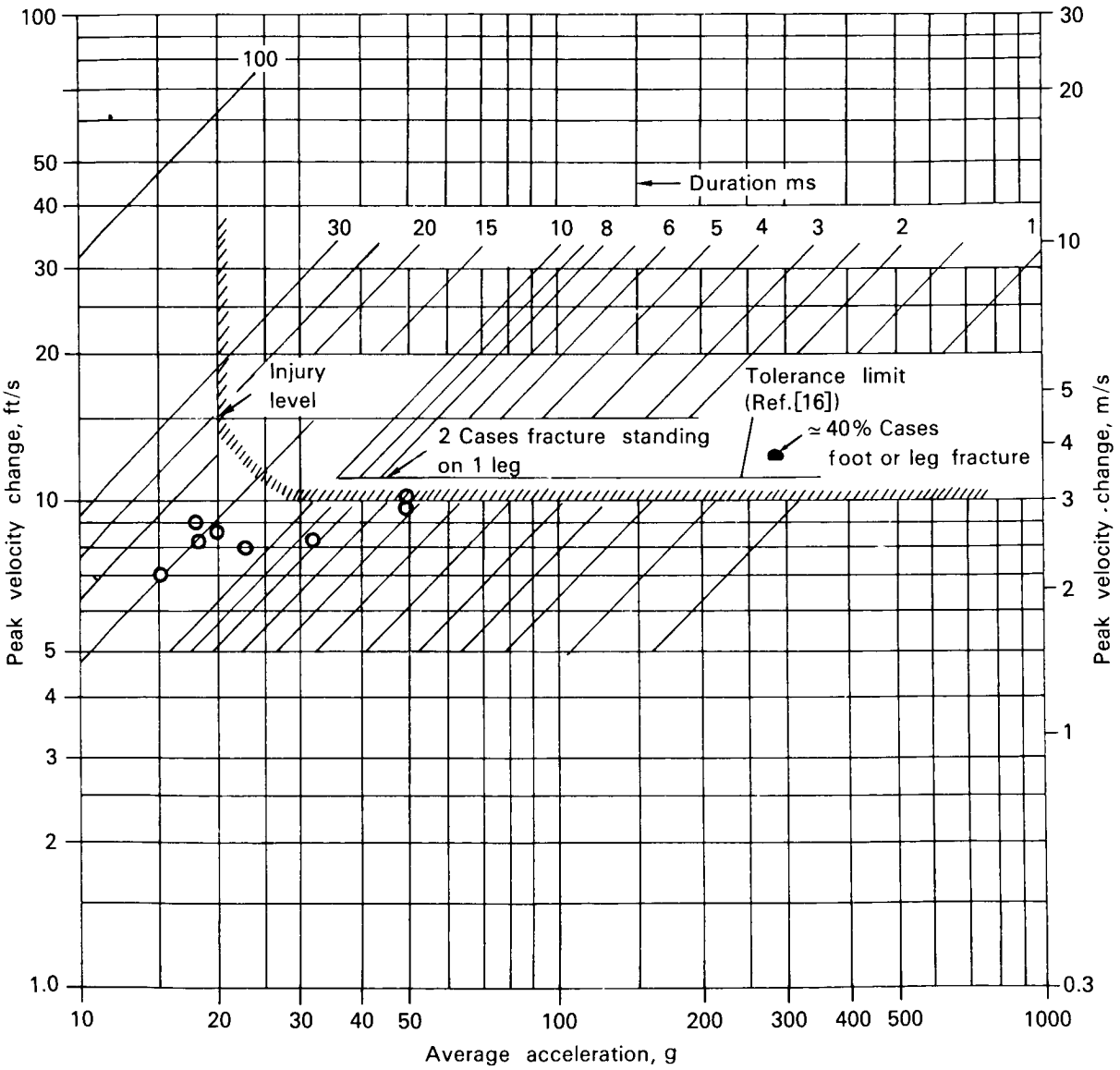


FIGURE 11.—Tolerance of stiff-legged standing men to shock motion of short duration. (After [41])

