CHAPTER 42

EFFECTS OF SHOCK AND VIBRATION ON HUMANS

Henning E. von Gierke
Anthony J. Brammer

INTRODUCTION

This chapter considers the following problems: (1) the determination of the structure and properties of the human body regarded as a mechanical as well as a biological system, (2) the effects of shock and vibration forces on this system, (3) the protection required by the system under various exposure conditions and the means by which this protection is to be achieved, and (4) tolerance criteria for shock and for vibration exposure. Man, as a mechanical system, is extremely complex and his mechanical properties readily undergo change. There is limited reliable information on the magnitude of the forces required to produce mechanical damage to the human body. To avoid damage to humans while obtaining such data, it is necessary to use cadavers, experimental animals, or simulations for most studies on mechanical injury. However, the data so obtained must be subjected to careful scrutiny to determine the degree of their applicability to humans. Occasionally it is possible to obtain useful information from situations involving accidental injuries to man, but while the damage often can be assessed, the forces producing the damage usually cannot, and so only rarely are useful data obtained in this way. It is also very difficult to obtain reliable data on the effects of mechanical forces on the performance of various tasks and on subjective responses to these forces largely because of the wide variation in the human being in both physical and behavioral respects. Measurement of some of the mechanical properties of man is, however, often practicable since only small forces are needed for such work. The difficulty here is in the variability and lability of the system and in the inaccessibility of certain structures.

One section of this chapter introduces methods used for mechanical shock and vibration studies on man and animals. Subsequent sections deal with the mechanical characteristics of the body, the effects of shock and vibration forces on man, the methods and procedures for protection against these forces, and safety criteria.

For general background material on the effects of shock and of vibration on man, see Refs. 1 through 4.

DEFINITIONS AND CHARACTERIZATION OF FORCES

Characterization of Forces. Forces may be transmitted to the body through a gas, liquid, or solid. They may be diffuse or concentrated over a small area. They may vary from tangential to normal and may be applied in more than one direction. The shape of a solid body impinging on the surface of the human is as important as the position or shape of the human body itself. All these factors must be taken into account in comparing injuries produced by vehicle crashes, explosions, blows, vibration, etc. Laboratory studies often permit fairly accurate control of force application, but field situations are apt to be extremely complex. Therefore it is often very difficult to predict what will happen in the field on the basis of laboratory studies. It is equally difficult to interpret field observations without the benefit of laboratory studies.

Shock. The term *shock* is used differently in biology and medicine than in engineering; therefore one must be careful not to confuse the various meanings given to the term. In this chapter the term *shock* is used in its engineering sense as defined in Chap. 1 of this Handbook, that is, for a nonperiodic excitation characterized by suddenness and severity. A *shock wave* is a discontinuous pressure change propagated through a medium at velocity greater than that of sound in the medium. In general, forces reaching peak values in less than a few tenths of a second and of not more than a few seconds' duration may be considered as shock forces in relation to the human system.

The term *impact* (i.e., a *blow*) refers to a force applied when the human body comes into sudden contact with a solid body and when the momentum transfer is considerable, as in rapid deceleration in a vehicle crash or when a rapidly moving solid body strikes a human body.

Vibration. Biological systems may be influenced by vibration at all frequencies if the amplitude is sufficiently great. This chapter is concerned primarily with the frequency range from about 1 Hz to 1 kHz.

METHODS AND INSTRUMENTATION

Most quantitative investigations of the effects of shock and vibration on humans are conducted in the laboratory in controlled, simulated environments. Meaningful results can be obtained from such tests only if measurement methods and instrumentation are adapted to the particular properties of the biological system under investigation to ensure noninterference of the measurement with the system's behavior. This behavior may be physical, physiological, and psychological although these parameters should be studied separately if possible. The complexity of a living organism makes such separation, even assuming independent parameters, only an approximation at best. In many cases if extreme care is not exercised in planning and conducting the experiment, uncontrolled interaction between these parameters can lead to completely erroneous results. For example, the dynamic elasticity of tissue of a certain area of the body may depend on the simultaneous vibration excitation of other parts of the body; or the elasticity may change with the duration of the measurement since the subject's physiological response varies; or the elasticity may be influenced by the subject's psychological reaction to the test or to the measurement equipment.

Control of, and compensation for, the nonuniformity of living systems is absolutely essential because of the variation in size, shape, sensitivity, and responsiveness of people and because these factors, for a single subject, vary with time, experience, and motivation. The use of adequate statistical experimental design is necessary and almost always requires a large number of observations and carefully arranged controls.

A range of mechanical and hydraulic vibration exciters have been developed specifically for human laboratory experiments with extensive safety systems. Similarly, acceleration and deceleration sleds have been developed for use in impact tests with human subjects.

PHYSICAL MEASUREMENTS

In determining the effects of shock and vibration on humans, the mechanical force environment to which the human body is exposed must be clearly defined. Force and vibration amplitudes should be specified for the area of contact with the body. Vibration measurements of the body's response should be made whenever possible by noncontact methods. X-ray methods can be used successfully to measure the displacement of internal organs. Optical, cinematographic, and stroboscopic observation can give the displacement amplitudes of parts of the body. If vibration pickups in contact with the body are used, they must be small and light enough so as not to introduce a distorting mechanical load. This usually places a weight limitation on the pickup of a few grams or less, depending on the frequency range of interest and the effective mass to which the pickup is attached. Figure 42.1 illustrates the effect of

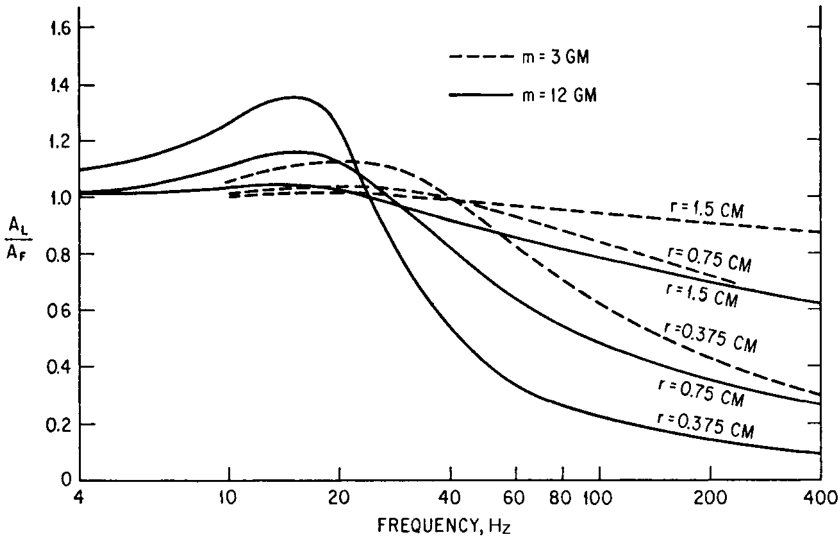

FIGURE 42.1 Amplitude distortion due to accelerometers of different mass m and size which are attached to a body surface over soft tissue of human subject exposed to vibration. The graph gives the ratio A_L/A_F of the response of the loaded to the unloaded surface for accelerometers having three different radii r. (*Values calculated from unpublished mechanical surface impedance data of E. K. Franke and H. E. von Gierke.*)

mass and size on the response of accelerometers attached to the skin overlying soft tissue. The lack of rigidity of the human body as a supporting structure makes measurements of acceleration usually preferable to those of velocity or displacement. The mechanical impedance of a sitting, standing, or supine subject is extremely useful for calculating the vibratory energy transmitted to the body by the vibrating structure. The mechanical impedance of small areas of the body surface can be measured in different ways (see Chap. 12), for example by vibrating pistons, resonating rods, and acoustical impedance tubes.

If the entire body is exposed to a pressure or blast wave in air or water, exact definition of the pressure environment is essential. The pressure distribution should be measured if possible. If the environment deviates from free-field conditions, it should be carefully specified because of its effect on peak pressure and pressure *vs.* time-history.

SIMULATION OF HUMAN SUBJECTS

The establishment of limits of human tolerance to mechanical forces, and the explanation of injuries produced when these limits are exceeded, frequently requires experimentation at various degrees of potential hazard. To avoid unnecessary risks to humans, animals are used first for detailed physiological studies. As a result of these studies, levels may be determined which are, with reasonable probability, safe for human subjects. However, such comparative experiments have obvious limitations. The different structure, size, and weight of most animals shift their response curves to mechanical forces into other frequency ranges and to other levels than those observed on humans. These differences must be considered in addition to the general and partially known physiological differences between species. For example, the natural frequency of the thorax-abdomen system of a human subject is between 3 and 4 Hz; for a mouse the same resonance occurs between 18 and 25 Hz. Therefore maximum effect and maximum damage occur at different vibration frequencies and different shock-time patterns in a mouse than in a human. However, studies on small animals are well worth making if care is taken in the interpretation of the data and if scaling laws are established. Dogs, pigs, and primates are used extensively in such tests.

Many kinematic processes, physical loadings, and gross destructive anatomical effects can be studied on dummies which approximate a human being in size, form, mobility, total weight, and weight distribution in body segments. In contrast to those used only for load purposes, dummies simulating basic static and dynamic properties of the human body are called *anthropometric* or *anthropomorphic* dummies. Several such dummies have been designed for specific simulations.[5] For automobile frontal collisions, the Hybrid III dummy shown in Fig. 42.2 has become the de facto standard, and is used in North America and Europe to simulate occupants in crash tests and tests of safety restraint systems. The original dummy was constructed to correspond to a 50th-percentile North American male. It possesses a metal “skeleton” covered with a vinyl skin and foam to produce the appropriate external shape, with a rubber lumbar spine curved to mimic a sitting posture, and a shoulder structure capable of supporting safety belt loads. The head, neck, chest, and knee responses of the Hybrid III are designed to mimic human responses, namely, the head acceleration resulting from forehead and side-of-the-head impacts, the fore-and-aft and lateral bending of the neck, the deflection of the chest to distributed forces on the sternum, and impacts on the knee.[4] The instrumentation required to record these responses together with local axial compressional loads is described in Fig. 42.2 and

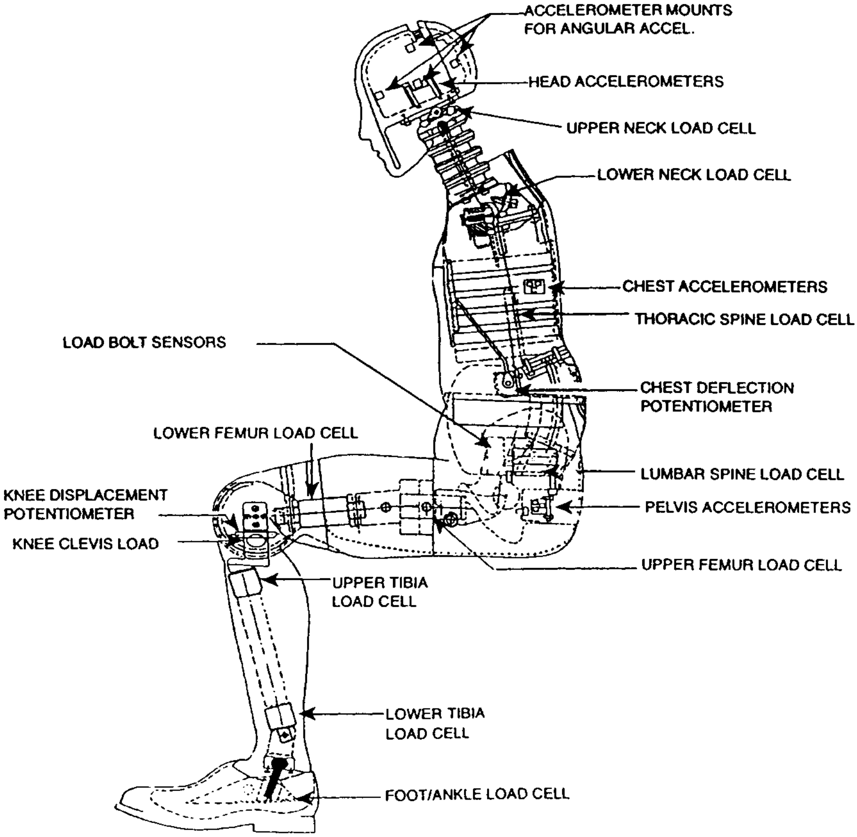

FIGURE 42.2 Hybrid III dummy designed for use in motor vehicle frontal crash tests. (*AGARD-AR-330.*[5])

results in 37 channels of data if head rotation is included. Hybrid III dummies are now available representing adult-sized small (5th-percentile) females and large (95th-percentile) males, as well as infants and children. A related dummy, SID (for Side-Impact Dummy), is available for automobile side collisions together with dummies developed for this purpose in Europe [EUROSID-1 and BIOSID (BIOfidelic Side-Impact Dummy)]. An advanced dummy, ADAM (Advanced Dynamic Anthropomorphic Manikin), has been developed for use in aircraft ejection seats, helicopter seats, and parachute tests. In addition to modeling body segments, surface contours, weights, centers of gravity, moments of inertia, and joint center locations, ADAM replicates human joint motion and the biodynamic response of the spine to vertical accelerations for both small-amplitude vibration and large impacts.

Efforts have also been made to simulate the mechanical properties of the human head in order to study the physical phenomena occurring in the brain during crash conditions. Although these head forms only approximate the human head, they are useful in the evaluation of the protective features of crash, safety, and antibuffet helmets. Plastic head forms, conforming to standard head measurements, are designed to fracture in the same energy range as that established for the human head. A cranial vault is provided to house instrumentation and a simulated brain mass with comparable weight and consistency (e.g., a mixture of glycerin, ethylene glycol, etc.). The static properties of the skin and scalp tissue are simulated with polyvinyl foam.

The static and dynamic breaking strength of bones, ligaments, and muscles and the forces producing fractures in rapid decelerations have been studied frequently on cadaver material. Extreme caution must be exercised in applying elastic and strength properties obtained in this way to a situation involving the living organism. The differences observed between properties of wet, dry, and embalmed materials are considerable; changes in these properties also result in changes in the force distribution of a composite structure. Thus a number of physiological, anatomical, and physical factors must be considered for each specific situation in which the use of animals, dummies, or cadavers as substitutes for live human subjects is planned.

MECHANICAL CHARACTERISTICS OF THE BODY

PHYSICAL DATA

This section summarizes the passive mechanical responses of the human body and tissues exposed to vibration and impact. The data can be used to calculate quantitatively the transmission and dissipation of vibratory energy in human body tissue, to estimate vibration amplitudes and pressures at different locations of the body, and to predict the effectiveness of various protective measures. Table 42.1 lists some dynamic mechanical characteristics of the body and indicates some of the fields where these data find application. In cases where detailed quantitative investigations are lacking, the information may serve as a guide for the explanation of observed phenomena or for the prediction of results to be expected. Most physical characteristics of the human body presented in this section (except for the strength data) have been derived from the analysis of experimental data in which it is assumed that the body is a linear, passive mechanical system. This is an idealization which holds only for very small amplitudes. Therefore these data may not apply in analyses of mechanical injury to tissue. There is actually considerable nonlinearity of response well below amplitudes required for the production of damage. This is indicated, for example, by the data given in Fig. 42.3, which shows how the mechanical stiffness and resistance of soft tissue vary with static deflection. Bone behaves more or less like a normal solid; however, soft elastic tissues such as muscle, tendon, and connective tissue resemble elastomers with respect to their Young's modulus and S-shaped stress-strain relation. These properties have been studied in connection with the quasi-static pressure-volume relations of hollow organs such as arteries, the heart, and the urinary bladder, assuming linear properties in studying dynamic responses. Then soft tissue can be described phenomenologically as a viscoelastic medium; plastic deformation need be considered only if injury occurs. Physical properties of human body tissue are summarized in Table 42.2 for frequencies less than 100 kHz.

The fatigue life of bone and soft tissue in response to cyclic dynamic stress at frequencies between 0.5 and 4 Hz is summarized in Fig. 42.4. In this diagram, the number of cycles to failure N of in vitro preparations is expressed as a function of the ratio of the applied dynamic stress to the ultimate static stress σ/σ_u. The straight lines

TABLE 42.1 Application of Mechanical Studies of Body

Dynamic mechanical quantity investigated	Field of application
Skull resonances and viscosity of brain tissue	Head injuries; bone-conduction hearing
Impedance of skull and mastoid	Matching and calibration of bone-conduction transducers; ear protection
Sound transmission through skull and tissue	Bone-conduction hearing
Mechanical properties of outer, middle, and inner ear	Theory of hearing; correction of hearing deficiencies
Resonances of mouth, nasal, and pharyngeal cavities	Theory of speech generation; correction of speech deficiencies; oxygen mask design
Resonance of lower jaw	Bone-conduction hearing
Response of mouth-thorax system	Blast-wave injury; respirators
Propagation of pulsed cardiac pressure	Circulatory physiology; hemodynamics
Heart sounds	Physiology of heart; diagnosis
Suspension of heart	Ballistocardiography; injury from severe vibrations and crash
Response of the thorax-abdominal mass system	Severe vibration and crash injury; crash protection
Impedance of subject sitting, standing, or lying on vibration platform	Isolation and protection against vibration and short-time accelerations; ballistocardiography
Hand-arm impedance	Isolation and protection against hand-transmitted vibration; design of power tools; design of test fixtures for hand-tool vibration assessment
Impedance of body surface, surface wave velocity, sound velocity in tissue, absorption coefficient of body surface	Theory of energy transmission and attenuation in tissue; determination of tissue elasticity, viscosity and compressibility; determination of acoustic and vibratory energy entering the body; vibration isolation; design of vibration pickups; transfer of vibratory energy to inner organs and sensory receptors; soft tissue and organ imaging

in Fig. 42.4 represent the functions $N = (\sigma/\sigma_u)^{-x}$, where the value of the index x in the relationship is indicated.

The combination of soft tissue and bone in the structure of the body together with the body's geometric dimensions results in a system which exhibits different types of response to vibratory energy depending on the frequency range: At low frequencies (below approximately 100 Hz), the body can be described for most purposes as a lumped parameter system; resonances occur due to the interaction of tissue masses with purely elastic structures. At higher frequencies, through the audio-frequency range and up to about 100 kHz, the body behaves more as a complex distributed parameter system—the type of wave propagation (shear waves, surface waves, or compressional waves) being strongly influenced by boundaries and geometrical configurations.

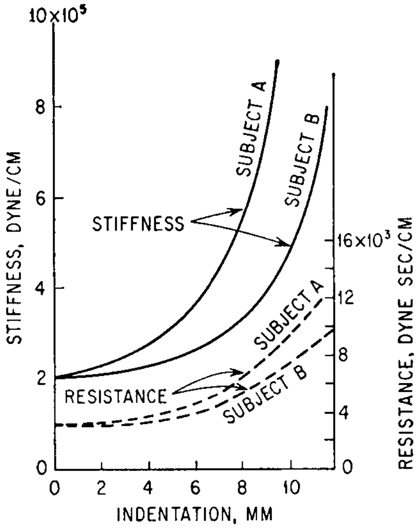

FIGURE 42.3 Mechanical stiffness and resistance of soft tissue, per square centimeter, as a function of indentation (i.e., static deflection). The nonlinearity shows the effect of loading of body surface on surface impedance of soft tissues for two experimental human subjects *A* and *B*. (*After Franke: USAF Tech. Rept. 6469, 1959.*)

LOW-FREQUENCY RANGE

Simple mechanical systems, such as the one shown in Fig. 42.5 for a standing and sitting man, are usually sufficient to describe and understand the important features of the response of the human body to low-frequency vibrations.[6,7] Nevertheless it is difficult to assign numerical values to the elements of the model, since they depend critically on the kind of excitation, the body type of the subject, and his posture and muscle tone. Large intersubject variability is therefore to be expected and is observed. Of the various factors influencing whole-body biodynamic responses, a reduction in intersubject variability can often be obtained by normalizing measured values by a subject's static mass.[1]

Subject Exposed to Vibrations in the Longitudinal Direction. The *mechanical impedance* of a man standing or sitting on a vertically vibrating platform, that is, the complex ratio between the dynamic force applied to the body and the velocity at the interface where vibration enters the body, is shown in Fig. 42.6. Below approximately 2 Hz the body acts as a unit mass. For the sitting man, the first resonance is between 4 and 6 Hz; for the standing man, resonance peaks occur at about 6 and 12 Hz. The numerical value of the impedance together with its phase angle provides data for the calculation of the total energy transmitted to the subject.

TABLE 42.2 Physical Properties of Human Tissue at Frequencies Less Than 100 kHz

		Bone, compact	
	Tissue, soft	Fresh	Embalmed, dry
Density, gm/cm^3	1–1.2	1.93–1.98	1.87
Young's modulus, dyne/cm^2	7.5×10^4	2.26×10^{11}	1.84×10^{11}
Volume compressibility,* dyne/cm^2	2.6×10^{10}	...	1.3×10^{11}
Shear elasticity,* dyne/cm^2	2.5×10^4	...	7.1×10^{10}
Shear viscosity,* dyne-sec/cm^2	1.5×10^2	...	...
Sound velocity, cm/sec	$1.5–1.6 \times 10^5$	3.36×10^5	...
Acoustic impedance, dyne-sec/cm^3	1.7×10^5	6×10^5	6×10^5
Tensile strength, dyne/cm^2	...	9.75×10^8	1.05×10^9
Shearing strength, dyne/cm^2, parallel	...	4.9×10^8	...
Shearing strength, dyne/cm^2, perpendicular	...	1.16×10^9	5.55×10^8

* Lamé elastic moduli.