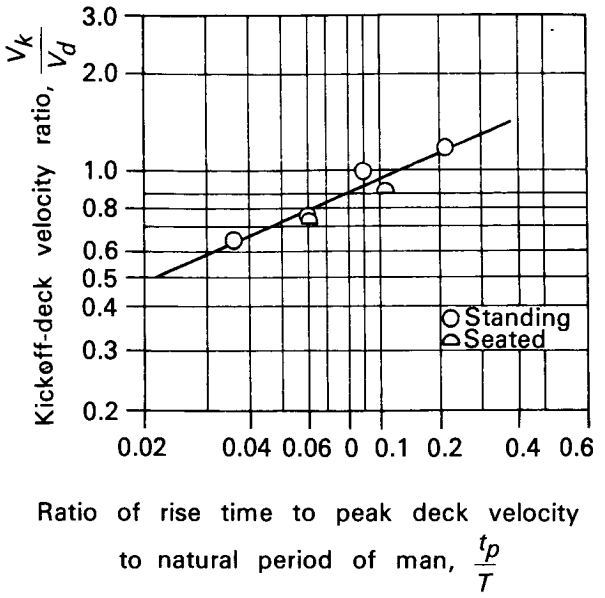


FIGURE 12.—Ratio of kickoff to peak deck velocity as a function of ratio of rise time to peak deck velocity to natural period of man. (After [41])

Lateral (+ G_y) Impact Tolerance

Human tolerance to lateral (+G_y) impact environments is not well-defined. A rather narrow range of acceleration pulse durations has been explored in tests. Volunteer subjects supported by a fully contoured couch were exposed to impacts up to 22 g with an onset rate of 1350 g/s where impact velocity was 5.9 m/s (19.3 ft/s) [13]. In another series of tests, volunteers were supported laterally by flat plates on which their shoulders would bear during impact [7]. The acceleration-time patterns used are discussed in detail in the section dealing with off-axis tolerance.

Tests with volunteers were conducted with more conventional restraints and seats, but the acceleration levels found tolerable were more moderate. A lap belt, shoulder harness, and crotch strap configuration were tested with human subjects up to 17.7 g without irreversible injury [76]. Tests have also been run with volunteers restrained only by a lap belt [107]; these tests were terminated when an acceleration level of 9 g was reached due to prolonged pain symptoms in the neck musculature.

Off-Axis Impact Tolerance

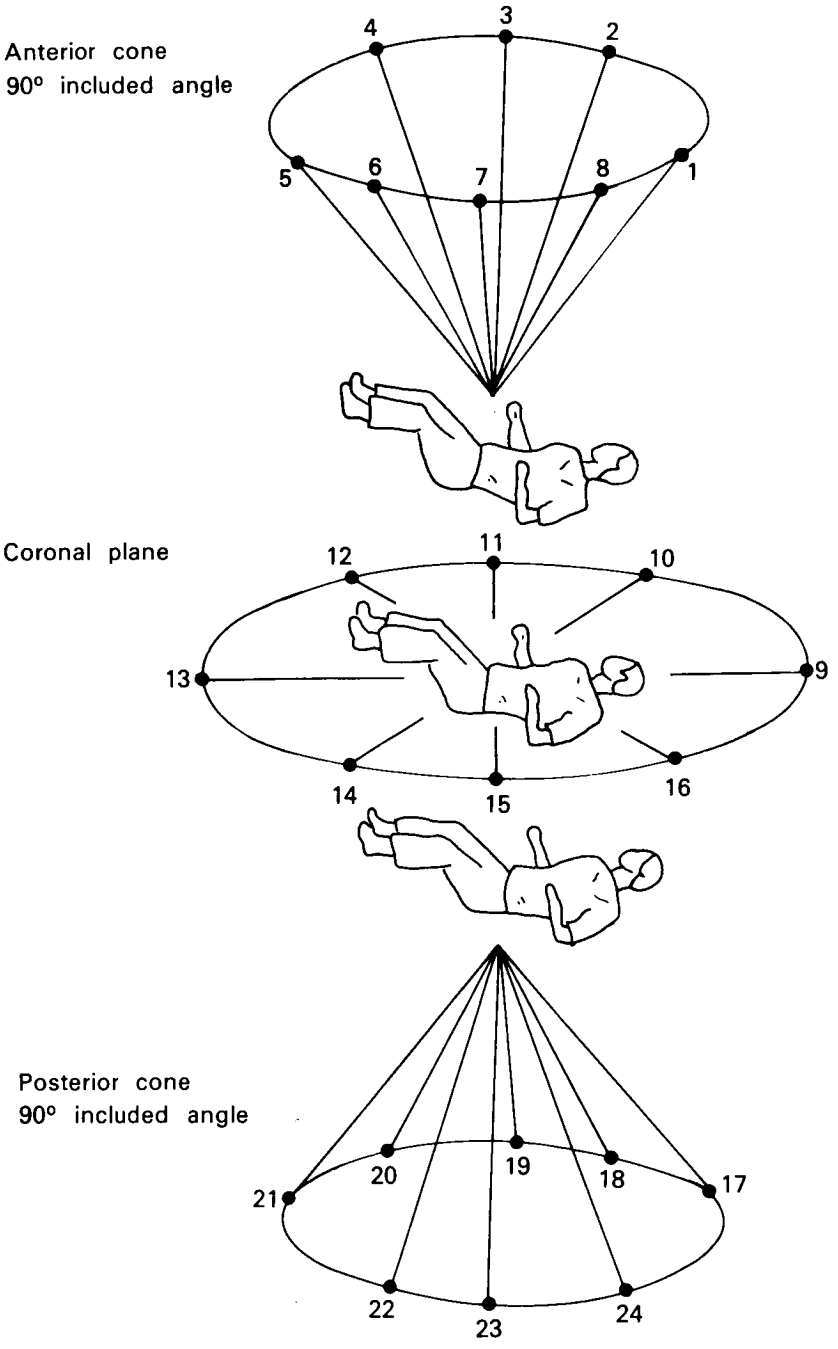
Impact exposure limits research has been concentrated on the cardinal axes, therefore, limits have not been developed for impact environments in other axes. Available data have been collected to evaluate acceptability of a narrow range of impact environments using body-support and restraint systems proposed or developed for specific aerospace systems. The most extensive work of this type was used to study impact effects resulting from descent velocities and crew module attitudes anticipated for Project Apollo landings. The scope of this work ranged from exploratory studies of the efficacy of methods to provide maximum body support and restraint to the evaluation of prototype spacecraft equipment. The effort was subdivided into several impact test programs conducted at different research facilities. The positions of the impact vector studied are described in Figure 13.

The initial series of impact tests was conducted on a vertical deceleration tower [7, 103]. In 32 tests to evaluate $\pm G_y$ impact vectors, there were no adverse subjective reactions to acceleration magnitudes to 22 g with velocity changes to 5.88 m/s. Subjects were restrained by lap belt, torso harness, and leg restraints and supported by a contoured pad filled with microspheres. This series was expanded to explore seven acceleration vector directions (positions 11, 15, 19, 20, 21, 22, and 23) and six acceleration-time histories; 20 volunteers were exposed to the impact profiles shown in Figure 14a. Peak accelerations ranged from 13.4 to 26.6 g with onsets from 426 to 1770 g/s; the power spectral density of each of these impact patterns is shown in "b" of Figure 14. Test subjects were restrained by shoulder straps and cross-chest straps converging at the sternum and a lap belt with crotch straps. The test seat included flat metal plates to support head, torso, and legs. No injuries were produced in this study, although some transient changes (abrupt rhythm changes and premature ventricular contractions) appeared in ECGs.

Sixty-one impact tests were conducted on a horizontal acceleration track with volunteer subjects to study the effects of $-G_z$ impact [40]. Subjects were restrained to a rigid couch by shoulder straps, cross-chest straps, lap belt, crotch straps,

and leg restraints. Impact magnitudes of 18.5 g were recorded on the accelerator sled with a velocity change of 5.94 m/s; onset rates ranged from 208 to 8140 g/s. Electrocardiograms of all subjects indicated transient sinus bradycardia for 2 s after impact. Bradycardia was observed in one subject for 30 min following impact.

Another series of 146 tests was conducted on a horizontal decelerator to supplement the above studies and to evaluate impact vector positions 1



Sled positions

Position no.	Roll	Pitch	Yaw
01	0	315	0
2	0	335	330
3	0	005	320
4	0	035	330
5	0	045	0
6	0	035	030
7	0	005	040
8	0	335	030
9	0	085	180
10	0	085	220
11	0	085	270
12	0	085	320
13	0	085	0
14	0	085	040
15	0	085	090
16	0	085	140
17	0	045	180
18	0	035	210
19	0	005	220
20	0	335	210
21	0	315	180
22	0	335	120
23	0	005	140
24	0	035	150

Note:
Deceleration force acts toward subject

FIGURE 13.—Deceleration force vector orientation for Apollo impact tests [9].