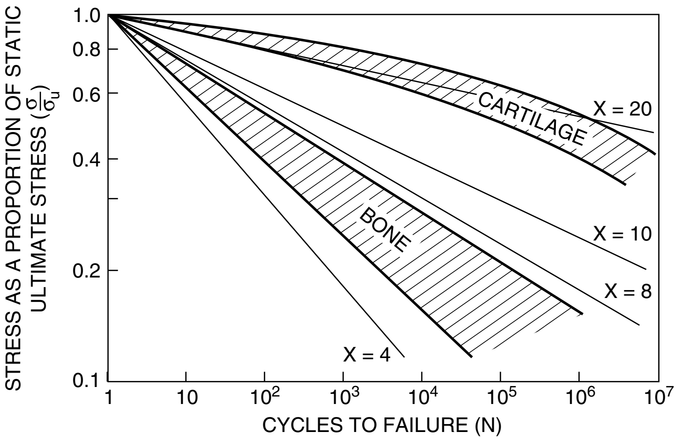

FIGURE 42.4 Fatigue failure of human tissue. The number of cycles of repeated stress N to failure of *in vitro* preparations is shown as a function of the ratio of the applied dynamic stress to the ultimate static stress σ/σ_u. (*von Gierke.*[6])

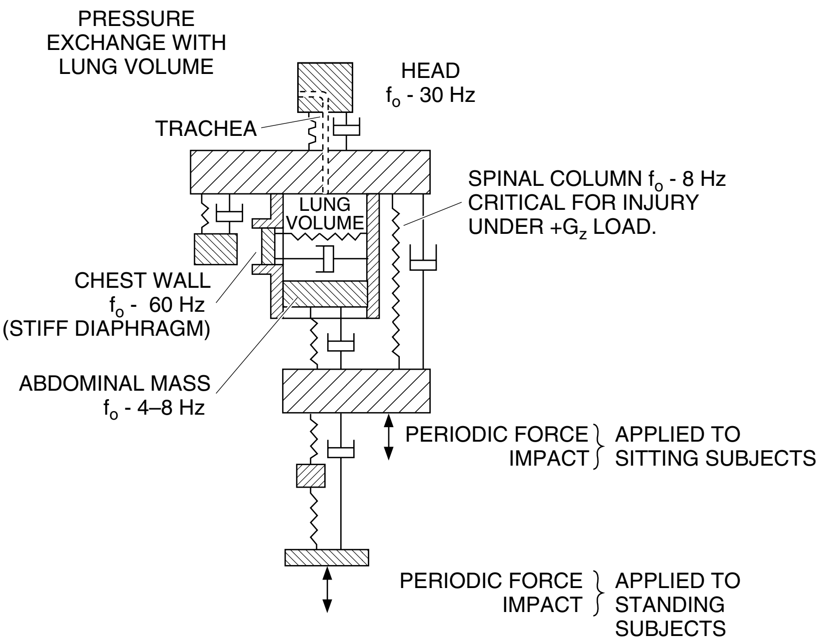

FIGURE 42.5 Lumped parameter biodynamic model of the standing and sitting human body for calculating motion of body parts and some physiological and subjective responses to vertical vibration. The approximate resonance frequencies of various subsystems are indicated by f_o. (*von Gierke.*[6])

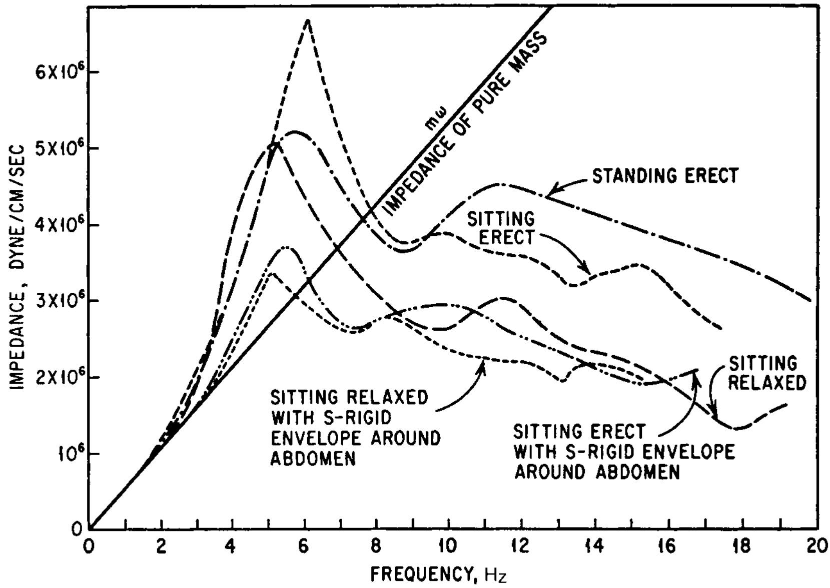

FIGURE 42.6 Mechanical impedance of a standing and sitting human subject vibrating in the direction of his longitudinal axis as a function of frequency. The effects of body posture and of a semirigid protective envelope around the abdomen are shown. The impedance of a mass *m* also is given. (*After Coermann:* Human Factors, ***4****:227, 1962, and Coermann et al.:* Aerospace Med., ***31****:443, 1960.*)

The resonances at 4 to 6 Hz and 10 to 14 Hz are suggestive of mass-spring combinations of (1) the entire torso with the lower spine and pelvis and (2) the upper torso with forward flexion movements of the upper vertebral column. The expectation that flexion of the upper vertebral column occurs is supported by observations of the transient response of the body to vertical impact loads and associated compression fractures. The greatest loads occur in the region of the twelfth thoracic to the second lumbar vertebra, which therefore can be assumed as the hinge area for flexion of the upper torso. Since the center-of-gravity of the upper torso is considerably forward of the spine, flexion movement will occur even if the force is applied parallel to the axis of the spine. Changing the direction of the force so that it is applied at an angle with respect to the spine (for example, by tilting the torso forward) influences this effect considerably. Similarly the center-of-gravity of the head can be considerably in front of the neck joint which permits forward-backward motion. This situation results in forward-backward rotation of the head instead of pure vertical motion.

Between 20 and 30 Hz the head exhibits a mechanical resonance. When subject to vibration in this range, the head displacement amplitude can exceed the shoulder amplitude by a factor of 3. This resonance is of importance in connection with the deterioration of visual acuity under the influence of vibration. Another frequency range of disturbances between 60 and 90 Hz suggests an eyeball resonance.

Typical values of mechanical impedance and *seat-to-head transmissibility,* that is, the ratio of the response amplitude and phase of the head to steady-state forced vibration of a seated person at that frequency, are described in an international standard.[8] They are based on a synthesis of measured values from different experimental studies, each of which was conducted under closely related, controlled measurement conditions, and involved a number of male subjects. The need for precise definition of measurement conditions, and hence the restricted applicability of the results, stems from the dependence of the biodynamic responses on body shapes (e.g., mass and height), posture, support (i.e., of buttocks, back, and/or feet), and state of ankle and knee joints. The remaining *unexplained* differences between the results of these studies, a situation commonly encountered in biodynamic experiments, led to the specification of the most probable values for the impedance and transmissibility as a function of frequency by an upper and lower envelope that encompasses the mean values of *all* data sets. The envelopes, which are shown by the thick continuous lines in Figs. 42.7 and 42.8, define a range of idealized values that characterize the bio-

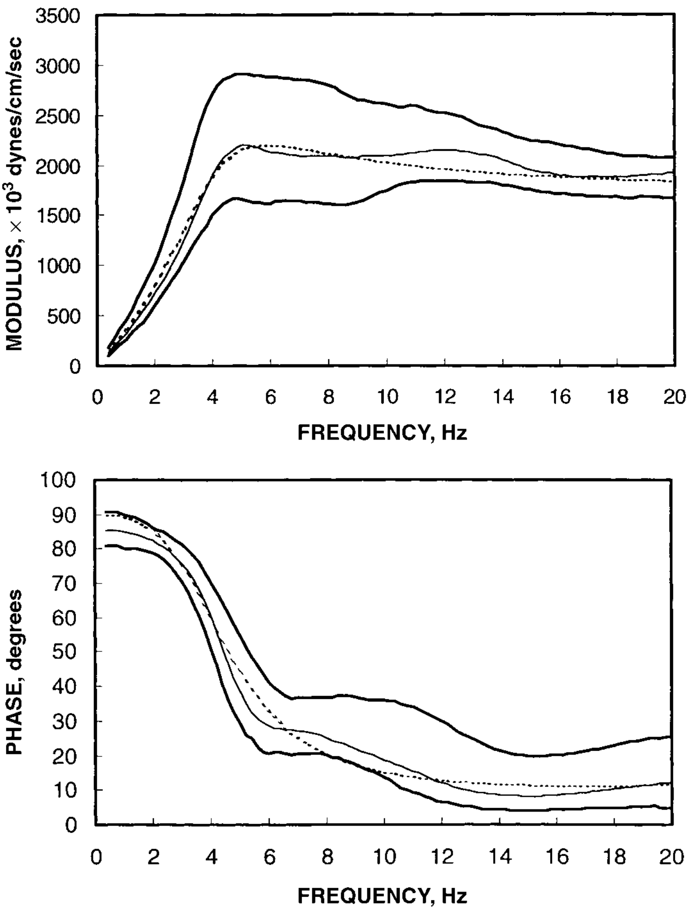

FIGURE 42.7 Driving-point mechanical impedance of the seated human body in the vertical direction (z-direction of Fig. 42.24), expressed as magnitude and phase. Maximum and minimum envelopes of mean values from the studies included in the data synthesis are shown by thick continuous lines, while the mean of these data sets is shown by the thin continuous line. The response of a three-degree-of-freedom biodynamic model is shown by the dashed line. (*ISO 5982.*[8])

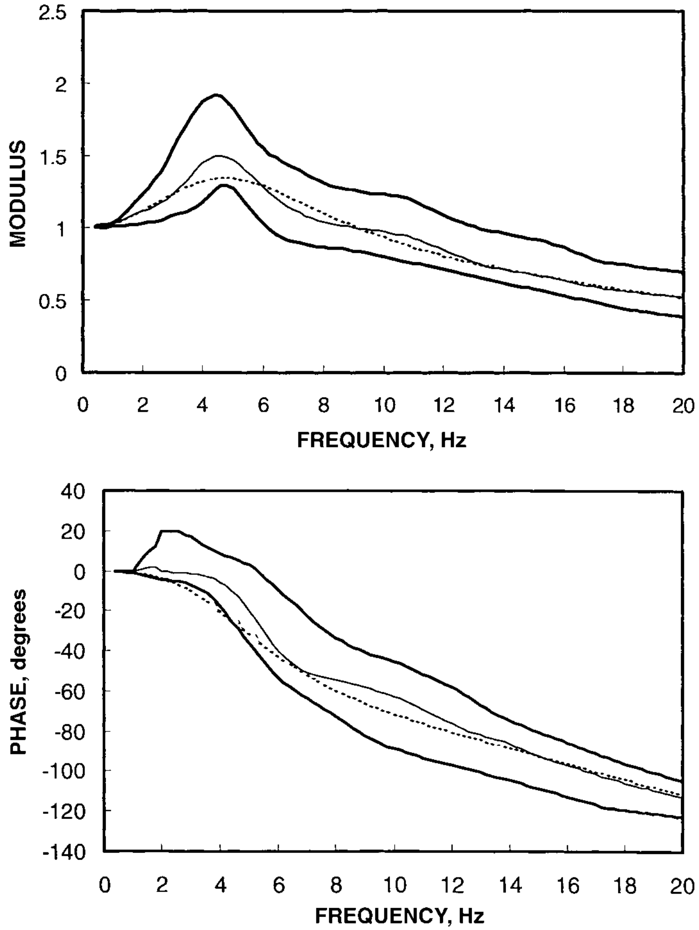

FIGURE 42.8 Seat-to-head transmissibility of the seated human body in the vertical direction (z-direction of Fig. 42.24), expressed as magnitude and phase. Maximum and minimum envelopes of mean values from the studies included in the data synthesis are shown by thick continuous lines, while the mean of these data sets is shown by the thin continuous line. The response of a three-degree-of-freedom biodynamic model is shown by the dashed line. (*ISO 5982.*[8])

dynamic response of a seated person when the back is unsupported and the feet are resting on a surface supporting a rigid seat. Note that data from some individuals will fall outside the range between the two envelopes, as a consequence of their definition (compare with Fig. 42.6). The mean value of all data sets is shown by the thin continuous line in these diagrams, and serves as a target for applications, such as mechanical simulation of the response of the seated human body to vertical vibration, or the development of seats for reducing impacts transmitted to the body. Also shown by the dotted lines in Figs. 42.7 and 42.8 are values calculated using the biodynamic model illustrated in Fig. 42.9.[8] The components of the model do not correspond to

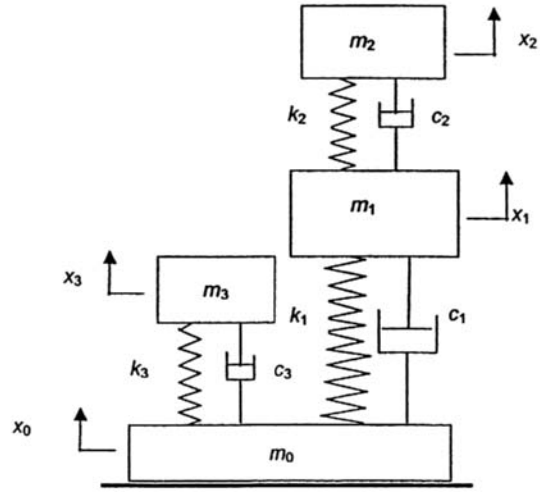

FIGURE 42.9 Three degree-of-freedom biodynamic model for the driving point mechanical impedance, apparent mass, and seat-to-head transmissibility of the seated human body in the vertical direction (z-direction of Fig. 42.24). The model is driven at its base (x_0). The parameters of this model do not possess direct anatomical correlates. (*ISO 5982.*[8])

those of identifiable body parts, though the motion of mass m_2 is taken to represent that of the head for the calculation of seat-to-head transmissibility.

The mechanical impedance of the human body, lying on its back on a rigid surface and vibrating in the direction of its longitudinal axis, has been determined in connection with ballistocardiograph studies. For tangential vibration, the total mass of the body behaves as a simple mass-spring system with the elasticity and resistance of the skin. For the average subject the resonance frequency is between 3 and 3.5 Hz, and the Q of the system is about 3. If the subject's motion is restricted by clamping the body at the feet and at the shoulders between plates connected to the table, the resonance is shifted to approximately 9 Hz and the Q is about 2.5.

One of the most important subsystems of the body, which is excited in the standing and sitting positions as well as in the lying position, is the thorax-abdomen system. The abdominal viscera have a high mobility due to the very low stiffness of the diaphragm and the air volume of the lungs and the chest wall behind it. Under the influence of both longitudinal and transverse vibration of the torso, the abdominal mass vibrates in and out of the thoracic cage. Vibrations take place in other than the (longitudinal) direction of excitation; during the phase of the cycle when the abdominal contents swing toward the hips, the abdominal wall is stretched outward and the abdomen appears larger in volume; at the same time, the downward deflection of the diaphragm causes a decrease of the chest circumference. At the other end of the cycle the abdominal wall is pressed inward, the diaphragm upward, and the chest wall is expanded. This periodic displacement of the abdominal viscera has a sharp resonance between 3 and 3.5 Hz, as can be seen from Fig. 42.10. The oscillations of the abdominal mass are coupled with the air oscillations of the mouth-chest system. Measurements of the impedance of the latter system at the mouth (by applying oscillating air pressure to the mouth) show that the abdominal wall and the anterior

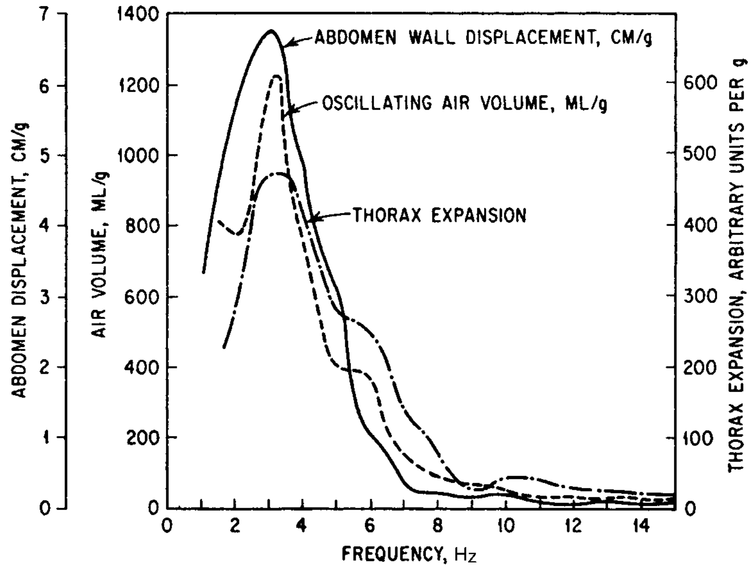

FIGURE 42.10 Typical response curves of the thorax-abdomen system of a human subject in the supine position exposed to longitudinal vibrations. The displacement of the abdominal wall (2 in. below umbilicus), the air volume oscillating through the mouth, and the variations in thorax circumference are shown per g longitudinal acceleration. (*Coermann et al.:* Aerospace Med., ***31***:*443, 1960.*)

chest wall respond to this pressure. The magnitude of the impedance is minimum and the phase angle is zero between 7 and 8 Hz. The abdominal wall has a maximum response between 5 and 8 Hz, the anterior chest wall between 7 and 11 Hz. Vibration of the abdominal system resulting from exposure of a sitting or standing subject is detected clearly as modulation of the air flow velocity through the mouth (Fig. 42.10). Therefore at large amplitudes of vibration, speech can be modulated at the exposure frequency. A lumped parameter model of the thorax-abdomen-airway system is used successfully to explain and predict these detailed physiological responses (Fig. 42.11).[7] The same model can also be used, when appropriately excited, to describe the effects of blast, infrasound, and chest impact and to derive curves of equal injury potential, i.e., tolerance curves.

Subject Exposed to Vibrations in the Transverse Direction. The physical response of the body to transverse vibration—i.e., horizontal in the normal upright position—is quite different from that for vertical vibration. Instead of thrust forces acting primarily along the line of action of the force of gravity on the human body, they act at right angles to this line. Therefore the distribution of the body masses is of the utmost importance. There is a greater difference in response between sitting and standing positions for transverse vibration than for vertical vibration where the supporting structure of the skeleton and the spine are designed for vertical loading.

For a standing subject, the displacement amplitudes of vibration of the hip, shoulder, and head are about 20 to 30 percent of the table amplitude at 1 Hz and decrease with increasing frequency. The sitting subject exhibits amplification of the hip (1.5 Hz) and head (2 Hz) amplitudes. All critical resonant frequencies are between 1 and

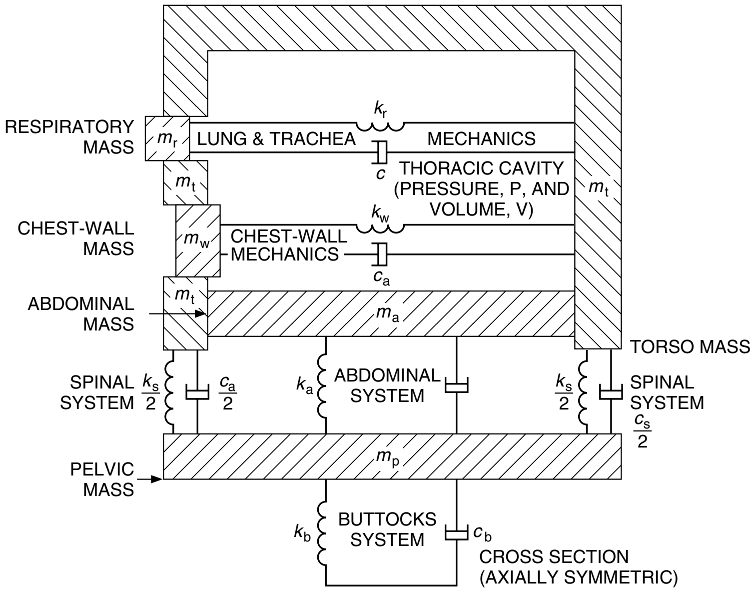

FIGURE 42.11 Five degree-of-freedom body model. The model is used to calculate body deformations (thorax compression, pressure in the lungs, airflow into and out of the lungs, diaphragm and abdominal mass movement) as a function of external longitudinal forces (vibration or impact) and pressure loads (blast, infrasonic acoustic loads). It has also been used to calculate thorax dynamics under impacts to the chest wall, m_w. (*von Gierke*.[7])

3 Hz. The transverse vibration patterns of the body can be described as standing waves, i.e., as a rough approximation one can compare the body with a rod in which transverse flexural waves are excited. Therefore there are nodal points on the body which become closer to the feet as the frequency of excitation increases, since the phase shift between all body parts and the table increases continuously with increasing frequency. At the first resonant frequency (1.5 Hz), the head of the standing subject has a 180° phase shift with respect to the table; between 2 and 3 Hz this phase shift is 360°.

There are longitudinal head motions excited by the transverse vibration in addition to the transverse head motions. The head performs a nodding motion due to the anatomy of the upper vertebrae and the location of the head's center-of-gravity. Above 5 Hz, the head motion for sitting and standing subjects is predominantly vertical (about 10 to 30 percent of the horizontal table motion).

Vibrations Transmitted from the Hand. The mechanical impedance of the hand-arm system measured at a hand grip under conditions representative of those associated with power-tool operation is shown in Fig. 42.12 for vibration directed essentially along the long axis of the forearm, that is, approximately in the direction of thrust. The precise direction is the Z component of the standardized coordinate system for the hand shown in Fig. 42.13, Z_h. Typical values of impedance have again been defined by a synthesis of measured values from different experimental studies, as described previously for whole-body impedance and transmissibility. The most

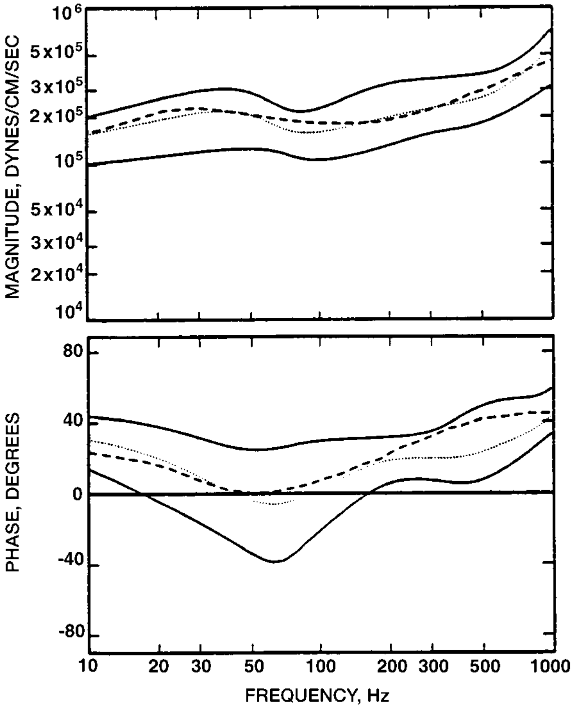

FIGURE 42.12 Mechanical impedance of the hand-arm system, expressed as magnitude and phase, in the Z_h direction specified in Fig. 42.13. Maximum and minimum envelopes of mean values from studies included in the data synthesis are shown by continuous lines, while the mean of these data sets is shown by the dotted line. The response of a 4-degree-of-freedom biodynamic model is shown by the dashed line. (*ISO 10068.*[9])

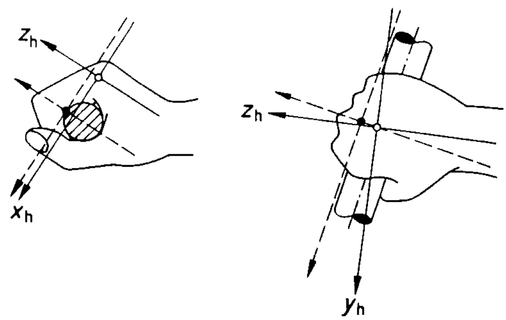

FIGURE 42.13 Standardized biodynamic (open circles—continuous lines) and basicentric (closed circles—dashed lines) coordinate systems for the hand. (*ISO 5349.*[35])

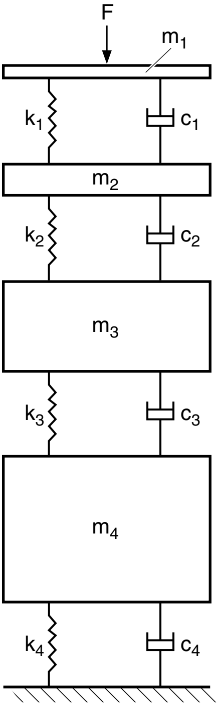

FIGURE 42.14 Biodynamic model for hand-arm impedance. The parameters of this model do not possess direct anatomical correlates.

probable values of impedance magnitude and phase are specified by an upper and lower envelope (the continuous lines in Fig. 42.12) and define a range of idealized impedances.[9] The mean of the data sets is shown by the dotted line. Also shown in the diagram are impedance values calculated by the 4 degree-of-freedom biodynamic model illustrated in Fig. 42.14. Equivalent data are available for the two orthogonal directions of the hand-arm coordinate system not shown in Fig. 42.12 (X_h and Y_h).[9] It should be noted that the parameters of these biodynamic models do not possess direct anatomical correlates, and, together with the idealized impedances, are intended to facilitate the development of devices for reducing vibration transmitted to the hands and of test rigs with which to measure power-tool handle vibration.

The mechanical impedance of the hand-arm system generally increases in magnitude with frequency, with a maximum at a frequency from 20 to 70 Hz. The model values suggest that resonances occur in structures within the hand, resulting in relative motion between tissue layers, and between tissue and the bone. The coupled mass in contact with the handle and subject to the vibration input is typically less than 20 grams. Small increases in impedance magnitude have been observed with increases in grip force (from the value of 25 *N* used for the data synthesis), which leads to increased indentation of the skin (see Fig. 42.3). The influence of the translational force with which the hand presses the handle (i.e., the thrust force) appears to be insignificant at frequencies above 100 Hz and to introduce variations in mechanical impedance magnitude and phase of less than 10 percent at frequencies between 20 and 70 Hz.

For hand tools involving a palm grip, the vibration amplitude decreases from the palm to the back of the hand. Further reductions in amplitude occur from the hand to the elbow and from the elbow to the shoulder.

MIDDLE-FREQUENCY RANGE (WAVE PROPAGATION)

Above about 100 Hz, simple lumped parameter models become more and more unsatisfactory for describing the vibration of tissue. At higher frequencies it is necessary to consider the tissue as a continuous medium for vibration propagation.

Skull Vibrations. The vibration pattern of the skull is approximately the same as that of a spherical elastic shell. The nodal lines observed suggest that the fundamental resonance frequency is between 300 and 400 Hz and that resonances for the higher modes are around 600 and 900 Hz. The observed frequency ratio between the modes for the skull is approximately 1.7, while the theoretical ratio for a sphere is 1.5. From the observed resonances, the calculated value of the elasticity of skull bone (a value of Young's modulus = 1.4×10^{10} dynes/cm^2) agrees reasonably well with static test results on dry skull preparations but is somewhat lower than the static test data obtained on the femur (Table 42.2). Mechanical impedances of small areas on the skull over the mastoid area have been measured to provide information for bone-conduction hearing. The impedance of the skin lining in the auditory canal has been investigated and used in connection with studies on ear protectors.

Vibration of the lower jaw with respect to the skull can be explained by a simple mass-spring system, which has a resonance, relative to the skull, between 100 and 200 Hz.

Mechanical Impedance of Soft Human Tissue. Mechanical impedance measurements of small areas (1 to 17 cm^2) over soft human body tissue have been made with vibrating pistons between 10 Hz and 20 kHz. At low frequencies this impedance is a large elastic reactance. With increasing frequency the reactance decreases, becomes zero at a resonance frequency, and becomes a mass reactance with a further increase in frequency (Fig. 42.15).[10] These data cannot be explained by a simple lumped parameter model, but require a distributed parameter system including a viscoelastic medium—such as the tissue constitutes for this frequency range. The high viscosity of the medium makes possible the use of simplified theoretical assumptions,

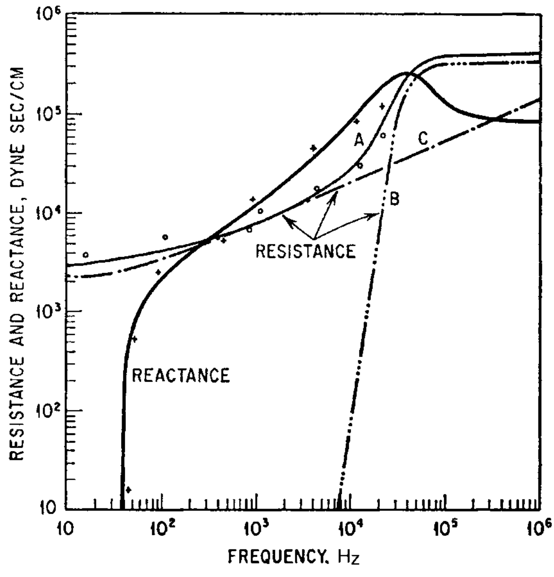

FIGURE 42.15 Resistance and reactance of circular area, 2 cm in diameter, of soft tissue body surface as a function of frequency. Crosses and circles indicate measured values for reactance and resistance. Smooth curves calculated for 2-cm-diameter sphere vibrating in (*A*) viscoelastic medium with properties similar to soft tissue (parameters as in Table 42.2), (*B*) frictionless compressible fluid, and (*C*) incompressible viscous fluid. (*From von Gierke et al.*[10])

such as a homogeneous isotropic infinite medium and a vibrating sphere instead of a circular piston. The results of such a theory agree well with the measured characteristics. As a consequence it is possible to assign absolute values to the shear viscosity and the shear elasticity of soft tissue (Table 42.2). The theory together with the measurements show that, over the audio-frequency range, most of the vibratory energy is propagated through the tissue in the form of transverse shear waves—not in the form of longitudinal compression waves. The velocity of the shear waves is about 20 meters per second at 200 Hz and increases approximately with the square root of the frequency. This may be compared with the constant sound velocity of about 1500 meters per second for compressional waves. Some energy is propagated along the body surface in the form of surface waves which have been observed optically. Their velocity is of the same order as the velocity of shear waves.[10]

Multibody and Finite Element Biodynamic Models.[5,11] In a multibody biodynamic model, the human is represented by a three-dimensional coupled system of rigid bodies and joints. The Articulated Total Body Model (ATBM) represents the human body by, typically, 15 ellipsoidal-shaped segments connected by 14 joints. The body segments possess masses and moments of inertia derived from human data, while the joints possess appropriate mechanical properties. When a representation of the immediate environment, such as a seat and safety harness, is included, the models can predict the gross body motion and forces resulting from external accelerations occurring, for example, during motor vehicle and aircraft crashes, and aircraft seat ejection. Models have also been developed to predict the response of the anthropometric dummies commonly used in motor vehicle testing.

Models that include finite elements (FEs) permit more detailed simulation of the interaction between the occupant of a motor vehicle, or aircraft, and the immediate environment, as well as the detailed response of some body parts.[38] The most sophisticated models include the active and passive behavior of muscles. The MAthematical and DYnamical MOdel (MADYMO) was originally developed for motor vehicle crash simulation, and can include the crash behavior of structural components. Elements that have been modeled with FE structures include the seat, seat frame, and vehicle interior (including padding). While it is at present impractical to construct an FE model for the complete human body, FE models of human body subsystems, such as the head and spine, can generate detailed information on, for example, the mechanical loads at each vertebra along the spinal column.[11]

HIGH-FREQUENCY RANGE

Above several hundred thousand Hz, in the ultrasound range, most of the vibratory energy is propagated through tissue in the form of compressional waves; for these conditions, geometrical acoustics offer a good approximation for the description of their path. Since the tissue dimensions under consideration are almost always large compared to the wavelength (about 1.5 mm at 1 MHz), the mechanical impedance of the tissue is equal to the characteristic acoustic impedance, i.e., sound velocity times density. This value for soft tissue differs only slightly from the characteristic impedance of water. The most important factor in this frequency range is the tissue viscosity, which brings about an increasing energy absorption with increasing frequency.

At very high frequencies this viscosity also generates shear waves at the boundaries of the medium, at the boundary of the acoustic beams, and in the areas of wave transition to media with somewhat different properties (e.g., boundary of muscle to fat tissue, or soft tissue to bone). These shear waves are attenuated so rapidly that they are of no importance for energy transport but are noticeable as increased local absorption, i.e., heating.[11]

EFFECTS OF SHOCK AND OF VIBRATION

The motions and mechanical stresses resulting from the application of mechanical forces to the human body have several possible effects: (1) the motion may interfere directly with physical activity; (2) there may be mechanical damage or destruction; (3) there may be secondary effects (including subjective phenomena) operating through biological receptors and transfer mechanisms, which produce changes in the organism. Thermal and chemical effects are usually unimportant, except in the ultrasonic frequency range.

EFFECTS OF MECHANICAL VIBRATION

Mechanical Damage. Damage is produced when the accelerative forces are of sufficient magnitude. Mice, rats, and cats have been killed by exposure to vibration.[12] There is a definite frequency dependence of the lethal accelerations coincident with resonance displacement of the visceral organs. Mice are killed at accelerations of 10 to 20g within a few minutes in the range 15 to 25 Hz; above and below this frequency range, the survival time is longer. Rats and cats may be killed within 5 to 30 minutes at accelerations above about 10g. Postmortem examination of these animals usually shows lung damage, often heart damage, and occasionally brain injury. The injuries to heart and lungs probably result from the beating of these organs against each other and against the rib cage. The brain injury, which is a superficial hemorrhage, may be due to relative motion of the brain within the skull, to mechanical action involving the blood vessels or sinuses directly, or to secondary mechanical effects. Tearing of intraabdominal membranes rarely occurs.

An increase in body temperature is also observed after exposure to intense vibration. Since this effect also occurs in dead animals, it is probably mechanical in origin. Estimates of energy dissipation from body mechanical impedance data suggest that appreciable heat can be generated at large vibration amplitudes.

In humans, mechanical damage to the heart and lungs, injury to the brain, tearing of membranes in the abdominal and chest cavities, as well as intestinal injury are possible, in principle. However, equinoxious contours of whole-body acceleration as a function of frequency have not been established for any of these phenomena, owing to an almost complete lack of data. Any effects would be expected to occur at lower frequencies than those in animals owing to the increased human visceral masses. Exposure for 15 minutes to an acceleration of 6g has been reported to cause gastrointestinal bleeding that persisted for several days in one subject.[1]

Chronic injuries may be produced by vibration exposure of long duration at levels which produce no acute effects.[1,2] There is epidemiological evidence that occupations with exposure to whole-body vibration are at greater risk of low back pain, sciatic pain, and herniated lumbar disc when compared with control groups not exposed to vibration.[13,14] There is also an increased risk of developing degenerative changes in the spine, including lumbar intervertebral disc disorders, for crane operators, tractor drivers, and drivers in the transportation industry. Nevertheless, it is difficult to differentiate between the relative roles of whole-body vibration and ergonomic risk factors, such as posture and awkward back movements, from epidemiological studies, though both are clearly cofactors in the development of the observed pathology. Exposure to repeated random jolts (in contrast to sinusoidal motion), such as the buffeting that occurs in aircraft, in small craft on rough water, or in off-the-road vehicles is commonly associated with the chronic injuries described.

Chronic injuries may also be produced when the hand is exposed to intense vibration, such as occurs during occupational use of some power tools (e.g., pneumatic drills and hammers, grinders, chain saws, and riveting guns).[3] Symptoms of numbness or paresthesias in the fingers are common and may be accompanied by episodes of finger blanching. Reduced grip strength and muscular weakness may also be experienced. The vascular, nerve, and muscular disorders associated with the use of hand-held vibrating power tools are known as the *hand-arm vibration syndrome* (HAVS). Pathological changes have been observed in the structure of the nerves and walls of the blood vessels in the fingers.[3] Changes in tactile function have been linked to changes in acuity of specific types of mechanoreceptive nerve endings at the fingertips.[15]

Few exposure-response relationships have been derived from epidemiological data for any sign, or symptom, of HAVS resulting from occupational use of hand-held power tools or industrial processes. For groups of workers who perform similar tasks throughout the workday, the *latency,* that is, the duration of exposure (in years) prior to the onset of episodes of finger blanching, and prevalence, may be predicted from the acceleration of a surface in contact with the hand.[16] These relationships serve as the basis for occupational exposure criteria (see *Human Tolerance Criteria*).

The tendons, tendon sheaths, muscles, ligaments, joints, and nerves in the hand and arm can also be damaged by repeated movement of the hand relative to the arm. These soft tissue and nerve injuries occur among blue- and white-collar workers performing tasks involving repeated hand-wrist flexure (e.g., keyboard operators) and are termed *repetitive strain injuries* (RSI).[17] Nerve compression may result from changes in the contents of restricted nerve passageways (e.g., the carpal tunnel at the wrist—*carpal tunnel syndrome*).[3] Pain and paresthesias in the hand and arm are common symptoms.

Physiological Responses. Vibration can induce physiological responses in the cardiovascular, respiratory, skeletal, endocrine, and metabolic systems and in muscles and nerves. The cardiovascular changes in response to intense vertical vibration are similar to those accompanying moderate exercise: increased heart rate, respiration rate, cardiac output, and blood pressure. Vibration of sufficient intensity will cause mechanical pumping of the respiratory system, as already noted, but is unlikely to produce significantly increased ventilation or oxygen uptake. Changes in blood and urine constituents are commonly used as indicators of generalized body stress and may, in consequence, be observed in persons exposed to vibration. It is difficult if not impossible, however, to relate specific endocrine and metabolic responses to a given vibration stimulus. Vibration can stimulate a tonic reflex contraction in muscles, which is a response to the stretching force (the *tonic vibration reflex*), disturb postural stability, and lead to body sway. Extremely low-frequency whole-body vibration, such as occurs in many transportation vehicles and ships, may also cause motion sickness (*kinetosis*).[1]

Vibration of the hand may cause peripheral vascular, neurological, and muscular responses.[3] Blood flow within the fingers may be reduced during stimulation, and tingling and paresthesias in the hands may be reported after exposure. Somatosensory perception and tactile function may be temporarily decreased. Grip strength may also be affected. Extremely low-frequency, large-amplitude motions, which are usually described as repetitive movements of the hand (and frequently involve repeated wrist rotation), may lead to tendon and muscle fatigue and to transitory parathesias or numbness.

Therapeutic applications of vibration include cardiac and circulatory assist devices and the control of spastic muscle. Ultrasonic frequencies are used in medical diagnosis, for soft tissue visualization, and for therapy. A common therapeutic use is to promote the return of limb function in rehabilitation medicine.

Subjective Responses. Feelings of discomfort and apprehension may be associated with exposure to whole-body and hand-arm vibration once the stimulus has been perceived. The extent of the discomfort depends on the magnitude, frequency, direction and duration of the exposure, and the posture and orientation of the body, as well as the point of contact with the stimulus. The response is also influenced by the environment in which the motion is experienced (e.g., floor motion in hospital versus aircraft). The range in response of different individuals to a given stimulus is large. In some circumstances, whole-body vibration may be exhilarating (e.g., a fairground ride) or soothing (e.g., rocking a baby in a cradle or a rocking chair).

In general, subjective responses to vibration may be subdivided into three broad categories: the threshold of perception, the onset of unpleasant sensations, and the limit of tolerance. The specification of acceptable vibration environments is discussed later in this chapter.

Once detected, the growth in sensation follows a Stevens' power law function with index k, in which the psychophysical magnitude of a stimulus, ψ, is related to its physical magnitude ϕ by

$$\psi = \text{constant}\{\phi^k\} \tag{42.1}$$

For discomfort associated with whole-body vibration, $k \approx 1$. Frequency contours of equal sensation magnitude depend principally on the direction in which vibration enters the body and whether the person is standing, seated, or recumbent.[1] Contours which summarize current knowledge may be inferred from the frequency-weighting functions employed in the international standard for whole-body vibration (i.e., by reciprocal curves to those shown later in Fig. 42.23). The effect of the duration of exposure t on subjective responses to suprathreshold vibration is often found to follow a power law relationship of the form

$$\phi^n t = \text{constant} \tag{42.2}$$

where the magnitude of the index n is from 2 to 4. For situations in which the *perception* of vibration is judged unacceptable, the boundary between acceptable and unacceptable exposures will be related to the physical magnitude of the stimulus corresponding to the threshold of perception, and will not depend on the duration of exposure. There is an extensive literature discussing the comfort/discomfort of passengers in road and rail vehicles, aircraft, and ships.[1,32]

The results of an experiment to establish subjective limits of tolerance to vertical vibration for short-duration exposures (less than 5 minutes) is shown in Fig. 42.16. The peak accelerations at which 10 subjects refused to continue exposure can be seen to depend on frequency, and describes an equi-noxious contour for these stimuli and experimental conditions. Subjects reported as the reason for discontinuing exposure either general discomfort or, within restricted frequency ranges, difficulty breathing (1 to 4 Hz) or chest and/or abdominal pain (3 to 9 Hz).

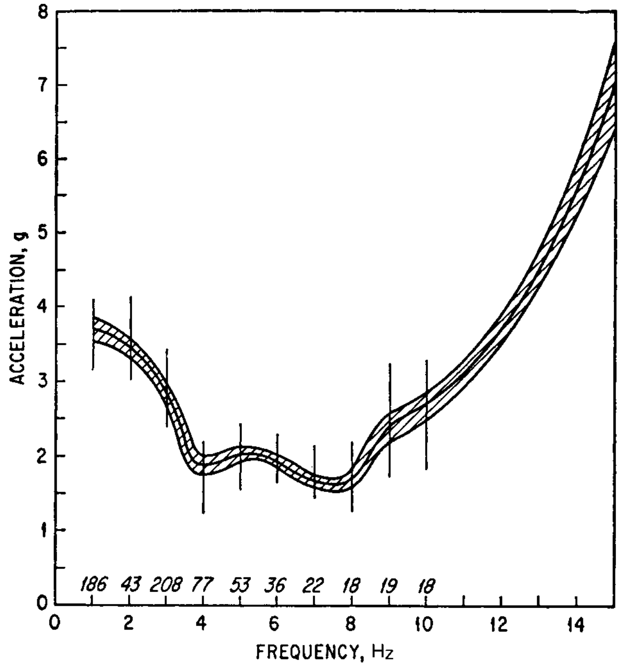

FIGURE 42.16 Peak acceleration at various frequencies at which subjects refuse to tolerate further a short exposure (less than 5 min) to vertical vibration. The figures above the abscissa indicate the exposure time in seconds at the corresponding frequency. The shaded area has a width of one standard deviation on either side of the mean (10 subjects). (*Ziegenruecker and Magid:* USAF WADC Tech. Rept. 59–18, *1959.*)

EFFECTS OF MECHANICAL SHOCK

Mechanical shock includes several types of force application which have similar, though not identical, effects. Explosions, explosive compression or decompression, and impacts and blows from rapid changes in body velocity or from moving objects produce shock forces of importance. Major damage, short of complete tissue destruction, is usually to lungs, intestines, heart, head, neck, or brain. Differences in injury patterns arise from differences in rates of loading, peak force, duration, and localization of forces.

Blast and Shock Waves.[18] The mechanical effects associated with rapid changes in environmental pressure are primarily localized to the vicinity of air-filled cavities in the body, i.e., the ears, lungs, and air-containing gastrointestinal tract. Here, heavy masses of blood or tissue border on light masses of air. The local impedance mismatch can lead to a relative tissue displacement, which is destructive, by several different mechanisms.