through 16, shown in Figure 13 [95]. Accelerations measured on the impact sled ranged from 6.0 to 26.3 g, onset rate varied from 250 to 2130 g/s, and the velocity of the sled at impact time ranged from 5.3 to 13.96 m/s. Acceleration magnitude and onset rate increased simultaneously. Restraint and support systems used in these experiments were similar to those used in the vertical deceleration tower [103]. No persistent or severe subjective complaints were found in 119 of the 146 tests conducted with volunteers. A forward-facing subject tipped back at 45° (position 5) sustained simultaneous compression and hyperflexion of the trunk which produced persistent soft tissue injury in the area of the 6th, 7th, and 8th thoracic vertebrae. Impact was 25 g at 960 g/s in 0.097 s. Blood and urine microscopic and chemical findings were within normal limits for all tests. Fifty-five of 144 ECGs showed significant bradycardia within 51 s after impact of more than 15 g. Incidents of bradycardia were associated with impacts with a $-G_z$ component.

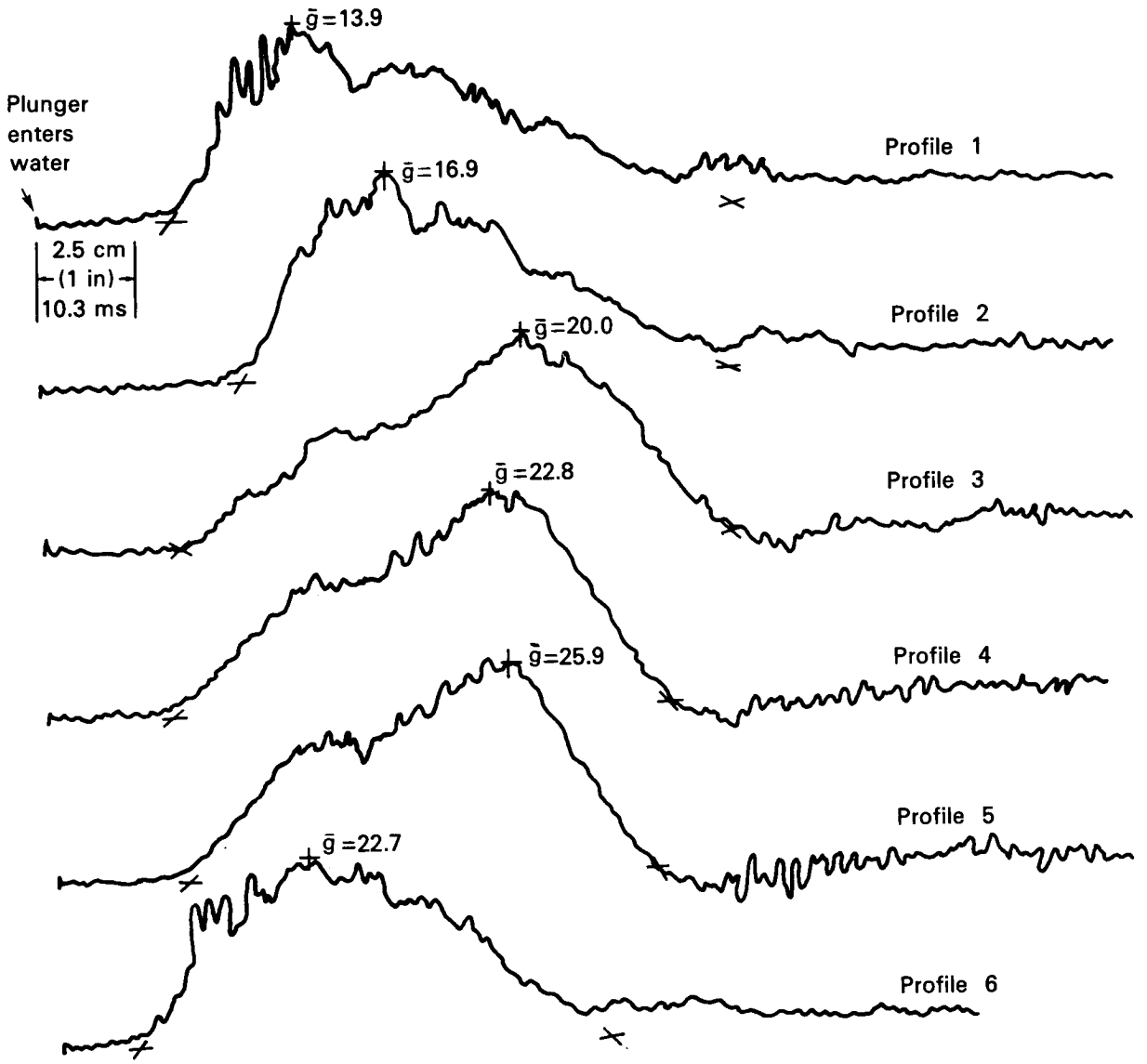


FIGURE 14a.—Vehicle acceleration profiles [103].