The ear is the part of the human body most sensitive to blast injury. Rupture of the tympanic membrane and injury to the conduction apparatus can occur singly or together with injury to the hair cells in the inner ear. The two first-mentioned injuries may protect the inner ear through energy dissipation. The degree of injury depends on the frequency content of the blast pressure function. The fact that the ear's greatest mechanical sensitivity occurs at frequencies between 1500 and 3000 Hz explains its vulnerability to short-duration blast waves. Peak pressures of only a few pounds per square inch can rupture the ear drum, and still smaller pressures can damage the conducting mechanism and the inner ear. There are wide variations in individual susceptibility to these injuries.

With very slow differential pressure changes, of approximately 1 sec duration or longer, dynamic mechanical effects are unimportant; the static pressure is responsible for destructive mechanical stress or physiological response. Such pressure-time functions occur with the explosive decompression of pressurized aircraft cabins at high altitude and with the slow response of well-sealed shelters to blast waves. If the pressure rise times or fall times are shortened (roughly to the order of tenths of seconds), the dynamic response of the different resonating systems of the body becomes important, in particular the thorax-abdomen system of Fig. 42.11. Available data for single pulse, "instantaneously rising" pressures suggest the existence of a minimum peak pressure which corresponds to natural frequencies for dogs of between 10 and 25 Hz; for humans this frequency is lower. Contours of equal injury potential are shown in Fig. 42.17 for various species. The theoretical curves are obtained by means of the thorax model of Fig. 42.11 after application of appropriate scaling laws to account for the different species sizes.[7] For pressures with total durations of milliseconds or less and much shorter rise times (duration of wave short compared to the natural period of the responding tissue), the effect and destruction seem to depend primarily on the momentum of the shock wave. The mass m of an oscillatory system located in a wall or body surface, which is struck by a shock wave, is set into motion according to the relation

$$P_r\,dt = mv_0 \tag{42.3}$$

where P_r = reflected pressure at body surface
t = time
v_0 = initial velocity

Experimental fatality curves on animals generally show this dependence on momentum for short pressure phenomena (close to center of detonation) and the transition to a dependence on peak pressure for phenomena of long duration (far away from center).[18] Fatal blast waves in air and water, for example, differ widely in peak pressure and duration (in air, 10 atm in excess of atmospheric pressure with a duration of 2.8 milliseconds, in water 135 atm in excess with a duration of 0.17 milliseconds), but their momenta are similar. In this most important range of short-duration blasts, the mechanical effects are localized because of the short duration, i.e., the high-frequency content of the wave. The upper respiratory tract and bronchial tree, as well as the thorax and abdomen system, are too large and have resonance frequencies too low to be excited; there is no general compression or overexpansion of the thorax, which leads to pulmonary injury as in explosive decompression. The blast waves go directly through the thoracic wall, producing an impact or grazing blow. Inside the tissue, blast injury has three possible causes: (1) spalling effects, i.e., injuries caused by the tensile stresses arising from the reflection of the shock wave at the boundary between media with different propagation velocity [for example, subpleural pulmonary hemorrhages along the ribs]; (2) inertia effects which lead to

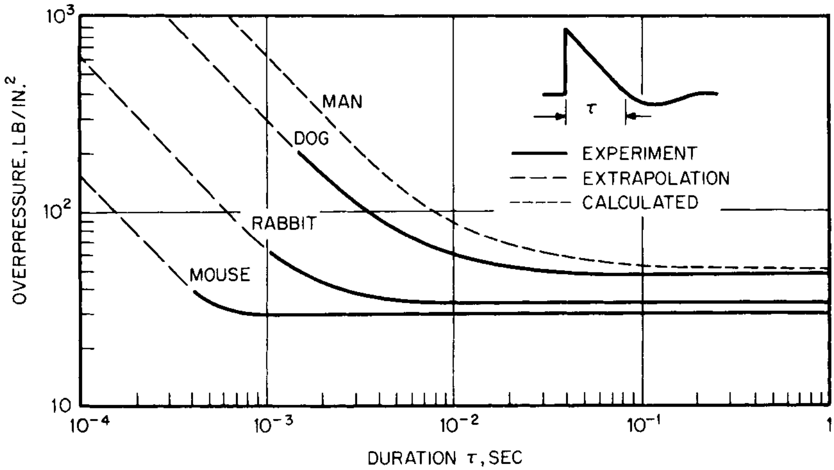

FIGURE 42.17 Estimated maximum tolerable blast overpressures for mouse, rabbit, dog, and man. The curve for man is calculated by means of a model of the type given in Fig. 42.11. Using the same model, dimensionally scaled to the animal sizes, results in curves matching closely the experimental data.

different accelerations of adjacent tissues with various densities, when the shock wave passes simultaneously through these media; and (3) implosion of gas bubbles enclosed in a liquid. These phenomena are compatible with observations made when high-velocity missiles pass through water near air-containing tissues.[19] The shock waves may produce not only pulmonary injuries but also hard, sharply circumscribed blows to the heart.

Of the injuries produced by exposure to high-explosive blast, lung hemorrhage is one of the most common. It may not of itself be fatal, since enough functional lung tissue may easily remain to permit marginal gas exchange. However, the rupture of the capillaries in the lung produces bleeding into the alveoli and tissue spaces, which can seriously hamper respiratory activity or produce various respiratory and cardiac reflexes. The heart rate is often very slow after a blast injury. Leakage of fluid through moderately injured, but not ruptured, capillaries may occur. There is also the possibility that air may enter the circulation to form bubbles or emboli and by reaching critical regions may fatally impair the heart or brain circulation or produce secondary damage to other organs.[18] Fat emboli also may be formed and these, too, are capable of blocking vessels supplying vital parts. When gas pockets are present in the intestines, the shock may produce hemorrhage and in extreme cases rupture the intestinal wall itself.

The effects of underwater shock waves on man and animals are in general of the same kind as those produced by air blast. Differences which appear are those of magnitude and often depend on the mode of exposure of the body. A person in the water may, for example, be submerged from the waist down only; in this case, damage is practically confined to the lower half of the body so that intestinal, rather than lung, damage will occur. Direct mechanical injury to the heart muscle and conducting mechanism is possible.

Cerebral concussion resulting directly from exposure to shock waves is unusual. Neurological symptoms following exposure to blast, however, may include general depression of nervous activity sometimes to the point of abolition of certain reflexes. Psychological changes such as memory disturbances and abnormal emotional states are found sometimes. In extreme cases, there may be paralysis or muscular dysfunction. Unconsciousness and subsequent amnesia for events immediately preceding the injury result more commonly from blows to the head than from air blast. Recovery from minor concussion apparently may be complete, but repeated concussion may produce lasting damage.

Impacts, Blows, Rapid Deceleration. This type of force is experienced in falls, in motor vehicle or aircraft crashes, in parachute openings, in seat ejections for escape from high-speed military aircraft, and in many other situations. Interest in the body's responses to these forces centers on mechanical stress limits. Accident statistics from the United States (from 1979 to 1986) indicate that serious injuries to occupants of automobiles involved in frontal impacts, and who were wearing seat belts, were most commonly to the head (approximately 35 percent), followed by the thorax (including abdomen), and lower extremities (approximately 25 percent each). The distribution of injuries in fatal accidents involving military helicopters and pilot ejections from fixed-wing aircraft is similar to that of the automobile statistics cited with, in addition, injuries to the spine in approximately 13 percent of cases.[5] For crewmen who survived seat ejection from military aircraft, the most common injury was to the spine, while for passengers surviving civil air transport accidents the most common injury remained to the head.[4]

Serious injuries to the head usually involve brain injury, either with or, commonly, without skull fracture. The brain may suffer either diffuse or focal injuries. The former consists of brain swelling, concussion, and *diffuse axonal injury,* that is, mechanical disruption of the nerve fibers; the latter consists of localized internal bleeding and contusions (coup and contrecoup). Concussion is the most common brain injury. Skull motion and fracture have been extensively investigated (see later in this section), and have led to criteria for head injury (see *Human Tolerance Criteria*).

The most common neck and spinal injury is caused by rearward flexion and forward extension of the neck, such as commonly occurs in rear-end motor vehicle collisions ("whiplash"), and results in localized pain in the neck and shoulders, and even cord injuries. The motion can also result in dislocation or fracture of the first and second vertebral joints, and may lead to the spinal cord being crushed or severed. Both neck and spine may be injured by vertical accelerations directed from the head or buttocks, leading to dislocation and fracture with, again, the potential for spinal cord injury. The nature and degree of injury is critically dependent on the body position at impact.

The chest encloses important organs—the heart, lungs, trachea, esophagus, and major blood vessels—and so injuries may be divided into those affecting the organs, and those affecting the rib cage. Injuries to the internal organs include ruptures of the heart, the lung, and of the arteries connected to the heart, while injuries to the rib cage involve fractures of the ribs and sternum, and sometimes dislocations and fractures of the thoracic vertebrae. Compound rib fractures may, if sufficiently displaced, also result in puncturing of internal organs. Organs within the abdomen (especially liver, kidneys, and spleen) are also subject to injury by external trauma involving transverse (e.g., front-to-back or side-to-side) accelerations.

Common injuries to the lower extremities involve fractures of the long bones and injuries to the joints.

Force Duration. The correlation between the response of the body system to continuous vibration and to spike and step-force functions may be used to guide and interpret exposures. The tissue areas stressed to maximum relative displacement at the various frequencies during steady-state excitation are preferred target areas for injury under impact load if the force-time functions of the impacts have appreciable energy in these frequency bands. If the impact exposure times are shorter, stress tolerance limits increase; if exposure times decrease to hundredths or thousandths of a second, the response becomes more and more limited and localized to the point of application of the force (blow). Elastic compression or injury will depend on the load distribution over the application area, i.e., the pressure, to which tissues are subjected. If tissue destruction or bone fracture occurs close to the area of application of the force, these will absorb additional energy and protect deeper-seated tissues by reducing the peak force and spreading it over a longer period of time. An example is the fracture of foot and ankle of men standing on the deck of warships when an explosion occurs beneath. The support may be thrown upward with great momentum; if the velocity reaches 5 to 10 ft/sec (which corresponds, under these conditions, to an acceleration of several hundred g) fractures occur.[20] However, the energy absorption by the fracture protects structures of the body which are higher up.

If the force functions contain extremely high frequencies, the compression effects spread from the area of force application throughout the body as compression waves. If these are of sufficient amplitude, they may cause considerable tissue disruption. Such compression waves are observed from the impact of high-velocity missiles.

If the exposure to the accelerating forces lasts long enough so that (as in most applications of interest) the whole body is displaced, exact measurement of the force applied to the body and of the direction and contact areas of application becomes of extreme importance. In studies of seat ejection, for example, a knowledge of seat acceleration alone is not sufficient for estimating responses. One must know the forces in those structures or restraining harnesses through which acceleration forces are transmitted. The location of the center-of-gravity of the various body parts such as arms, head, and upper torso must be known over the time of force application so that the resulting body motion and deformation can be analyzed and controlled for protection purposes. In addition to the primary displacements of body parts and organs, there are secondary forces from decelerations if, due to the large amplitudes, the motions of parts of the body are stopped suddenly by hitting other body parts. Examples occur in linear deceleration where, depending on the restraint, the head may be thrown forward until it hits the chest or, if only a lap belt is used, the upper torso may jackknife and the chest may hit the knees. There is always the additional possibility that the body may strike nearby objects (e.g., automobile dashboard or door post), thus initiating a new impact deceleration history.

Longitudinal Acceleration. The study of positive longitudinal (headward) acceleration of short duration is connected closely with the development of upward ejection seats for escape from aircraft. Since the necessary ejection velocity of approximately 60 ft/sec and the available distance for the catapult guide rails of about 3 ft are determined by the aircraft, the minimum acceleration required (step function) is approximately 18.6g. Since the high jolt of the instantaneous acceleration increase is undesirable because of the high dynamic load factor in this direction for the frequency range of body resonances, slower build-up of the acceleration with higher final acceleration is preferable to prevent injury. Investigations show that the body's ballistic response can be predicted by means of analog computations making use of the frequency-response characteristics of the body. The simplest analog used for the study of headward accelerations is the single degree-of-freedom mechanical resonator composed of the lumped-parameter elements of a spring, mass, and damper. A diagram of this model is shown in Fig. 42.18*A*. The model is used to simulate the maximum stress developed within the vertebral column (the first failure mode in this direction) for any given impact environment. The maximum dynamic deflection of

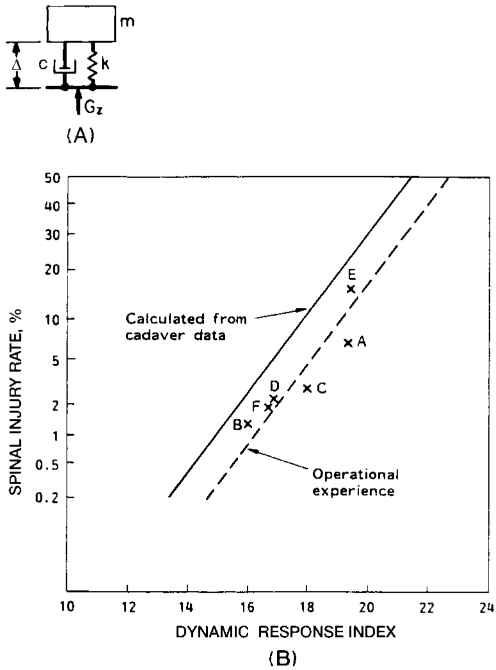

FIGURE 42.18 Prediction of spinal compression injury from pilot ejection seat accelerations. (*A*) Model for the study of spinal compression Δ with mass *m,* spring stiffness *k* and damping *c;* (*B*) relation between the dynamic response index (DRI) and spinal injury rate for 361 nonfatal ejections from six different types of aircraft (dashed line) (aircraft type A, 64 ejections; B, 62; C, 65; D, 89; E, 33; F, 48). Data from cadavers (continuous line). (*After Griffin,*[1] *and von Gierke.*[6])

the spring, Δ_{max}, may be calculated for a given input acceleration-time history to the model. The potential for spinal injury is estimated by forming the *dynamic response index* (DRI), which is defined as $\omega_n^2\Delta_{max}/g$, where the natural frequency of the model $\omega_n = (k/m)^{1/2}$ is 52.9 rad/sec and the damping ratio $c/2(km)^{1/2}$ is 0.224. Experience with nonfatal ejections from military aircraft, shown by the crosses and dashed line in Fig. 42.18*B,* suggests a 5 percent probability of spinal injury from exposure to a dynamic response index of 18. An estimate of the rate of spinal injury from cadavers is shown in this diagram by the continuous line. The success of the model has led to its adoption for the specification of ejection seat performance, for its extension to accelerations in three orthogonal directions,[21] and to measures of ride comfort for exposure to repeated impacts in some land vehicles and high-speed boats.[22]

For negative (tailward) acceleration (downward ejection) no firm point for application of the accelerating force is accessible as for positive acceleration. If the force is applied as usual through harness and belt at shoulder and groin, the mobility of the shoulder girdle together with the elasticity of the belts results in a lower resonance frequency than the one observed in upward ejection. To avoid overshooting with standard harnesses, the acceleration rise time must be at least 0.15 sec. This type of impact can excite the thorax-abdomen system (Fig. 42.11). The diaphragm is pushed upward by the abdominal viscera; as a result, air rushes out of the lungs (if the glottis is open) or high pressures develop in the air passages.

Transverse Accelerations. The forward- and backward-facing seated positions are most frequently exposed to high transverse (i.e., horizontal) components of crash loads. Human tolerance to these forces has been studied extensively by volunteer tests on linear decelerators, in automobile crashes, and by the analysis of the records of accidental falls. The results indicate the importance of distributing the decelerative forces or impact over as wide an area as possible. The tolerable acceleration amplitudes of well over 50g (100g and over for falling flat on the back with minor injuries, 35 to 40g for 0.05-sec voluntary deceleration when seated with restraining harness) are probably limited by injury to the brain. An indication that the latter might be sensitive to and based on specific dynamic responses is the fact that the tolerance limit depends strongly on the rise time of the acceleration. With rise times around 0.1 sec (rate of change of acceleration 500g/sec), no overshooting of head and chest accelerations is observed, whereas faster rise times of around 0.03 sec (1000 to 1400g/sec) result in overshooting of chest accelerations of 30 percent (acceleration front to back) and even up to 70 percent (acceleration back to front). All these results depend critically on the harness for fixation and the back support used (see *Protection Methods and Procedures*). These dynamic load factors indicate a natural frequency of the body system between 10 and 20 Hz. Impact of the heart against the chest wall is another possible injury discussed and noted in some animal experiments.

The head and neck supporting structures seem to be relatively tough. Injury seems to occur only upon backward flexion and extension of the neck ("whiplash") when the body is accelerated from back to front without head support, as already noted.

Head Impact.[23] The reaction of the head to a blow is a function of the velocity, duration, area of impact, and the transfer of momentum. Near the point of application of the blow there will be an indentation of the skull. This results in shear strains in the brain in a superficial region close to the dent. Compression waves emanate from this area, which have normally small amplitudes since the brain is nearly incompressible. In addition to the forces on the brain resulting from skull deformation there are acceleration forces, which also would act on a completely undeformable skull. The centrifugal forces and linear accelerations producing compressional strains are negligible compared to the shear strains produced by the rotational accelerations. The maximum strains are concentrated at regions where the skull has a good grip on the brain owing to inwardly projecting ridges, especially at the wing of the sphenoid bone of the skull. Shear strains also must be present throughout the brain and in the brain stem. Many investigators consider these shear strains, resulting from rotational accelerations due to a blow to the unsupported head, as the principal event leading to concussion. Blows to the supported, fixed head are supposed to produce concussion by compression of the skull and elevation of cerebrospinal fluid pressure. There is now general acceptance that rotational acceleration can be one of the main causes of concussion. This follows from animal experiments in which typical primate brain injuries were reproduced, ranging from subconcussive injury with little histological evidence of axonal damage, to prolonged traumatic coma (lasting hours or days) with extensive axonal injury throughout the white matter and brainstem, to immediate fatal injury.[4]

In general a high-velocity projectile (for example, a bullet of 10 grams with a speed of 1000 ft/sec) with its high kinetic energy and low momentum produces plainly visible injury to scalp, skull, and brain along its path. The high-frequency content of the impact is apt to produce compression waves which in the case of very high energies may conceivably lead to cavitation with resulting disruption of tissue. Skull fracture is not a prerequisite for these compression waves. However, if the head hits a wall or another object whose mass is large compared to the head's mass, the local, visible damage is small and the damage due to rotational acceleration may be large. Blows to certain points, especially on the midline, produce no rotation. Blows to the chin upward and sideward produce rotation relatively easily ("knock-out" in boxing). Velocities listed in the literature for concussion from impact of large masses range from 15 to 50 ft/sec. At impact velocities of about 30 ft/sec, approximately 200 in.-lb of energy is absorbed in 0.002 sec, resulting in an acceleration of the head of 47g. Impact energies for compression concussion are probably approximately in the same range.

Scalp, skin, and subcutaneous tissue reduce the energy applied to the bone. If the response of the skull to a blow exceeds the elastic deformation limit, skull fracture occurs. Impact by a high-velocity, blunt-shaped object results in localized circumscribed fracture and depression. Low-velocity blunt blows, insufficient to cause depression, occur frequently in falls and crashes. Given enough energy, two, three, or more cracks appear, all radiating from the center of the blow. The skull has both weak and strong areas, each impact area showing well-defined regions for the occurrence of the fracture lines.

The total energy required for skull fracture varies from 400 to 900 in.-lb, with an average often assumed to be 600 in.-lb. This energy is equivalent to the condition that the head hits a hard, flat surface after a free fall from a 5-ft height. Skull fractures occurring when a batter is accidentally hit by a ball (5 oz) of high velocity (100 ft/sec) indicate that about the same energy (580 in.-lb) is required. Additional energy 10 to 20 percent beyond the single linear fracture demolishes the skull completely. Dry skull preparations required only approximately 25 in.-lb for fracture. The reason for the large energy difference required in the two conditions is attributed mainly to the attenuating properties of the scalp.

In automobile and aircraft crashes the form, elasticity, and plasticity of the object injuring the head is of extreme importance and determines its "head injury potential." For example, impact with a 90° sharp corner requires only a tenth of the energy for skull fracture (60 in.-lb) that impact with a hard, flat surface requires.

EFFECTS OF SHOCK AND VIBRATION ON TASK PERFORMANCE

The performance of tasks requiring a physical response to some stimulus involves peripheral (e.g., perceptual and motor) and central neurological processes, with multiple feedback paths characteristic of a sophisticated control system. Each of these processes is complex, is more or less developed in different individuals, and may be influenced by training and the general state of health. In consequence, unique relationships between vibration and task performance are unlikely, except for well-defined situations in which some part of the body reaches a physical or physiological limit to performance. For example, movement of images on the retina may cause defocusing and a reduction of visual acuity. The movement may be caused by vibration of the display (i.e., the source), the head (and/or observer), or both. At frequencies below approximately 1 Hz, a pursuit reflex assists visual acuity. At frequencies above 20 Hz an eyeball resonance can degrade acuity. The effects of whole-body vibration on visual acuity therefore depend on the frequency and amplitude, as well as the viewing distance.[1] As already discussed, whole-body vibration can affect speech.

Vibration may also degrade the manual control of objects. The influence of whole-body vibration on writing and drinking is a common experience in public transportation vehicles and ships. Vibration may interfere with the performance of manually controlled systems. The extent of the effect depends on hand motion, the type of control (e.g., a "stiff" control that responds to the application of force without moving or one that moves and responds with little force applied), and the dynamics of the control and the controlled system. A control that responds to hand displacement may be disrupted by vertical vibration at frequencies between 2 and 6 Hz. The effect of the duration of vibration exposure on task performance is influenced by motivation, arousal, and adaptation and may therefore be observed to improve or degrade performance over time.

Exposure of the hand to vibration can lead to sensorineural dysfunction sufficient to reduce the ability to perform fine manual tasks, such as buttoning clothing.[3]

The motion associated with a shock is unlikely to interfere directly with the performance of most tasks unless it is coincident with some critical component of the task. This condition may occur with shocks repeated at very short intervals.