This series of impact experiments was later expanded to 288 tests to explore each of 24 positions of the impact vector [9]. Impact acceleration magnitude ranged from 5.5 to 30.7 g, rate of onset varied from 300 to 2500 g/s, and impact velocity ranged from 2.8 to 13.7 m/s. Significant findings of postimpact physical examination are summarized in Table 4.

Data are also available from a series of 11 (volunteer) impact tests conducted on the horizontal decelerator using a less restrictive body support and restraint system [79]. These tests were used to evaluate adequacy of an aircraft restraint harness configuration consisting of shoulder straps, lap belt, and inverted V crotch straps, and a noncontoured seat with a shallow, 5.08–cm deep head support. The configuration proved adequate for impact magnitudes to 14 g with velocity change of 10.9 m/s and onset rate of 1070 g/s.

Missile Impact

Injuries due to the impact of objects propelled by blast pressures, winds, ground or floor shock, and others are dependent upon a number of factors. Among them are mass, velocity, character, density, and impact angle of the projectile whether or not penetration occurs; the area and

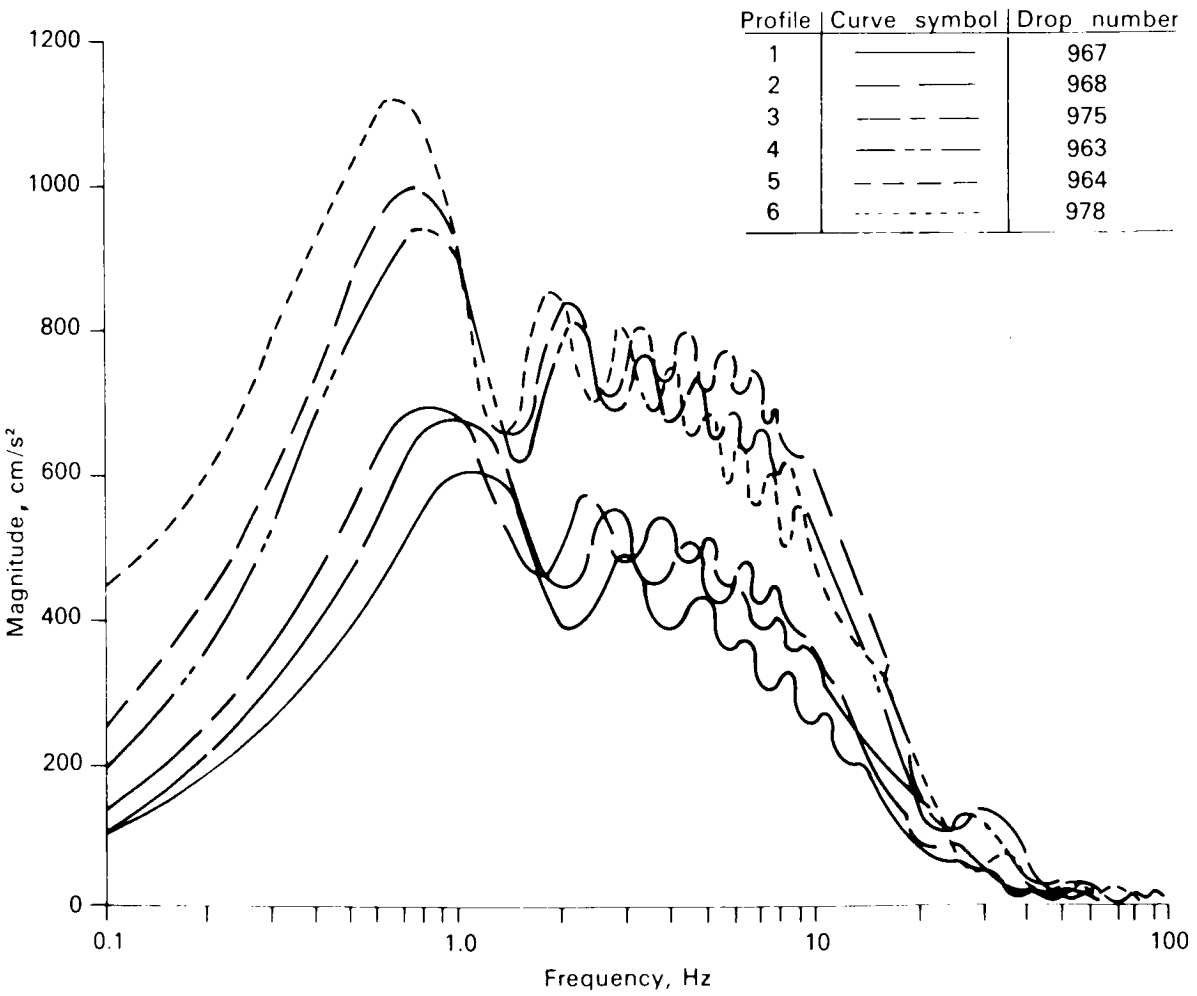


FIGURE 14b.—Power density spectra [103].

organ of the body involved; the amount and kind of clothing; and immunological status and general health condition of the injured individual [14, 105]. Studies of tissue damage by impact of small objects show that the energy of small objects striking a body surface overlying soft tissue is absorbed in the surrounding tissue and does not bring about motion of the whole body [21]. Tentative criteria for missile damage in humans are shown in Table 5.

Impact Protection

Impact protection of man or animal is dependent upon the manner in which impact stress is transmitted to the body and the degree of body support and restraint that have been provided. The method of fixation of the subject to the impacted structure is perhaps the most fundamental consideration. Seat structure and restraint reinforce the body to prevent injurious hyperflexion or hyperextension of anatomical joints and excursions of body organs [63]. Body support and restraint act to distribute impact loads over the body surface. Restraint systems constructed of webbing materials are usually designed to distribute impact loads into the skeletal system. Impact loads should generally be distributed uniformly over as wide an area as possible to avoid concentration of pressure. An exception to this rule would be where the body may act to attenuate the load being transmitted to vital parts, whereas direct coupling might be more injurious. Of many experimental approaches used to provide maximum load distribution, one was to immerse the body in fluid. Effectiveness of this technique to increase tolerance to long-duration acceleration has been demonstrated in centrifuge experiments.

Impact experiments with mice and dogs immersed in water and congealing gypsum have shown that tolerance may be increased up to six times higher than without immersion [63, 64, 65]. Covering the walls of the immersion vessel with porous rubber to attenuate high hydraulic pressure was a critical factor in animal survival [64].

Effects of several other methods of body support and restraint upon the probability of lethality have been demonstrated with guinea pigs [56, 57, 77]. In these experiments, the differences were explored between various degrees of support and restraint ranging from rigid, fully enclosing contoured shell, to a more conventional arrangement of flat seat pan and seat back with a webbing restraint configuration. In one series of experiments [77], guinea pigs were exposed to $+G_x$, $-G_x$, and $+G_z$ accelerations at impact velocities of 12.2, 18.3, and 24.4 m/s in two types of support and restraint systems (SARS). One support and restraint configuration, referred to as SARS IIa or the isovolumetric concept, consisted of a rigid, contoured support and a one-piece fabric apron and retention straps in the shoulder, upper chest, lower abdominal, and crotch regions. The fabric apron covered the ventral thoracic-abdominal area.

The second configuration, referred to as SARS IIIa, consisted of flat plates to provide back support and a seat pan and straps restraining the thoracic and abdominal-crotch areas. Head restraint used on both configurations was identical. The system using the thoracic-abdominal apron was markedly superior in $-G_x$ impacts, slightly superior in $-G_x$ impacts, and approximately equal in the $+G_z$ orientation. Major pathology associated with each of the support and restraint configurations is summarized in Table 6 in terms of occurrence percentage. The 50% probability of lethality values using average g ranged from 209 to 325 for $+G_x$, 287 to 350 for $-G_x$, and 103 to 135 for $+G_z$.