PROTECTION METHODS AND PROCEDURES

Protection of man against mechanical forces is accomplished in two ways: (1) isolation to reduce transmission of the forces to the man and (2) increase of man's mechanical resistance to the forces. Isolation against shock and vibration is achieved if the natural frequency of the system to be isolated is lower than the exciting frequency at least by a factor of 2. Both linear and nonlinear resistive elements are used for damping the transmission system; irreversible resistive elements or energy-absorbing devices can be used once to change the time and amplitude pattern of impulsive forces (e.g., progressive collapse of automobile engine compartment in frontal crash). Human tolerance to mechanical forces is strongly influenced by selecting the proper body position with respect to the direction of forces to be expected. Man's resistance to mechanical forces also can be increased by proper distribution of the forces so that relative displacement of parts of the body is avoided as much as possible. This may be achieved by supporting the body over as wide an area as possible, preferably loading bony regions and thus making use of the rigidity available in the skeleton. Reinforcement of the skeleton is an important feature of seats designed to protect against crash loads. The flexibility of the body is reduced by fixation to the rigid seat structure. The mobility of various parts of the body, e.g., the abdominal mass, can be reduced by properly designed belts and suits. The factor of training and indoctrination is essential for the best use of protective equipment, for aligning the body in the least dangerous positions during intense vibration or crash exposure, and possibly for improving operator performance during vibration exposure.

PROTECTION AGAINST VIBRATIONS

The transmission of vibration from a vehicle or platform to a man is reduced by mounting him on a spring or similar isolation device, such as an elastic cushion. The degree of vibration isolation theoretically possible is limited, in the important resonance frequency range of the sitting man, by the fact that large static deflections of the man with the seat or into the seat cushion are undesirable. Large relative movements between operator and vehicle controls interfere in many situations with man's performance. Therefore a compromise must be made. Cushions are used primarily for static comfort, but they are also effective in decreasing the transmission of vibration above man's resonance range. They are ineffective in the resonance range and may even amplify the vibration in the subresonance range. In order to achieve effective isolation over the 2- to 5-Hz range, the natural frequency of the man-cushion system should be reduced to 1 Hz, i.e., the natural frequency should be small compared with the forcing frequency (see Chap. 30). This would require a static cushion deflection of 10 in. If a seat cushion without a back cushion is used (as is common in some tractor or vehicle arrangements), a condition known as "back scrub" (a backache) may result. Efforts of the operator to wedge himself between the controls and the back of the seat often tend to accentuate this uncomfortable condition.

For severe low-frequency vibration, such as occur in tractors and other field equipment, suspension of the whole seat is superior to the simple seat cushion. Hydraulic shock absorbers, rubber torsion bars, coil springs, and leaf springs all have been successfully used for suspension seats. A seat that is guided so that it can move only in a linear direction seems to be more comfortable than one in which the seat simply pivots around a center of rotation. The latter situation produces an uncomfortable and fatiguing pitching motion. Suspension seats can be built which are capable of preloading for the operator's weight so as to maintain the static position of the seat and the natural frequency of the system at the desired value. Suspension seats for use on tractors and on similar vehicles are available which reduce the resonance frequency of the man-seat system from approximately 4 to 2 Hz. This can be seen from the comparison of the transmissibility of a rigid seat, a truck suspension seat, and a conventional foam and metal sprung car seat in Fig. 42.19. The transmissibility

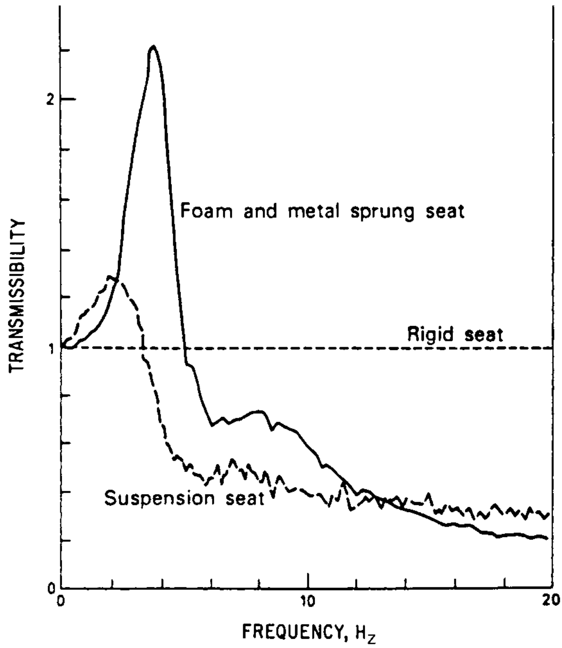

FIGURE 42.19 Comparison of the transmissibilities of a rigid seat, a foam-covered metal sprung seat, and a truck suspension seat. (*Griffin.*[1])

of the car seat is in excess of 2 at the resonance frequency (4 Hz), implying that the seat motion reaching the body is amplified by this ratio. In contrast, the amplification introduced by the suspension seat is at most a factor of 1.3 at the resonance frequency (2 Hz), and improved attenuation of vibration is obtained throughout the frequency range from 4 to 12 Hz. At frequencies below 2 Hz and above 12 Hz, less vibration is transmitted to the subject by the foam and metal sprung seat. There are large differences in the performance of suspension seats, with transmissibilities in excess of 2 being recorded in some designs at the resonance frequency (which is usually close to 2 Hz).[1] In consequence, the selection of a seat for a particular application must take into account both the performance of the seat and the critical seat vibration frequencies to be attenuated.

For severe vibrations, close to or exceeding normal tolerance limits, such as those which may occur in military operations, special seats and restraints can be employed to provide maximum body support for the subject in all critical directions. In general, under these conditions, seat and restraint requirements are the same for vibration and rapidly applied accelerations (discussed in the next section). Laboratory experiments show that protection can be achieved by the use of rigid or semirigid body enclosures. Immersion of the operator in a rigid, water-filled container with proper breathing provisions has been used in laboratory experiments to protect subjects against large, sustained static g loads. This principle can be used to provide protection against large alternating loads.

Isolation of the hand and arm from the vibration of hand-held or hand-guided power tools is accomplished in several ways. A common method is to isolate the handles from the rest of the power tool, using springs and dampers (see Chap. 30). The application of vibration-isolation systems to chain saws for occupational use in forestry has become commonplace and has led to a reduction in the incidence of HAVS. A second method is to modify the tool so that the primary vibration is counterbalanced by an equal and opposite vibration source. This method takes many different forms, depending on the operating principle of the power tool.[24] An example is shown for a pneumatic scaling chisel in Fig. 42.20, in which an axial impact is applied to a work piece to remove metal by a chisel *P*. The chisel is driven into the work piece by compressed air and is returned to its initial position by a spring *S*. The axial motion of the chisel is counterbalanced by a second mass *m* and spring *k* which oscillate out of phase with the chisel motion. The design of an appropriate vibration-isolation system must include the dynamic properties of the hand-arm system. The model of Fig. 42.14 is suitable for this purpose.

Conventional gloves do not attenuate the vibration transmitted to the hand but may increase comfort and keep the hands warm. So-called antivibration gloves also fail to reduce vibration at frequencies below 100 Hz, which are most commonly responsible for HAVS, but may reduce vibration at high frequencies (the relative

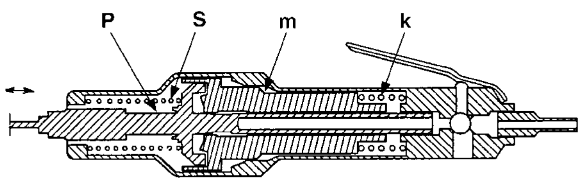

FIGURE 42.20 An antivibration power tool design for a pneumatic scaler: *P*—vibrating chisel; *S*—chisel return spring; *m*—counterbalancing oscillating weight; and *k*—counterbalance return spring. (*Lindqvist.*[24])

importance of different frequencies in causing the vascular component of HAVS is shown in Fig. 42.26).

Preventive measures for HAVS to be applied in the work place include: minimizing the duration of exposure to vibration, using minimum hand-grip force consistent with safe operation of the power tool or process ("let the tool do the work"), wearing sufficient clothing to keep warm, and maintaining the tool in good working order, with minimum vibration. As recovery from HAVS has only been demonstrated for early vascular symptoms, medical monitoring of persons exposed to vibration is essential. Monitoring should include a test of peripheral neurological function,[25] since this component of HAVS appears to persist.

PROTECTION AGAINST RAPIDLY APPLIED ACCELERATIONS (CRASH)

The study of automobile and aircraft crashes and of experiments with dummies and live subjects shows that complete body support and restraint of the extremities provide maximum protection against accelerating forces and give the best chance for survival. If the subject is restrained in the seat, he makes full use of the force moderation provided by the collapse of the vehicle structure, and is protected against shifts in position which would injure him by bringing him in contact with interior surfaces of the cabin structure. The decelerative load must be distributed over as wide a body area as possible to avoid force concentration with resulting bending moments and shearing effects. The load should be transmitted as directly as possible to the skeleton, preferably directly to the pelvic structure—not via the vertebral column.

Theoretically, a rigid envelope around the body will protect it to the maximum possible extent by preventing deformation. A body restrained to a rigid seat approximates such a condition; proper restraints against longitudinal acceleration shift part of the load of the shoulder girdle and arms from the spinal column to the back rest. Arm rests can remove the load of the arms from the shoulders. Semirigid and elastic abdominal supports provide some protection against large abdominal displacements. The effectiveness of this principle has been shown by animal experiments and by impedance measurements on human subjects (see Fig. 42.6). Animals immersed in water, which distributes the load applied to the rigid container evenly over the body surface, or in rigid casts are able to survive acceleration loads many times their normal tolerance.

Many attempts have been made to incorporate energy-absorptive devices, either in a harness or in a seat, with the intent to change the acceleration-time pattern by limiting peak accelerations. For example, consider an aircraft which is stopped in a crash from 100 mph in 5 ft; it is subjected to a constant deceleration of 67g. An energy-absorptive device designed to elongate at 17g would require a displacement of 19 in. In traveling through this distance, the body or seat would be decelerated relative to the aircraft by 14.4g and would have a maximum velocity of 36.8 ft/sec relative to the aircraft structure. A head striking a solid surface (e.g., cabin interior surface) with this velocity has many times the minimum energy required to fracture a skull. The available space for seat or passenger travel using the principle of energy absorption therefore must be considered carefully in the design. Seats for jet airliners have been designed which have energy-absorptive mechanisms in the form of extendable rear-legs. The maximum travel of the seats is 6 in.; their motion is designed to start between 9 and 12g horizontal load, depending on the floor strength. During motion, the legs pivot at the floor level—a feature considered to be beneficial if the floor wrinkles in the crash. Theoretically, such a seat can be exposed to a deceleration of 30g for 0.037 sec or 20g for 0.067 sec without transmitting a deceleration of more than $9g$ to the seat. However, the increase in exposure time must be considered as well as the reduction in peak acceleration. For very short exposure times where the body's tolerance probably is limited by the transferred momentum and not the peak acceleration, the benefits derived from reducing peak loads would disappear.

The high tolerance limits of the well-supported human body to decelerative forces suggest that in aircraft and other vehicles, seats, floors, and the whole inner structure surrounding crew and passengers should be designed to resist crash decelerations as near to $40g$ as weight or space limitations permit.[26] The structural members surrounding this inner compartment should be arranged so that their crushing reduces forces on the inner structure. Protruding and easily loosened objects should be avoided. To allow the best chance for survival, seats should also be stressed for dynamic loadings between 20 and $40g$. Civil Air Regulations require a minimum static strength of seats of $9g$. A method for computing seat tolerance for typical survivable airplane crash decelerations is available for seats of conventional design.[26] It has been established that a passenger who is riding in a seat facing backward has a better chance to survive an abrupt crash deceleration since the impact forces are then more uniformly distributed over the body. Neck injury must be prevented by proper head support. Objections to riding backward on a railway or in a bus are minimized for air transportation because of the absence of disturbing motion of objects in the immediate field of view. Another consideration concerning the direction of passenger seats in aircraft stems from the fact that for a rearward-facing seat the center of passenger support during deceleration is about 1 ft above the point where the seat belt would be attached for a forward-facing passenger. Consequently, the rearward-facing seat is subjected to a higher bending moment; in other words, for seats of the same weight the forward-facing seat will sustain higher crash forces without collapse. For the same seat weight, the rearward-facing seat will have approximately only half the design strength of the forward-facing seat and about one-third its natural frequency.

Increased safety in automobile as well as airplane crashes can be obtained by distributing the impact load over larger areas of the body and fixing the body more rigidly to the seat. Shoulder straps, thigh straps, chest straps, and hand holds are additional body supports used in experiments. They are illustrated in Fig. 42.21. Table 42.3 shows the desirability of these additional restraints to increase possible survivability to acceleration loads of various direction. In airplane crashes, vertical and horizontal loads must be anticipated. In automobile crashes, horizontal loads are most likely.

Safety lap, or seat, belts are used to restrain the occupants of aircraft or automobiles and to prevent their being hurled about within, or being ejected from, the car or aircraft. Their effectiveness has been proved by many laboratory tests and in actual crash accidents. A forward-facing passenger held by a seat belt flails about when suddenly decelerated; his hands, feet, and upper torso swing forward until his chest hits his knees or until the body is stopped in this motion by hitting other objects (back of seat in front, cabin wall, instrument panel, steering wheel, control stick, see Fig. 42.22). Since 15 to $18g$ longitudinal deceleration can result in 3 times higher acceleration of the chest hitting the knees, this load appears to be about the limit a human can tolerate with a seat belt alone. Approximately the same limit is obtained when the head-neck structure is considered.

The effectiveness of adequately engineered shoulder or chest straps in automobile crashes is illustrated in Fig. 42.22. Lap straps always should be as tight as comfort will permit to exclude available slack. During forward movement, about 60 percent of the body mass is restrained by the belt, and therefore represents the belt load. If the upper torso is fixed to the back of the seat by any type of harness (shoulder harness,

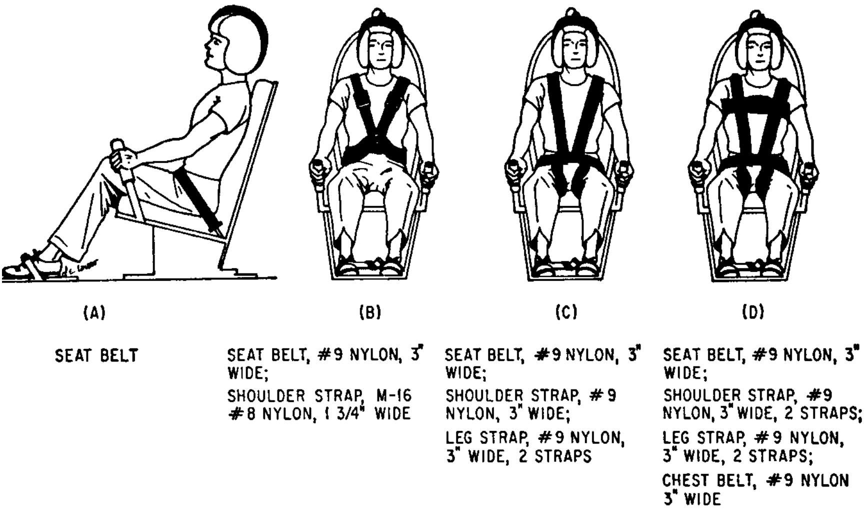

FIGURE 42.21 Protective harnesses for rapid accelerations or decelerations. The following devices were evaluated in sled deceleration tests: (*A*) Seat belt for automobiles and commercial aviation. (*B*) Standard military lap and shoulder strap. (*C*) Like (*B*) but with thigh straps added to prevent headward rotation of the lap strap. (*D*) Like (*C*) but with chest strap added. (*Stapp:* USAF Tech. Rept. 5915, *pt. I, 1949; pt. II, 1951.*)

chest belt, etc.), the load on the seat is approximately the same for forward- and aft-facing seats. The difference between these seats with respect to crash tolerance as discussed above no longer exists. These body restraints for passenger and crew must be applied without creating excessive discomfort.

A rapidly inflating "air bag" situated in front of an automobile driver, and often the front passenger, and inflated on frontal collision, has been installed in most vehicles during the last decade. While initially conceived as an alternative to passive restraints, that is, as a safety system that would operate when an automobile occupant was not wearing a seat belt, air bags are now recognized to provide most benefit when considered as a complementary system to lap and shoulder seat belts. The device consists of a crash sensor, or sensors, mounted near the front of the vehicle that signal velocity changes to a controller; those in excess of about 20 ft/sec cause a pyrotechnic reaction to generate gas that inflates a porous fabric bag within, typically, 25 m/sec, so that the bag is inflated sufficiently to distribute the deceleration forces over a large surface area on contact with the occupant. Accident data have shown that while air bags do save lives, believed to be some 2,620 people in the United States from 1990 to 1997, they were also responsible for the deaths of at least 44 children and 36 adults during this period.[27] Most of the fatalities have been attributed to the size and position of the occupant at the time of impact with the air bag, which is not defined if a seat belt is not worn (see Fig. 42.22). In these circumstances, the air bag may impact the occupant with sufficient force to produce fatal injury. Systems are under development to mitigate these effects (e.g., reducing the inflation rate of the bag and monitoring occupant position).[27]

The dynamic properties of seat cushions are extremely important if an acceleration force is applied through the cushion to the body. In this case the steady-state

TABLE 42.3 Human-Body Restraint and Possible Increased Impact Survivability (*After Eiband*.[36])

Direction of acceleration imposed on seated occupants	Conventional restraint	Possible survivability increases available by additional body supports*
Spineward:		
Crew	Lap strap Shoulder straps	Forward facing: (*a*) Thigh straps (assume crew members will be performing emergency duties with hands and feet at impact)
Passengers	Lap strap	Forward facing: (*a*) Shoulder straps, (*b*) thigh straps, (*c*) nonfailing arm rests, (*d*) suitable hand holds, and (*e*) emergency toe straps in floor
Sternumward:		
Passengers only	Lap strap	Aft facing: (*a*) Nondeflecting seat back, (*b*) integral, full-height head rest, (*c*) chest strap (axillary level), (*d*) lateral head motion restricted by padded "winged back," (*e*) leg and foot barriers, and (*f*) arm rests and hand holds (prevent arm displacement beyond seat back)
Headward:		
Crew	Lap strap Shoulder straps	Forward facing: (*a*) Thigh straps, (*b*) chest strap (axillary level), and (*c*) full, integral head rest (assume crew members will be performing emergency duties; extremity restraint useless)
Passengers	Lap strap	Forward facing: (*a*) Shoulder straps, (*b*) thigh straps, (*c*) chest strap (axillary level), (*d*) full, integral head rest, (*e*) nonfailing contoured arm rests, and (*f*) suitable hand holds Aft facing: (*a*) Chest strap (axillary level), (*b*) full, integral head rest, (*c*) nonfailing arm rests, and (*d*) suitable hand holds
Tailward:		
Crew	Lap strap Shoulder straps	Forward facing: (*a*) Lap-belt tie-down strap (assume crew members will be performing emergency duties; extremity restraint useless)
Passengers	Lap strap	Forward facing: (*a*) Shoulder straps, (*b*) lap-belt tie-down strap, (*c*) hand holds, (*d*) emergency toe straps Aft facing: (*a*) Chest strap (axillary level), (*b*) hand holds, and (*c*) emergency toe straps
Berthed occupants	Lap strap	Feet forward: Full-support webbing net Athwart ships: Full-support webbing net

* Exposure to maximum tolerance limits (see Figs. 42.28 to 42.35) requires straps exceeding conventional strap strength and width.

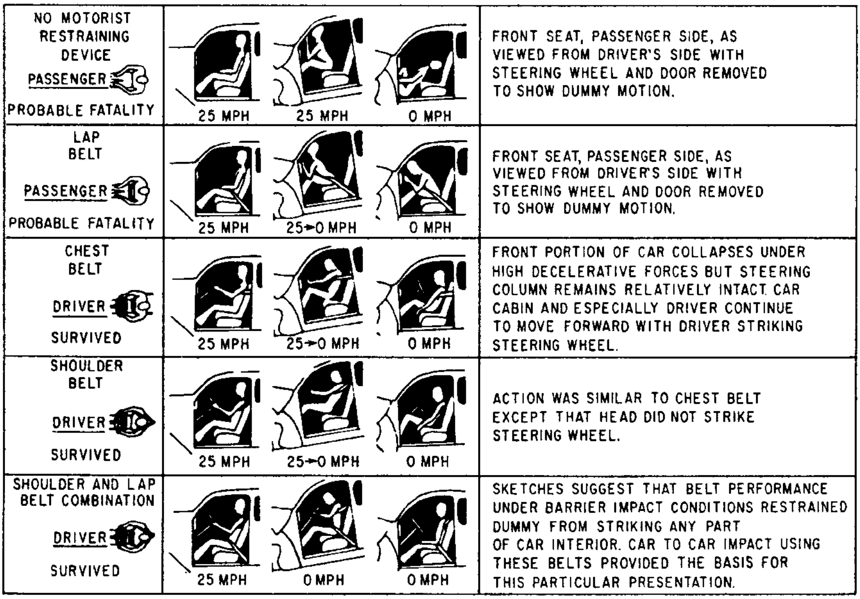

FIGURE 42.22 Effect of varying safety-belt arrangements on driver and passenger for a 25-mph automobile collision with a fixed barrier. The sketches and evaluations are based on actual collision tests. (*Severy and Matthewson:* Trans. SAE, ***65***:*70, 1957.*)

response curve of the total man-seat system (Fig. 42.19) provides a clue to the possible dynamic load factors under impact. Overshooting should be avoided, at least for the most probable impact rise times. This problem has been studied in detail in connection with seat cushions used on upward ejection seats. The ideal cushion is approached when its compression under static load spreads the load uniformly and comfortably over a wide area of the body and if almost full compression is reached under the normal weight. The impact acceleration then acts uniformly and almost directly on the body without intervening elastic elements. A slow-responding foam plastic, such as an open cell rate-dependent polyurethane foam, of thickness from 2 to 2.5 in. satisfies these requirements.[28]

A significant factor in human impact tolerance appears to be the acceleration-time history of the subject immediately preceding the impact event. A so-called *dynamic preload* consists of an imposed acceleration preceding, and/or during, and in the same direction as the impact acceleration.[29] A dynamic preload occurs, for example, when the brakes are applied to a moving automobile before it hits a barrier. The phenomenon is found experimentally to reduce the acceleration of body parts on impact, thereby potentially mitigating adverse health effects. The dynamic preload should not be confused with the static preload introduced by a protective harness. The latter brings the occupant into contact with the seat or restraint but does not introduce the dynamic displacement of body parts and tissue compression necessary to reduce the body's dynamic response.

A summary of methods used to improve the performance of occupant restraint systems in motor vehicles is given in Table 42.4.

TABLE 42.4 Occupant Restraint Performance Enhancers (*After R. Eppinger, in "Accidental Injury: Biomechanics and Prevention," p. 196, Springer-Verlag, New York, 1993.*)

Belt systems
Pretensioner—A device, activated by a sensor detecting the onset of the crash, which retracts belts rapidly, removing any slack and coupling the occupant to the vehicle structure sooner than a standard belt system would • Effect: Maximizes time and distance over which belt forces are applied, applies greater restraint forces earlier in the crash event, and therefore affords greater energy extraction
Variable stiffness seat cushion—A cushion that is stiffer at the front edge than toward the rear by the seat back • Effect: Prevents the pelvis from moving down while being restrained by a lap-belt system, thus maintaining correct belt geometry to ensure the belt remains on the bony pelvis and does not slip up and over the pelvis into the soft abdomen and cause injuries
Force-limiting belt webbing—Seat-belt webbing designed to stretch at a predetermined level • Effect: Limits the maximum force applied to the body and allows the body to translate more within the occupant compartment as well as over the ground, thus extracting a greater amount of the initial kinetic energy
Retracting steering system—A steering system designed to move forward within the compartment as the front of the car crushes during an impact • Effect: Provides a greater translation distance within the compartment for the driver using a safety-belt system, thus increasing the energy extraction potential with reduced force
Inflating belts—A torso belt system, which, upon sensing the initiation of a crash, inflates to large-diameter cylinder • Effect: Increases the contact area between the belt and the thorax as well as removing slack for the system and coupling the occupant to the vehicle earlier in the crash sequence
Web lockers—A device that clamps the torso belt and prevents it from unwrapping from its take-up spool or reel • Effect: Prevents torso belt from becoming longer and allowing the occupant to translate within the compartment without substantial restraint forces being applied to occupant
Air bag systems
Dual inflation levels—Air bag systems that, depending on the logic provided, are either crash-sensitive and/or occupant-sensitive, will inflate with different rates and/or volumes of gas depending on the intensity of the crash and/or the size or proximity of occupant to the inflation module • Effect: Crash- or size-sensitive systems modulate the forces applied to the occupant according to need (greater forces in higher-intensity crashes or for heavier occupants), thus allowing optimal stroke within compartment. Proximity-sensitive systems reduce inflation rate when occupant near module to prevent unnecessarily high forces being applied
Dual air bags—An air bag within an air bag where the inflator directly inflates the inner bag and then vents the gases into the outer bag to inflate it • Effect: Allows the small-volume inner bag to inflate rapidly and apply forces earlier in the crash event with the subsequent large outer bag adding area and force capability later in the crash event
Precrash sensing (anticipatory)—Means by which an imminent crash is sensed prior to the actual initiation of the crash and restraint operations are begun • Effect: Allows the restraint system to initiate its application of forces earlier in the crash event by either pretensioning-belt systems or initiating inflation of air bag sooner
Stroking columns—Specifically designed column support structure that deforms in a specified direction while applying a controlled force • Effect: Allows the occupant to have greater stroke within the compartment, thus extending the time over which he accomplishes the necessary velocity change and extracting more of his kinetic energy
Air bag/seat-belt combination—Safety system designed to exploit advantages of both restraint systems • Effect: Employs the best operational characteristics of both restraint concepts to provide optimal restraint for the occupant

PROTECTION AGAINST HEAD IMPACT

The impact-reducing properties of protective helmets are based on two principles: the distribution of the load over a large area of the skull and the interposition of energy-absorbing systems. The first principle is applied by using a hard shell, which is suspended by padding or support webbing at some distance from the head (⅝ to ¾ in.). High local impact forces are distributed by proper supports over the whole side of the skull to which the blow is applied. Thus, skull injury from relatively small objects and projectiles can be avoided. However, tests usually show that contact padding alone over the skull results in most instances in greater load concentration, whereas helmets with web suspension distribute pressures uniformly. Since helmets with contact padding usually permit less slippage of the helmet, a combination of web or strap suspension with contact padding is desirable. The shell itself must be as stiff as is compatible with weight considerations; when the shell is struck by a blow, its deflection must not be large enough to permit it to come in contact with the head.

Padding materials can incorporate energy-absorptive features. Whereas foam rubber and felt are too elastic to absorb a blow, foam plastics like polystyrene or Ensolite result in lower transmitted accelerations.

Most helmets constitute compromises among several objectives such as pressurization, communication, temperature conditioning, minimum bulk and weight, visibility, protection against falling objects, etc.; usually, impact protection is but one of many design considerations.[30] The protective effect of helmets against concussion and skull fracture has been proved in animal experiments and is apparent from accident statistics.

PROTECTION AGAINST BLAST WAVES

Individual protection against air blast waves is extremely difficult since only very thick protective covers can reduce the transmission of the blast energy significantly. Furthermore, not only the thorax but the whole trunk would require protection. In animal experiments, sponge-rubber wrappings and jackets of other elastic material have resulted in some reduction of blast injuries.[18] Enclosure of the animal in a metal cylinder with the head exposed to the blast wave has provided the best protection—short of complete enclosure of the animal. Therefore it is generally assumed that shelters are the only practical means of protecting humans against blast. They may be of either the open or closed type; both change the pressure environment. Changes in pressure rise time introduced by the door or other restricted openings are physiologically most important.

HUMAN TOLERANCE CRITERIA

WHOLE-BODY VIBRATION EXPOSURE

International Standard ISO 2631 defines methods for the measurement of periodic, random, and transient whole-body vibration. The standard also describes the principal factors that combine to determine the acceptability of an exposure and suggests the possible effects, recognizing the large variations in responses between individuals.[31]

Measurement. Whole-body vibration is measured at the principal interface between the human body and the source of vibration. For seated persons, this interface is most likely to be the seat surface and seat back, if any; for standing persons, the feet; and for reclining persons, the supporting surface(s) under the pelvis, torso, and head. When vibration is transmitted to the body through a nonrigid or resilient material (e.g., a seat cushion), the measuring transducer should be within a mount, in contact with the body, formed to minimize the change in surface pressure distribution of the resilient material.[1] The measurement should be of sufficient duration to ensure that the data are representative of the exposure being assessed and, for random signals, contain acceptable statistical precision.

Frequency-Weighted Acceleration. The magnitude of the exposure is characterized by the *rms frequency-weighted acceleration* calculated in accordance with the following equation or its equivalent in the frequency domain:

$$a_W = \left[\frac{1}{T} \int_0^T a_W^2(t)\, dt \right]^{1/2} \tag{42.4}$$

where $a_w(t)$ is the frequency-weighted acceleration, or angular acceleration, at time t expressed in meters per second squared (m/s^2), or radians per second squared (rad/s^2), respectively; and T is the duration of the measurement in seconds. The frequency weightings to be employed for different applications are shown in Fig. 42.23, with their applicability summarized in Table 42.5. The coordinate systems for the directions of motion referred to in Table 42.5 are shown in Fig. 42.24. Frequency weightings W_d and W_k are the principal weightings for the assessment of the effects of vibration on health, comfort, and perception, with W_f used for motion sickness. W_c, W_e, and W_j apply to specific situations involving, respectively: motion coupled to the body from a seat back (W_c); body rotation (W_e); and head motion in the X direction of reclining persons (W_j). Application of a frequency weighting selected according to Table 42.5, Fig. 42.23, and Fig. 42.24 to one component of vibration transmitted to the body provides a measure of the *component frequency-weighted acceleration* for that direction of motion and human response.

Equation (42.4) is suitable for characterizing vibrations with a *crest factor* less than 9, where the crest factor is here defined as the magnitude of the ratio of the peak value of the frequency-weighted acceleration signal to its rms value.

Vibration Containing Transient Events. For exposures to whole-body vibration containing transient events resulting in crest factors in excess of 9, either the *running rms* or the *fourth-power vibration dose,* or both, may be used in addition to the rms frequency-weighted acceleration to ensure that the effects of transient vibrations are not underestimated. The running rms is calculated for a short integration time τ ending at time t_0 in the time record as follows:

$$a_W(t_0) = \left[\frac{1}{\tau} \int_{(t_0 - \tau)}^{t_0} a_W^2(t)\, dt \right]^{1/2} \tag{42.5}$$

A correlation with some subjective human responses to transient vibration may be obtained by constructing the *maximum transient vibration value* $\mathrm{MTVV}_{(T)}$ during the measurement

$$\mathrm{MTVV}_{(T)} = |a_W(t_0)|_{\max} \tag{42.6}$$

where the right-hand side of this equation is determined by the maximum value of the running rms acceleration obtained using Eq. (42.5) when $\tau = 1$ sec.

The fourth-power *vibration dose value* VDV is defined by

$$\mathrm{VDV} = \left[\int_0^T a_W^r(t)\, dt \right]^{1/r} \tag{42.7}$$

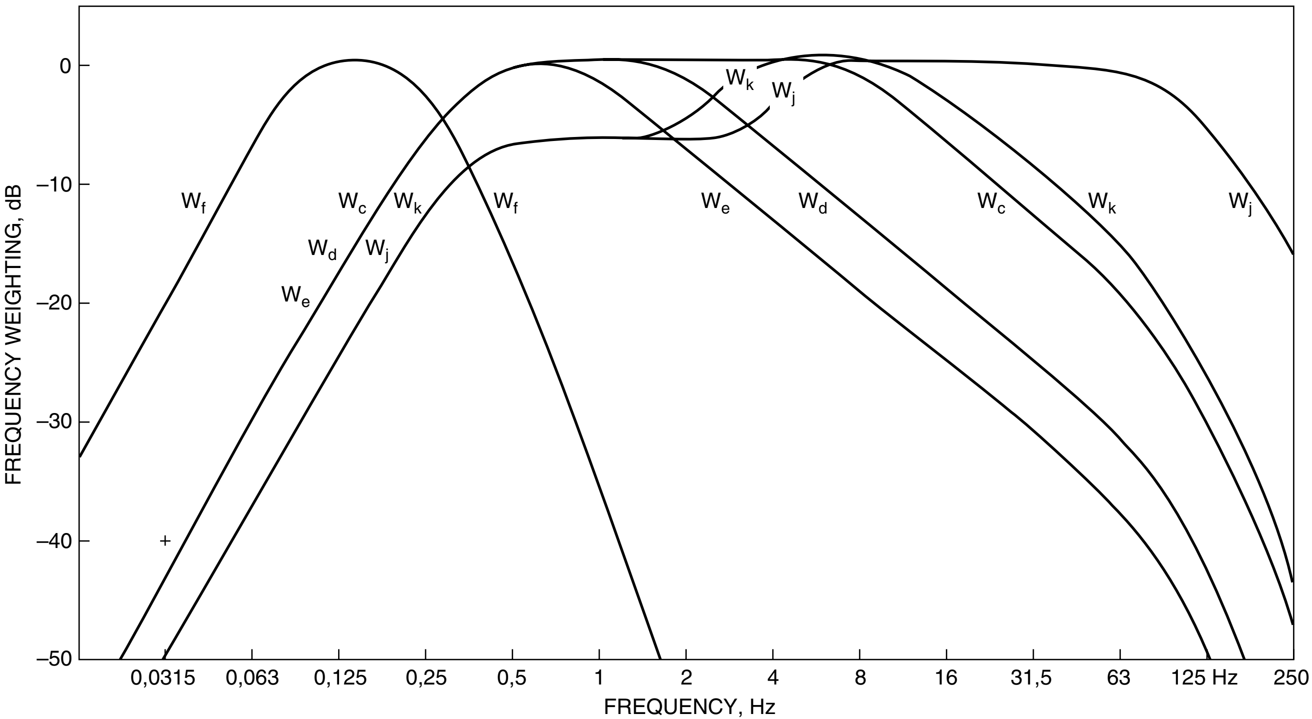

FIGURE 42.23 Frequency-weighting curves for the assessment of whole-body vibration. The application of these curves to the assessment of human health, comfort, perception, and motion sickness is summarized in Table 42.5. (*ISO 2631-1*.[31])

TABLE 42.5 Applicability of Whole-Body Vibration Frequency Weightings W_k, W_d, W_f, W_c, W_e, and W_j, Shown in Fig. 42.23, for the Vibration Directions X, Y, Z, R_x, R_y, and R_z Specified in Fig. 42.24 (*ISO 2631-1*.[31])

Frequency weighting	Health	Comfort*	Perception	Motion sickness
		Principal weighting		
W_k	Z	Z-seat X,Y,Z-feet Z-standing X-lying	Z	
W_d	X-seat Y-seat	X-seat Y-seat X,Y-standing Y,Z-lying Y,Z-back	X-, Y-	
W_f				Z
		Additional weighting		
W_c	X-seat back	X-seat back		
W_e		R_x, R_y, R_z		
W_j		X-lying (head)		

* Values of the multiplying factor k to be applied to component accelerations for assessing the comfort of seated persons in situations in which vibration enters the body at several points, e.g., the seat pan, seat back, and the feet (see text).

Component Acceleration	*Value of k*
X direction at seat back	0.8
Y direction at seat back	0.5
Z direction at seat back	0.4
X & Y directions at feet	0.25
Z direction at feet	0.4
R_x axis at seat	0.63 m/rad
R_y axis at seat	0.4 m/rad
R_z axis at seat	0.2 m/rad

For other component accelerations the value of k is unity.

with $r = 4$, and provides a measure of exposure that is more sensitive to large amplitudes by forming the fourth power of the frequency-weighted acceleration time-history, $a_w^4(t)$. If the total exposure consists of i exposure elements with different vibration dose values $(\text{VDV})_i$ then

$$\text{VDV}_{\text{total}} = \left[\sum_i (\text{VDV})_i^4 \right]^{1/4} \tag{42.8}$$

Use of the maximum transient vibration value or the total vibration dose value in addition to the rms frequency-weighted acceleration is advisable whenever

$$\text{MTVV}_{(T)} > 1.5 a_W \tag{42.9}$$

or

$$\text{VDV}_{\text{total}} > 1.75 a_W T^{1/4} \tag{42.10}$$

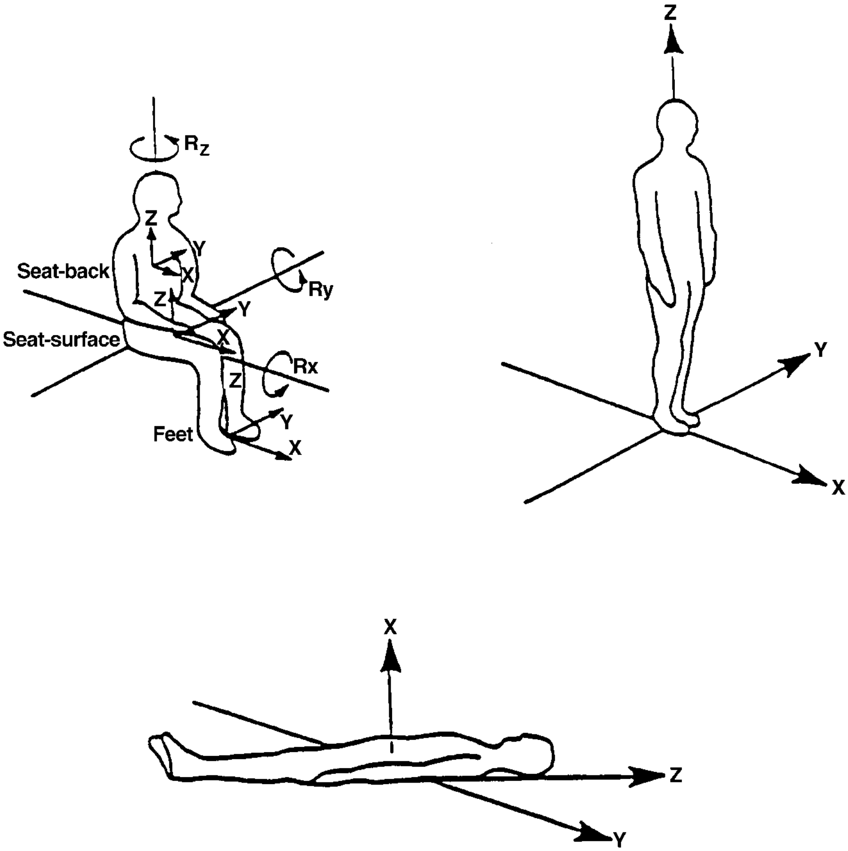

FIGURE 42.24 Basicentric axes of the human body for translational (*X, Y,* and *Z*) and rotational (R_x, R_y, and R_z) whole-body vibration. (*ISO 2631-1.*[31])

The total vibration dose value will integrate the contribution from *each* transient event, irrespective of magnitude or duration, to form a time- and magnitude-dependent dose. In contrast, the maximum transient vibration value will provide a measure dominated by the magnitude of the most intense event occurring in a 1-second time interval, and will be little influenced by events occurring at times significantly greater than 1 second from this event. Application of either measure to the assessment of whole-body vibration should take into consideration the nature of the transient events, and the anticipated basis for the human response (i.e., source and variability of, and intervals between, transient motions, and whether the human response is likely to be dose related).

Health. Guidance for the effect of whole-body vibration on health is provided in international standard ISO 2631-1 for vibration transmitted through the seat pan in the frequency range from 0.5 to 80 Hz.[31] The assessment is based on the largest measured translational component of the frequency-weighted acceleration (see Fig. 42.24 and Table 42.5). If the motion contains transient events that result in the con-

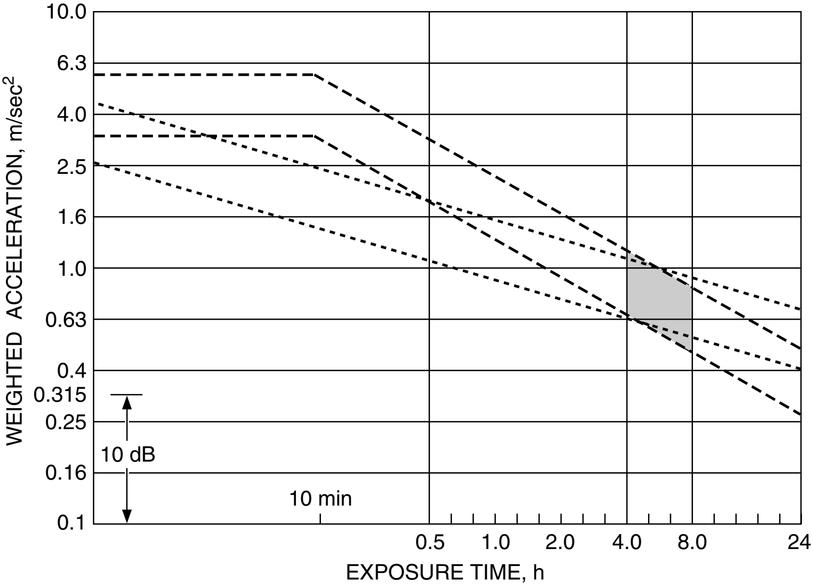

FIGURE 42.25 Health guidance caution zone for exposure to whole-body vibration. The dashed lines employ a relationship between stimulus magnitude and exposure time in hours [Eq. (42.2)] with $n = 2$ and the dotted lines $n = 4$. For exposures below the shaded zone, health effects have not been reproducibly observed; for exposures above the shaded zone, health effects may occur. The lower and upper dotted lines correspond to vibration dose values of 8.5 and 17, respectively. (*ISO 2631-1*.[31])

dition in Eq. (42.10) being satisfied, then a further assessment may be made using the vibration dose value. The frequency weightings to be applied, W_d and W_k (see Table 42.5), are to be multiplied by factors of unity for vibration in the Z direction and 1.4 for the X and Y directions of the coordinate system shown in Fig. 42.24. The largest component-weighted acceleration is to be compared at the daily exposure duration with the shaded health caution zone in Fig. 42.25. The dashed lines in this diagram correspond to a relationship between the physical magnitude of the stimulus and exposure time with an index of $n = 2$ in Eq. (42.2), while the dotted lines correspond to an index of $n = 4$ in this equation. The lower and upper dotted lines in Fig. 42.25 correspond to vibration dose values of 8.5 and 17, respectively. For exposures below the shaded zone, which has been extrapolated to shorter and longer daily exposure durations in the diagram, health effects have not been reproducibly observed; for exposures within the shaded zone, the potential for health effects increases; for exposures above the zone, health effects are expected.[13]

If the total daily exposure is composed of several exposures for times t_i to different frequency-weighted component accelerations $(a_W)_i$ then the equivalent acceleration magnitude corresponding to the total time of exposure $(a_W)_{\text{equiv}}$ may be constructed using

$$(a_W)_{\text{equiv}} = \left[\frac{\sum_i (a_W)_i^2 t_i}{\sum_i t_i} \right]^{1/2} \tag{42.11}$$

To characterize occupational exposure to whole-body vibration, the 8-hour frequency-weighted component accelerations may be measured according to Eq. (42.4) with $T = 28{,}800$ seconds. The total daily vibration dose value is constructed using Eq. (42.8).