Seating and Other Devices

In many early studies of impact tolerance for spaceflight operations, molded couches of rigid plastic foam were used to support both animal and human subjects. Seats of varying degrees of contouring have been used in spacecraft applications to provide crew protection. Individually molded seats were used in Mercury, Gemini, and Voskhod spacecraft [4]. In Apollo spacecraft, simpler seat structures were used to enhance interchangeability of crew stations throughout long-duration flights [79]. These seat structures are supported within the spacecraft by impact-attenuating struts, shown in Figure 15, a and b. Various attenuation devices which have been studied range from simple, crushable honeycomb structures to more complex, hydraulically

damped spring systems and cyclic strain mechanisms [58, 99].

Impact protection to the crewman can be achieved to the greatest extent within the impact transmission pathway with devices such as externally mounted air bags or internally mounted impact-attenuating struts. Control of the entire pulse shape is essential to optimum protection, in addition to maintaining the most advantageous parameters in magnitude, duration, and mean rate of onset. Depending on elastic properties of the body being accelerated, a waveform can be chosen in which the ratio of the acceleration in the body being accelerated to the magnitude of the acceleration of the body imparting the acceleration will be equal to unity [28]. This can be accomplished by careful design of all acceleration transmission pathways in addition to impact-attenuation devices. Crushing of the vehicle structure provides some energy absorption and this characteristic can be enhanced by the vehicle

TABLE 4.—*Significant Postimpact Physical Exam Findings* [9]

Significant physical findings	Test position	Sled g
Harness burns (all first degree)	2	20.0
	7	23.0
Dazed and disoriented (lasting no longer than 2 min postimpact)	17	17.4
	17	18.9
	17	21.7
	17	25.8
	17	19.6
	19	30.0
	24	28.1
	24	24.6
	24	16.5
	9	9.8
	1	17.2
	21	19.0
Respiratory difficulty (lasting no longer than 1 min postimpact)	17	18.9
	23	19.5
	18	24.6
	18	23.2
	24	23.7
	24	16.5
Blood pressure difference (20 mm Hg at pre- and postrun physical exam)	19	30.0
Pulse difference (20 beats/min at pre- and postrun physical exam)	23	19.4
	17	19.6
	24	20.2
	12	19.5
Engorged retinal vessels	17	21.7
	3	9.2
Back and/or neck pain and decreased range of motion	17	25.8
	1	17.2
	5	25.1
	5	21.0

TABLE 5.—*Tentative Criteria for Indirect Blast Effects Involving Impact from Secondary Missiles* [106]

Missile, type	Critical organ/event	Related impact velocity (m/s)
Nonpenetrating 4.54 kg object	Cerebral concussion:	
	mostly "safe"	3.05
	threshold	4.58
	Skull fracture:	
	mostly "safe"	3.05
	threshold	4.58
	near 100%	7.02
Penetrating 10-g glass fragments	Skin laceration[1]:	
	threshold	15.3
	Serious wounds[1]:	
	threshold	30.5
	50%	54.9
	near 100%	91.5

[1] Represent impact velocities with unclothed skin. A serious wound arbitrarily defined as a laceration of the skin with missile penetration into tissues to 10 mm or more.

TABLE 6.—*Occurrence Percent of Major Pathology in Guinea Pigs at Cumulative Velocities*[1] [77]

Injury type	SARS IIIa			SARS IIa		
	$-G_x$	$+G_x$	$+G_z$	$-G_x$	$+G_x$	$+G_z$
Brain hemorrhage	42	61	30	42	91	22
Pulmonary hemorrhage	91	82	80	74	100	32
Cardiovascular pathology	33	52	48	19	0	0
Hepatic laceration	83	56	5	45	3	2
Gastrointestinal pathology	77	19	15	19	30	17
Paralysis	0	0	18	0	0	80
Total nonsurvivors	180			161		

[1] Entrance velocities, 12 to 24 m/s.

structure designer. Deformation of cushioning materials and restraint system can also be designed to minimize transmittal of energy at frequencies where particular segments of the human body, such as the head, are most sensitive; however, care must be taken to assure that these elements of the protective system do not, in fact, amplify accelerations transmitted to the body.

Control of impact vector direction can be used to take advantage of differences in impact tolerance levels for each body axis. Design of seat angles may also be critical in providing maximum tolerance [8].