A method for assessing the effect of repeated, large magnitude (i.e., in excess of the acceleration of gravity), transient events on health is described under *Multiple Shocks in the Vertical Direction.*

Discomfort. Guidance for the evaluation of comfort and vibration perception is provided in international standard ISO 2631-1 for the exposure of seated, standing, and reclining persons (the last-mentioned supported primarily at the pelvis).[31] The guidance concerns translational and rotational vibration in the frequency range from 0.5 to 80 Hz that enters the body at the locations, and in the directions, listed in Table 42.5. The assessment is formed from rms component accelerations. For transient vibration, the maximum transient component vibration values should be considered if the condition in Eq. (42.9) is satisfied, while the magnitude of the vibration dose value may be used to compare the relative comfort of events of different durations. Each measure is to be frequency weighted according to the provisions of Table 42.5 and Fig. 42.24. Frequency weightings other than those shown in Fig. 42.23 have been found appropriate for some specific environments, such as for passenger and crew comfort in railway vehicles.[32]

Overall Vibration Value. The vibration components measured at a point where motion enters the body may be combined for the purposes of assessing comfort into a so-called *frequency-weighted acceleration sum* a_{WAS}, which for orthogonal translational component accelerations a_{WX}, a_{WY}, and a_{WZ}, is

$$a_{\mathrm{WAS}} = [a_{\mathrm{WX}}^2 + a_{\mathrm{WY}}^2 + a_{\mathrm{WZ}}^2]^{1/2} \tag{42.12}$$

An equivalent equation may be used to combine rotational acceleration components.

When vibration enters a seated person at more than one point (e.g., at the seat pan, the backrest, and the feet), a weighted acceleration sum is constructed for each entry point. In order to establish the relative importance of these motions to comfort, the values of the component accelerations at a measuring point are ascribed a magnitude multiplying factor k so that, for example, a_{WX}^2 in Eq. (42.12) is replaced by $k^2 a_{\mathrm{WX}}^2$, etc. The values of k are listed in Table 42.5, and are dependent on vibration direction and where motion enters the seated body. The *overall vibration total value* a_{overall} is then constructed from the root sum of squares of the frequency-weighted acceleration sums recorded at different measuring points, i.e.

$$a_{\mathrm{overall}} = [a_{\mathrm{WAS}_1}{}^2 + a_{\mathrm{WAS}_2}{}^2 + a_{\mathrm{WAS}_3}{}^2 + \cdots]^{1/2} \tag{42.13}$$

where the subscripts 1,2,3, etc., identify the different measuring points.

Many factors, in addition to the magnitude of the stimulus, combine to determine the degree to which whole-body vibration causes discomfort (see *Effects of Mechanical Vibration* above). Probable reactions of persons to whole-body vibration in public transport vehicles are listed in Table 42.6 in terms of overall vibration total values.

Fifty percent of alert, sitting or standing, healthy persons can detect vertical vibration with a frequency-weighted acceleration of 0.015 meter per sec^2.

ACCEPTABILITY OF BUILDING VIBRATION

The vibration of buildings is commonly caused by motion transmitted through the building structure from, for example, machinery, road traffic, and railway and sub-

TABLE 42.6 Probable Subjective Reactions of Persons Seated in Public Transportation to Whole-Body Vibration Expressed in Terms of the Overall Vibration Value (defined in text) (*ISO 2631-1.*[31])

Vibration	Reaction
Less than 0.315 m/s^2	Not uncomfortable
0.315 to 0.63 m/s^2	A little uncomfortable
0.5 to 1 m/s^2	Fairly uncomfortable
0.8 to 1.6 m/s^2	Uncomfortable
1.25 to 2.5 m/s^2	Very uncomfortable
Greater than 2 m/s^2	Extremely uncomfortable

way trains. Experience has shown that the criterion of acceptability for continuous, or intermittent, building vibration lies at, or only slightly above, the threshold of perception for most living spaces. Furthermore, complaints will depend on the specific circumstances surrounding vibration exposure. Guidance is provided for building vibration in Part 2 of the international standard for whole-body vibration, for the frequency range from 1 to 80 Hz,[33] and is adapted here to reflect alternate procedures for estimating the acceptability of building vibration (see Ref. 1).

In order to estimate the response of occupants to building vibration, the motion is measured on a structural surface supporting the body at, or close to, the point of entry of vibration into the body. For situations in which the direction of vibration and the posture of the building occupants are known (i.e., standing, sitting, or lying), the evaluation is based on the magnitudes of the component frequency-weighted accelerations measured in the X, Y, and Z directions shown in Fig. 42.24, using the frequency weightings for comfort, W_k and W_d, as appropriate (see Table 42.5 and Fig. 42.23). If the posture of the occupant with respect to the building vibration changes or is unknown, a so-called *combined* frequency weighting may be employed which is applicable to all directions of motion entering the human body, and has attenuation proportional to

$$10 \log[1 + (f/5.6)^2] \tag{42.14}$$

where the frequency f is expressed in hertz. No adverse reaction from occupants is expected when the rms frequency-weighted acceleration of continuous or intermittent building vibration is less than 3.6×10^{-3} meter/sec^2.

Transient building vibration, that is, motion which rapidly increases to a peak followed by a damped decay (which may, or may not, involve several cycles of vibration), may be assessed either by calculating the maximum transient vibration value or the total vibration dose value using Eqs. (42.6) and (42.8), respectively. No adverse human reaction to transient building vibration is expected when the maximum rms frequency-weighted transient vibration value is less than 3.6×10^{-3} meters/sec^2, or the total frequency-weighted vibration dose value is less than 0.1 meter/sec$^{1.75}$.

Human response to building vibration depends on the use of the living space. In circumstances in which building vibration exceeds the values cited to result in no adverse reaction, the use of the room(s) should be considered. Site-specific values for acceptable building vibration are listed in Table 42.7 for common building and room uses. Explanatory comments applicable to particular room and/or building uses are provided in footnotes to that table.

TABLE 42.7 Maximum RMS Frequency-Weighted Acceleration, RMS Transient Vibration Value, MTVV, and Vibration Dose Value, VDV (defined in text) for Acceptable Building Vibration in the Frequency Range 1–80 Hz[1]

Place	Time[2]	Continuous/intermittent vibration (meters/sec^2)	Transient vibration MTVV (meters/sec^2)	Transient vibration VDV (meters/sec$^{1.75}$)
Critical working areas (e.g., hospital operating rooms)[3]	Any	0.0036	0.0036	0.1
Residences[4,5]	Day	0.0072	$0.07/n^{1/2}$	0.2
	Night	0.005	0.007	0.14
Offices[5]	Any	0.014	$0.14/n^{1/2}$	0.4
Workshops[5]	Any	0.028	$0.28/n^{1/2}$	0.8

[1] The probability of adverse human response to building vibration that is less than the weighted accelerations, MTVVs, and VDVs listed in this table is small. Complaints will depend on specific circumstances. For an extensive review of this subject, see Ref. 1. Note that: (*a*) VDV has been used for the evaluation of continuous and intermittent, as well as for transient, building vibration; and (*b*) annoyance from acoustic noise caused by vibration (e.g., of walls or floors) has not been considered in formulating the guidance in Table 42.7.

[2] Daytime may be taken to be from 7 AM to 9 PM and nighttime from 9 PM to 7 AM.

[3] The magnitudes of transient vibration in hospital operating theaters and critical working places pertain to those times when an operation, or critical work, is in progress.

[4] There are wide variations in human tolerance to building vibration in residential areas.

[5] n is the number of discrete transient events that are 1 second or less in duration. When there are more than 100 transient events during the exposure period, use $n = 100$.

It should be noted that building vibration at frequencies in excess of 30 Hz may cause undesirable acoustical noise within rooms, a subject not considered in this chapter. In addition, the performance of some extremely sensitive or delicate operations (e.g., microelectronics fabrication) may require control of building vibration more stringent than that acceptable for human habitation.

MOTION SICKNESS

Guidance for establishing the probability of whole-body vibration causing motion sickness is obtained from international standard ISO 2631-1 by forming the *motion sickness dose value,* MSDV_z.[31] This energy-equivalent dose value is given by the term on the right-hand side of Eq. (42.7) with $r = 2$, and the acceleration time-history frequency-weighted using W_f (see Fig. 42.23). If the exposure is to continuous vibration of near constant magnitude, the motion sickness dose value may be approximated by the frequency-weighted acceleration recorded during a measurement interval τ of at least 240 seconds by

$$\mathrm{MSDV}_z \approx [a_{WZ}{}^2 \tau]^{1/2} \tag{42.15}$$

While there are large differences in the susceptibility of individuals to the effects of low-frequency vertical vibration (0.1 to 0.5 Hz), the percentage of persons who may vomit is

$$P = K_m(\mathrm{MSDV}_z) \tag{42.16}$$

where K_m is a constant equal to about one-third for a mixed population of males and females. Note that females are more prone to motion sickness than males.

Further guidance for the evaluation of exposure to extremely low frequency whole-body vibration (0.063 to 1 Hz) such as occurs on off-the-shore structures is to be found in ISO 6987.[34]

HAND-TRANSMITTED VIBRATION

Guidance for the measurement and assessment of hand-transmitted vibration is provided in international standard ISO 5349-1.[35] Three, rms frequency-weighted component accelerations, a_{hwx}, a_{hwy}, and a_{hwz}, are first determined at the hand-handle interface for the directions described in Fig. 42.13, using the frequency weighting specified for all directions of vibration coupled to the hand (shown in Fig. 42.26). The values are constructed according to Eq. (42.4). The *vibration total value,* a_{hv}, is then formed, which is defined as the frequency-weighted acceleration sum constructed from the hand-transmitted component accelerations, i.e., using Eq. (42.12), but with a_{WAS} replaced by a_{hv}, a_{WX} by a_{hwx}, a_{WY} by a_{hwy}, and a_{WZ} by a_{hwz}.

If it is not possible to record the vibration in each of the three coordinate directions, then an estimate of a_{hv} is made from the largest component acceleration measured (i.e., either a_{hwx}, a_{hwy}, or a_{hwz}) by multiplying by a factor in the range from 1.0 to 1.7. The factor is designed to account for the contribution to the vibration total value from any unmeasured vibration.

The assessment of vibration exposure is based on the *8-hour energy equivalent vibration total value,* $(a_{hv})_{eq(8)}$. If the measurement procedure results in the daily exposure being composed of i exposures for times t_i to vibration total values a_{hvi}, then the 8-hour energy equivalent vibration total value is obtained by forming the sum:

$$(a_{hv})_{eq(8)} = \left[\frac{1}{28{,}800} \sum_i a_{hvi}^2 t_i\right]^{1/2} \tag{42.17}$$

If, alternatively, the measurement procedure provides a time-history of the vibration total value, $a_{hv}(t)$, then $(a_{hv})_{eq(8)}$ may be calculated directly by energy averaging for an eight-hour period, T_0:

$$(a_{hv})_{eq(8)} = \left[\frac{1}{28{,}800} \int_0^{T_0} a_{hv}^2(t)\,dt\right]^{1/2} \tag{42.18}$$

Development of White Fingers (Finger Blanching). For groups of persons who are engaged in the same work using the same, or similar, vibrating hand tools, or industrial processes in which vibration enters the hands (e.g., forestry workers using chain saws, chipping and grinding to clean castings, etc.), the number of years of exposure, on average, before 10 percent of the group experience episodes of *finger blanching,* D_y, is related to the 8-hour energy equivalent vibration total value by the relationship, shown in Fig. 42.27:

$$[(a_{hv})_{eq(8)}]^{1.06} D_y = 31.8 \tag{42.19}$$

The expression assumes that $(a_{hv})_{eq(8)}$ is expressed in m/sec^2, and D_y in years. Exposures below the line in Fig. 42.27 incur less risk of developing HAVS (hand-arm vibration syndrome). There is no epidemiologic evidence for finger blanching occur-

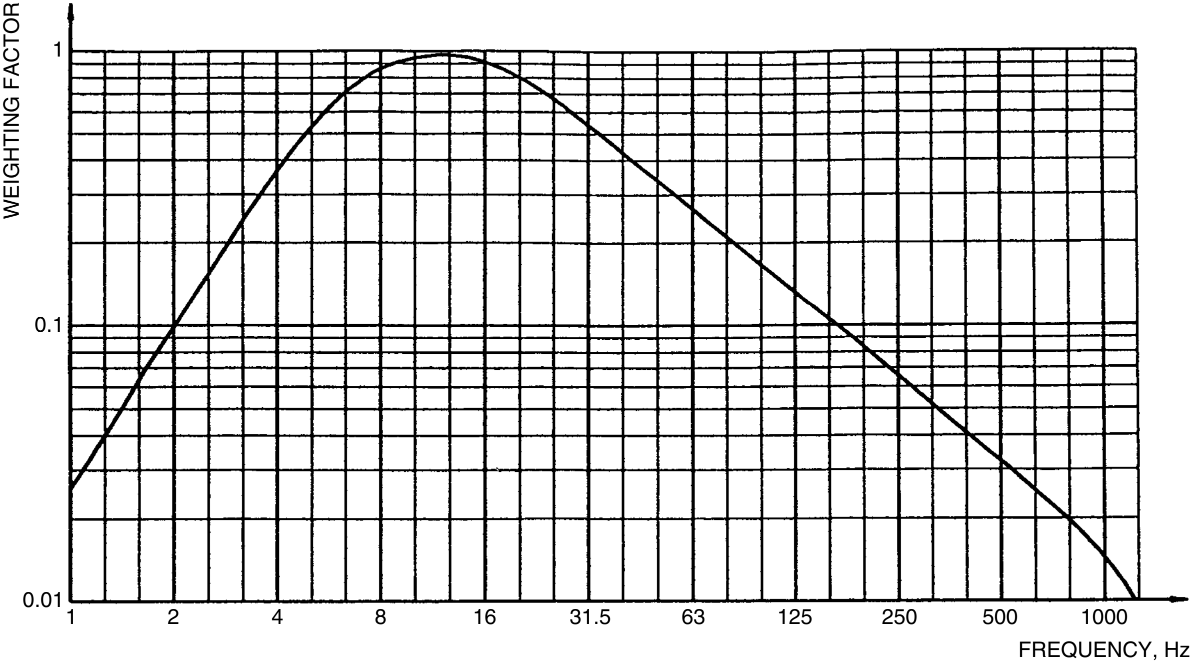

FIGURE 42.26 Frequency-weighting curve for the assessment of hand-transmitted vibration. The response shown includes band-limiting filters. (*ISO 5349.*[35])

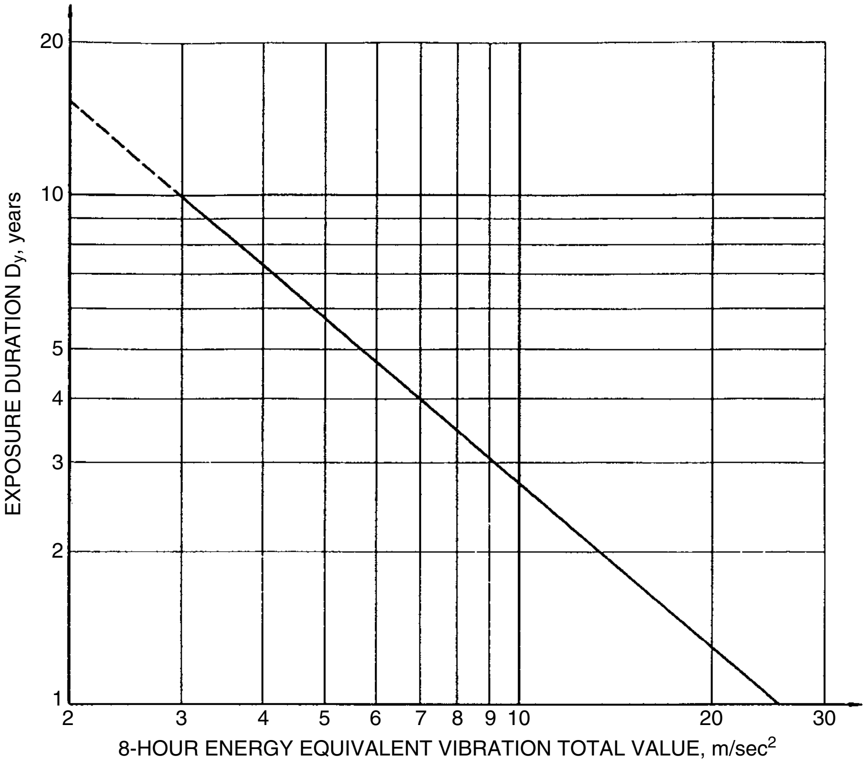

FIGURE 42.27 Duration of employment D_y, expressed in years, for 10 percent of a group of workers, all of whom perform essentially the same operations that result in exposure to effectively the same 8-hour energy equivalent vibration total value, $(a_{hv})_{eq(8)}$, to develop episodes of finger blanching. (*ISO 5349.*[35])

ring at values of $(a_{hv})_{eq(8)}$ of less than 1 m/sec^2. Deviation from the relationship shown in Fig. 42.27 may be expected for industrial situations that differ significantly from common practice (e.g., mixed occupations, such as painting for a week followed by chipping for a week).

SHOCK, IMPACT, AND RAPID DECELERATION

Human and animal experiments, frequently conducted using pneumatic or rocket-powered test sleds and water-brake deceleration, have established the tolerance of seated persons to short deceleration pulses. This unique body of information, which is unlikely to be extended for ethical reasons, was consolidated by Eiband who characterized the impacts at the seat by idealized trapezoidal time-histories, with a constant onset acceleration rate, a constant peak acceleration, and a constant decay rate.[36] The tolerance limits so obtained are shown for accelerations directed toward the spine (from in front), the sternum (from behind), the head (upward), and the tail bone (downward) in Figs. 42.28 to 42.35. The results are presented in terms of peak accelerations and their durations for the four impact directions (Figs. 42.28, 42.30,

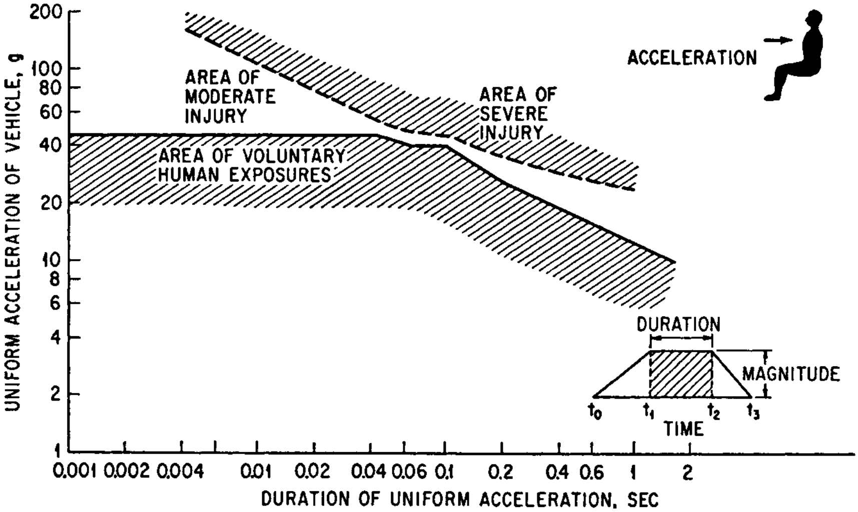

FIGURE 42.28 Tolerance to spineward acceleration as a function of magnitude and duration of impulse. (*Eiband.*[36])

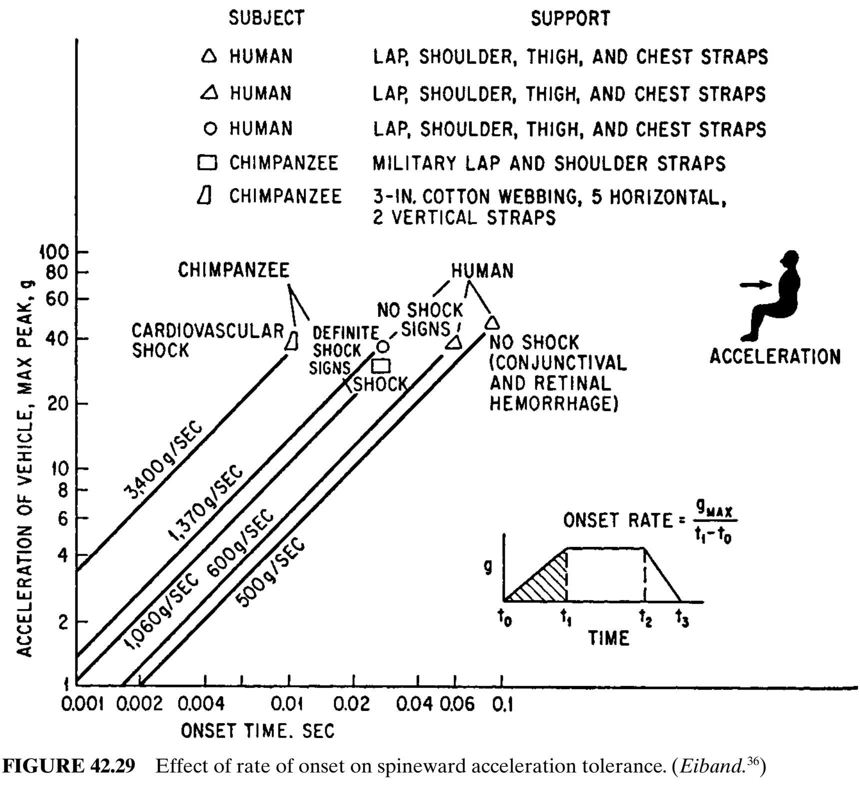

FIGURE 42.29 Effect of rate of onset on spineward acceleration tolerance. (*Eiband.*[36])