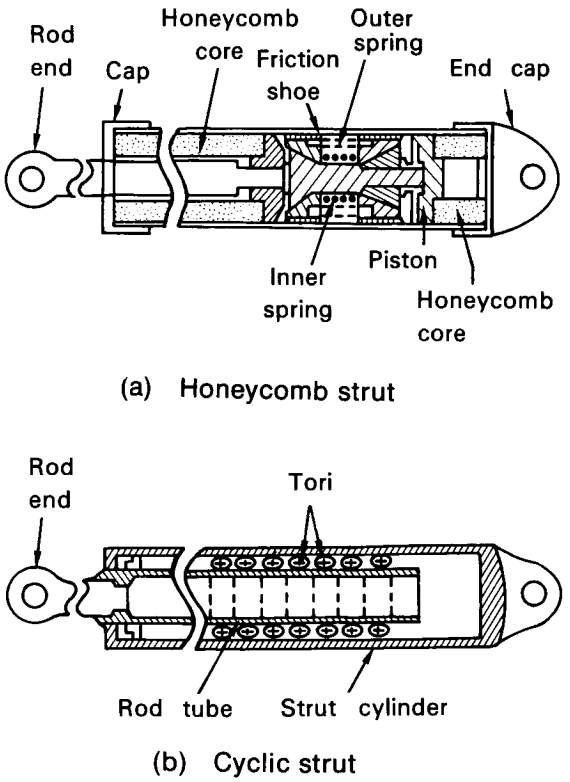


FIGURE 15.—Comparison of honeycomb and cyclic strain impact-attenuation systems for Apollo [98].

Mathematical models of both protection system and human body have greatly improved the designer's capability to select appropriate materials for crew seat cushioning and restraint systems, and impact-attenuation device performance characteristics [71, 72]. The same modeling techniques provided insight into effects of initial conditions of the crewman within his personal equipment. For example, these analytical techniques demonstrated the importance of eliminating slack or deadspace between crewman and his body-support and restraint system, and similarly provided design criteria for restraint-harness tensioning devices.

Other methods of crew protection include crew conditioning and use of pharmacological agents. Crew conditioning has been considered from several aspects. First, by assuring the best physical condition of the crewman through a sound program of preflight physical exercise. Second, by alterating the crewman's reaction to impact through crew training and exposure to mechanical stresses during simulated missions [28]. And finally, where long-duration missions may cause deconditioning of the musculoskeletal system, exercise and in the future, perhaps, use of chemotherapy to retard deconditioning are indicated.

Summary

The degree to which impact accelerations are an important factor in spaceflight environments depends primarily upon the technology of capsule landing deceleration and the weight permissible for the associated hardware: parachutes or deceleration rockets, inflatable air bags, or other impact-attenuation systems. Safe capsule landings on any type of terrestrial and extraterrestrial surface must be the goal of these hardware developments so that the restrictions imposed on most USSR and US space missions in the past can be relaxed. However, design for emergency situations such as crew escape during unforeseen failure on the launch pad will always require the most accurate information available on the limits of human tolerance and risk involved.

A considerable body of information has been available on human tolerance to impact and impact protection from aircraft escape, and aviation, as well as automotive crash research. However, both the USSR and US space programs had to define specific limits of human tolerance with higher accuracy and reliability than were previously known. Particular contributions in this area include: (a) exploration of impact tolerance for all impact directions; (b) definition of injury prob-

ability for low injury probabilities consistent with high reliability/safety requirements of space missions; and (c) development of mathematical models to predict injury probability for complex acceleration functions, and to calculate the crewman's biodynamic response when coupled to various support and restraint systems. These advances, as well as experience with new impact-attenuating crushable materials and structures, are of significance beyond the specific realm of space biotechnology.

The problem most specific to space medicine is the potential change of impact tolerance due to reduced bone mass and muscle stength caused by prolonged weightlessness and physical inactivity. Although valuable contributions to this area have been made through animal experimentation in the USSR and the US, considerably more research is required as space missions become extended over many weeks and months. Relationships between bone strength, bone mass, and muscle strength must be explored as a function of gravitational load, isotonic/isometric exercise, time pattern, and diet. For osteoporosis of disuse, appropriate time-scaling factors for bone dynamics as a function of gravitational exposure and activity time patterns must be established by relating animal experiments to human conditions. Changes in injury patterns due to these changes in the musculoskeletal system must be known and understood. Based on such studies, proper impact limit values, protection equipment, preventive measures such as exercise and possibly chemotherapy, and postflight care can be selected.

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