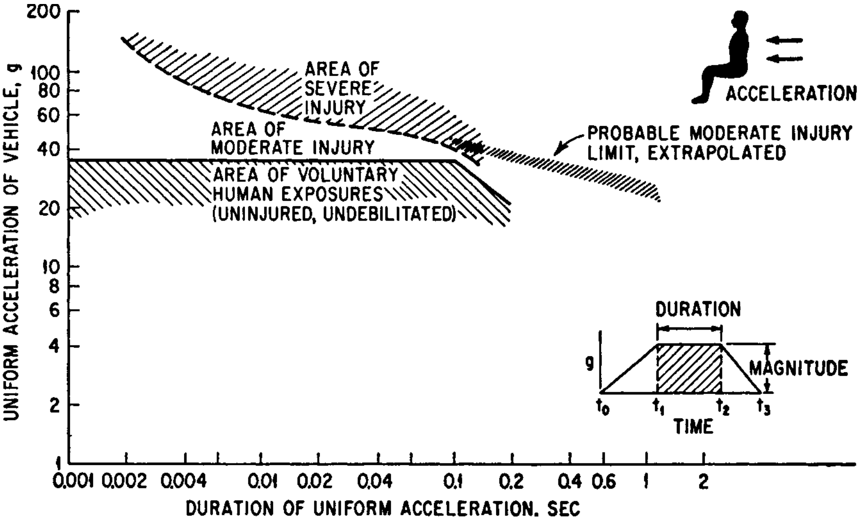

FIGURE 42.30 Tolerance to sternumward acceleration as a function of magnitude and duration of impulse. (*Eiband.*[36])

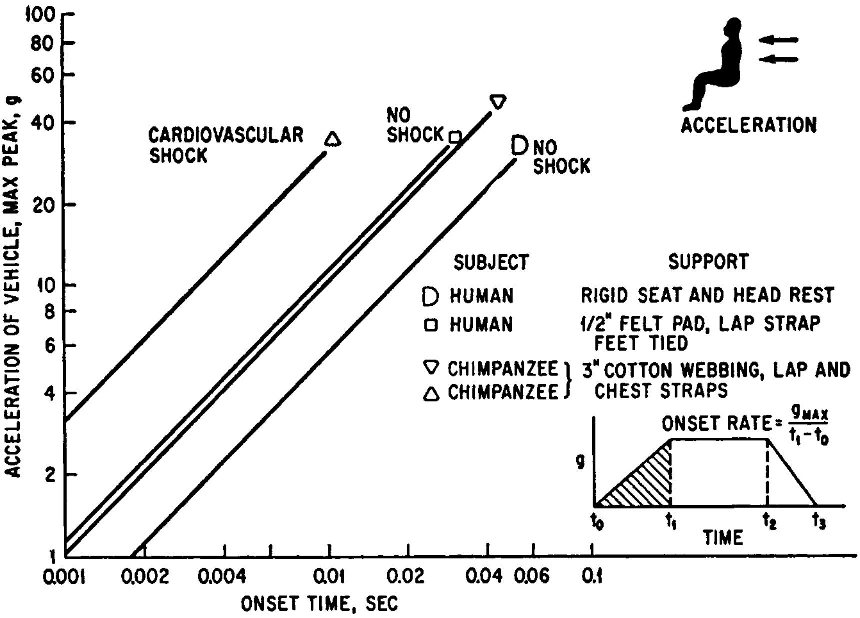

FIGURE 42.31 Effect of rate of onset on sternumward acceleration tolerance. (*Eiband.*[36])

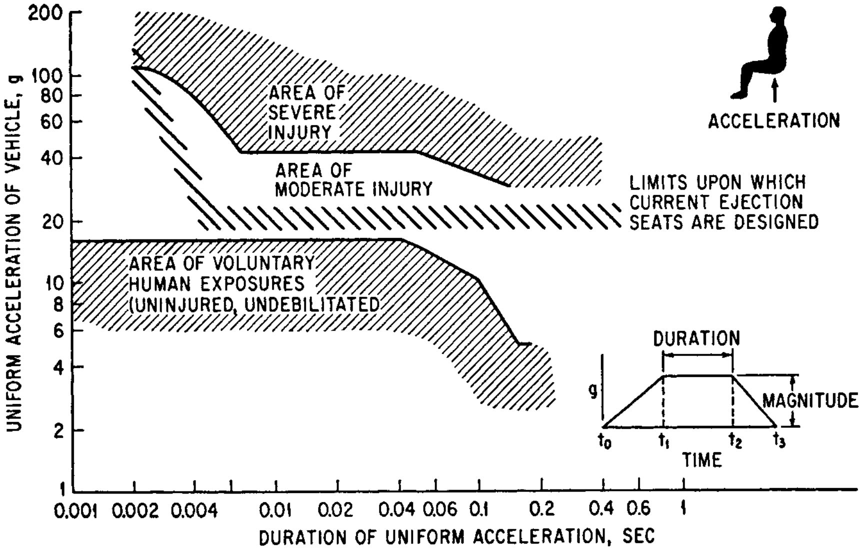

FIGURE 42.32 Tolerance to headward acceleration as a function of magnitude and duration of impulse. (*Eiband.*[36])

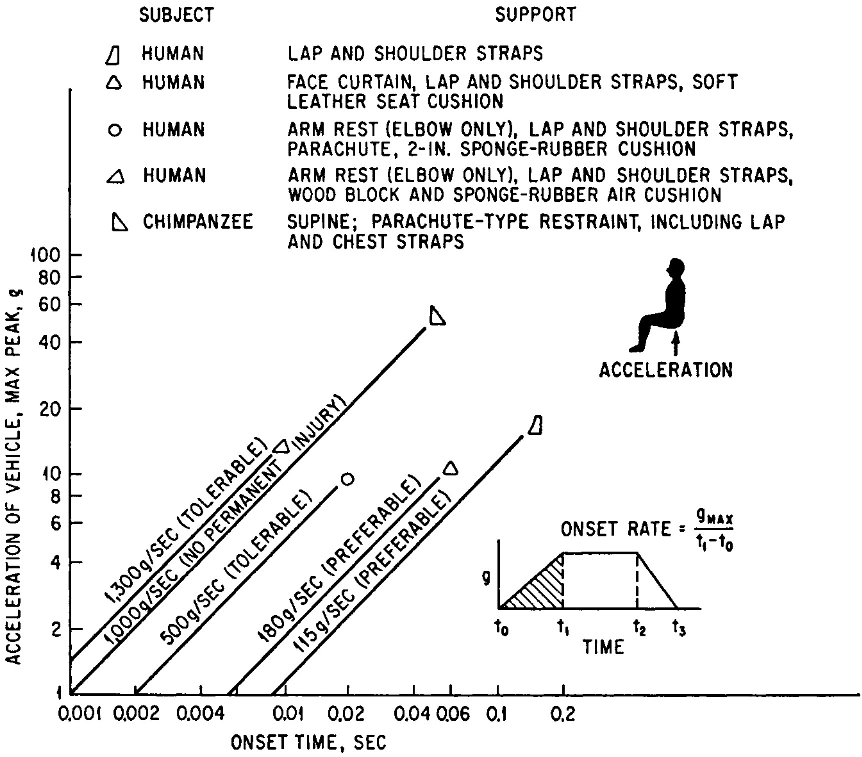

FIGURE 42.33 Effect of rate of onset on headward acceleration tolerance. (*Eiband.*[36])

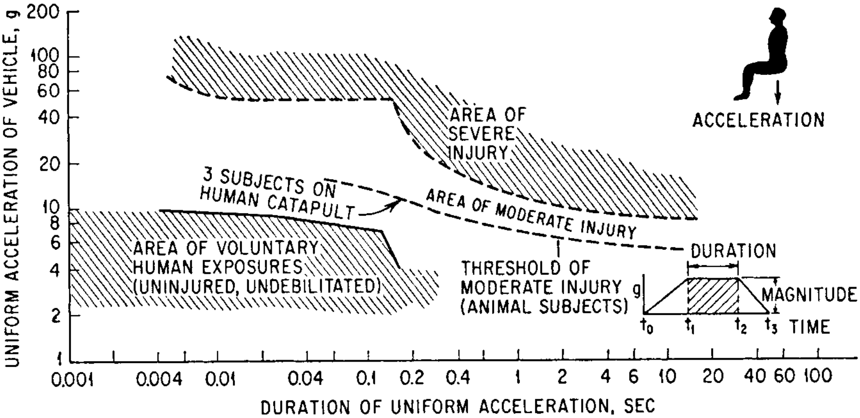

FIGURE 42.34 Tolerance to tailward acceleration as a function of magnitude and duration of impulse. (*Eiband.*[36])

42.32, and 42.34), and in terms of onset acceleration rates, which are characterized by the onset time ($t_1 - t_0$) and plotted on the abscissa of Figs. 42.29, 42.31, 42.33, and 42.35. The upper boundary of the lower shaded area in Figs. 42.28, 42.30, 42.32, and 42.34 defines the limit of voluntary human exposures that resulted in no injury. The corresponding lower boundary of the upper shaded area delineates the limit of seri-

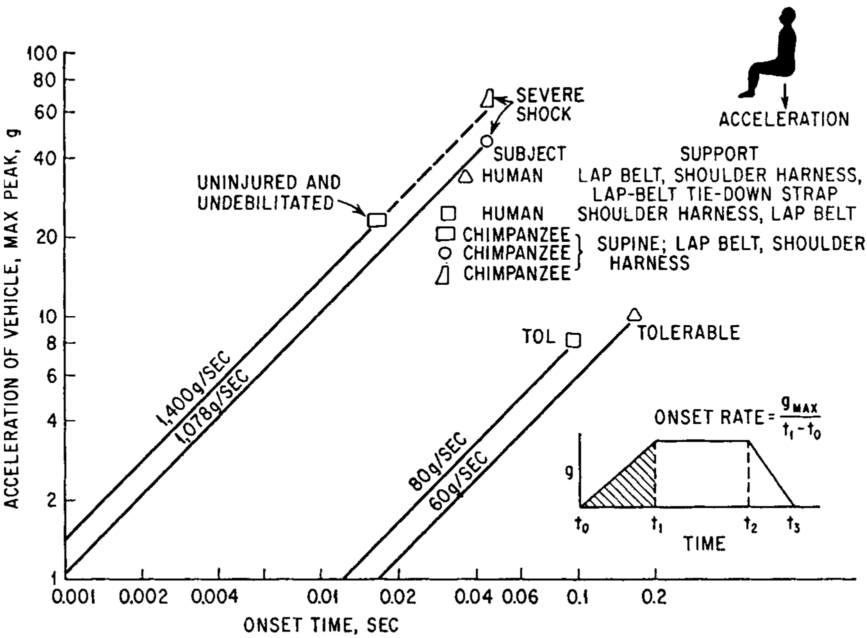

FIGURE 42.35 Effect of rate of onset on tailward acceleration tolerance. (*Eiband.*[36])

ous injury in animal experiments involving hogs and chimpanzees. No corrections for size or species differences were attempted. Maximum body support was provided to the subject in all experiments (i.e., lap belt, shoulder harness, thigh and chest straps, and arm rests, as appropriate; see Table 42.3 and Fig. 42.21).

While caution must be exercised in applying these tolerance curves, since they are based on experiments involving healthy young volunteers and animals, rigid seats, well-designed body supports, and minimum slack in harnesses, they form the primary information on which to base safety requirements for transportation vehicles. Additional sources of information have been used for specific impact conditions which, for this reason, will be described separately. Examples of short duration accelerations to illustrate the magnitudes and durations experienced in practice are listed in Table 42.8.

Single Shock in the Vertical Direction. The most common exposures of this type occur in aircraft seat ejection for which an extensive body of information and an accepted criterion exist, the latter based on a biodynamic model, namely the dynamic response index, DRI (see *Effects of Mechanical Shock*). As already noted, a DRI of 18 is predicted to correspond to a 5 percent risk of spinal injury. It should also be noted that a maximum upward acceleration of 18 to 22g is shown as the design limit for ejection seats in Fig. 42.32, which is from a 1944 ejection seat study

TABLE 42.8 Approximate Duration and Magnitude of Some Short-Duration Acceleration Loads

Type of operation	Acceleration, g	Duration, sec
Elevators:		
Average in "fast service"	0.1–0.2	1–5
Comfort limit	0.3	
Emergency deceleration	2.5	
Public transit:		
Normal acceleration and deceleration	0.1–0.2	5
Emergency stop braking from 70 mph	0.4	2.5
Automobiles:		
Comfortable stop	0.25	5–8
Very undesirable	0.45	3–5
Maximum obtainable	0.7	3
Crash (potentially survivable)	20–100	<0.1
Aircraft:		
Ordinary take-off	0.5	>10
Catapult take-off	2.5–6	1.5
Crash landing (potentially survivable)	20–100	
Seat ejection	10–15	0.25
Man:		
Parachute opening, 40,000 ft	33	0.2–0.5
6,000 ft	8.5	0.5
Parachute landing	3–4	0.1–0.2
Fall into fireman's net	20	0.1
Approximate survival limit with well-distributed forces (fall into deep snow bank)	200	0.015–0.03
Head:		
Adult head falling from 6 ft onto hard surface	250	0.007
Voluntarily tolerated impact with protective headgear	18–23	0.02

in Germany that describes the level as the limit of static and dynamic tolerance of vertebrae.[4]

Control or prevention of injury is critically dependent on optimal body positioning and restraint to minimize unwanted and forceful flexion of the spinal column. The fracture tolerance limits are influenced by age, physical condition, clothing, weight, and many other factors which detract from the optimum. If the tolerance limits are exceeded, fractures of the lumbar and thoracic vertebrae occur first. While in and of itself this injury may not be classified as severe, small changes in orientation may be enough to involve the spinal cord, an injury which is extremely severe and may be life-threatening. Neck injuries from headward accelerations appear to occur at considerably higher levels.

There have been 126 fatalities among the 620 crewmen who have escaped from a variety of U.S. Air Force aircraft from 1975 to 1991.[5] While the causes of the fatal injuries in addition to the rapid acceleration are not known (e.g., wind blast, impacting cockpit/canopy on ejection), these statistics would suggest that the single shock limit should be applied with caution and only when the body is well restrained.

Tolerance limits for downward acceleration probably are set by the compression load on the thoracic vertebrae, which are exposed to the load of the portion of the body below the chest. This load on the vertebrae is higher than that for the positive acceleration case due to the greater weight; therefore a tolerance limit for acceleration has been set at 13g. Shoulder accelerations of 13g have been tolerated by human subjects without injury, when the load was divided between hips and shoulders.

Multiple Shocks in the Vertical Direction. For evaluating exposures consisting of multiple shocks to the body, the following procedure should be considered.[37] This is based on an extension of the concept of the dynamic response index (DRI), which was introduced to quantify the potential for spinal injury associated with one large vertical acceleration (see *Effects of Mechanical Shock*). The tentative criterion for exposure to multiple shocks during a 24-hour period is given in Fig. 42.36. Upper limits of exposure are proposed for an estimated 5 percent risk of injury (dashed line), and for varying degrees of discomfort. The circles with crosses indicate exposures in which the risk of injury has been documented. The ordinate is given in terms of the DRI, which is equivalent to the maximum static acceleration (above normal gravity) and may be obtained by applying the acceleration time-history to the DRI model (Fig. 42.18).

To evaluate exposures consisting of multiple shocks of various magnitudes, if there are n_q shocks of magnitude DRI_q, where $q = 1,2,3,4 \ldots Q$, then the exposure is considered acceptable if

$$\sum_{q=1}^{Q}\left[\frac{\mathrm{DRI}_q}{(\mathrm{DRI}_{\max})_{n_q}}\right] \leq 1 \qquad (42.20)$$

In this expression, the denominator is the *maximum allowable* DRI corresponding to the *observed* number of shocks n_q with magnitude DRI_q, and is obtained from the chosen criterion curve in Fig. 42.36.

Crash. Motor vehicle and aircraft crashes commonly result in injury to occupants or fatalities from horizontal impacts. There are no internationally accepted guidelines for occupant protection, with most safety requirements being either required by law or applied voluntarily by vehicle manufacturers. Federal Motor Vehicle Safety Standards (FMVSS), promulgated in the U.S.A. by the National Highway Traffic Safety Administration (NHTSA), have had the most influence on automo-

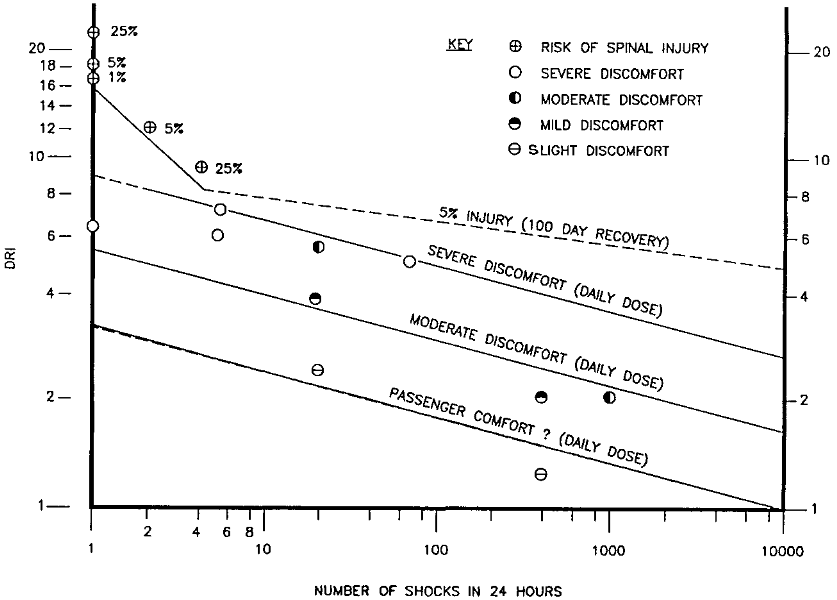

FIGURE 42.36 Tentative injury and discomfort limits for whole-body exposure to multiple impacts. The magnitude of the shocks is expressed in terms of the dynamic response index, DRI (see Fig. 42.18). (*After Allen.*[37])

tive safety, commencing with a proposal to restrict injury from the head hitting the instrument panel in 1966 (FMVSS 201). The primary concern has continued to be to reduce head injuries, considered below in more detail. Broader requirements for occupant protection including passive restraints were subsequently included in FMVSS 208 ("Occupant Crash Protection") which, as amended and expanded to include different crash configurations and injuries, forms the basis for current safety regulations.[38] In parallel with the development of regulations, the results of research on human, cadaver, animal, and surrogate exposure to impacts characteristic of those occurring in motor vehicle collisions have been summarized by the Society of Automotive Engineers (SAE).[39] SAE J885 provides biomechanical data for injuries to the head, neck, thorax, abdomen, and the lower extremities, and suggests some maximum loads, deflections, and accelerations for use in vehicle design.

Federal Aircraft Administration (FAA) regulations for improved seat strength and occupant crash injury protection in large transport aircraft [see *Protection Against Rapidly Applied Accelerations* (*Crash*)] were promulgated in the U.S.A. in 1988.[4]

Head Injury Criterion. The goal of protecting the head from irreversible brain damage in motor vehicle collisions involving unrestrained occupants led to the formulation of the Wayne State Concussion Tolerance Curve, which is shown in Fig. 42.37 as reported in SAE J885.[39] The original curve, shown by the continuous line, was derived from experiments in which instrumented, embalmed human cadavers were positioned horizontally and then dropped so that their foreheads fractured on impact with steel anvils or other targets (including motor-vehicle instrument panels). Impact durations measured on the skull of from 1 to 6 milliseconds could be

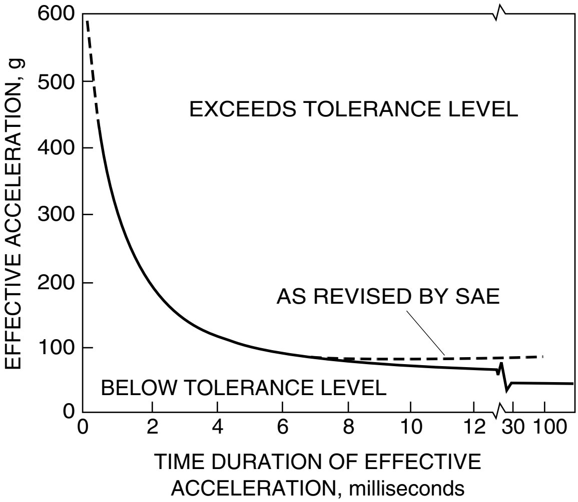

FIGURE 42.37 Wayne State Concussion Tolerance Curve. The continuous curve shows skull accelerations for impact durations of from 1 to 6 milliseconds found to produce skull fracture in embalmed cadaver heads. It has been extended to longer durations using the data of Fig. 42.28. The dashed line for durations in excess of 6 milliseconds was proposed for forehead impacts on padded surfaces. (*SAE J885.*[39])

obtained from this experiment. The tolerance curve was extended to impact durations of 100 milliseconds using an asymptotic acceleration of 42*g*, which corresponds to the limit of voluntary human exposure that resulted in no injury in Fig. 42.28 (the duration of motor vehicle crashes depends primarily on vehicle speed and typically lasts for less than 100 milliseconds). The asymptotic limit was subsequently raised to a head acceleration of 80*g* for impacts of the forehead on padded surfaces that were believed to be survivable (shown by the dashed line in Fig. 42.37).

The Wayne State Concussion Tolerance Curve has proved difficult to apply to complex acceleration-time impact waveforms, because of uncertainty in determining the effective acceleration and time. A straight-line approximation to the power curve (between 2.5 and 25 milliseconds) led to the definition of the *severity index* SI as:

$$SI = \int_0^T a^{2.5}(t)dt \tag{42.21}$$

where T is the impact duration, and $a(t)$ the acceleration time-history of the head (in units of *g*). The maximum value was proposed to be 1000. A revised index has been defined by the NHTSA for use in the frontal crash tests specified in motor vehicle regulations, which has become known as the *head injury criterion* (HIC):

$$HIC = \left| (t_2 - t_1) \left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} \right|_{\max} \tag{42.22}$$

where t_1 and t_2 are the initial and final times (in seconds) of the interval during which the HIC attains the maximum value, and $a(t)$ is measured at the center of gravity of the manikin's head. This measure is to be applied to tests using instrumented anthropometric dummies, in which a maximum value of 1000 is allowed. FMVSS 208 specifies the time interval $(t_2 - t_1)$ to be 33 milliseconds.

There are several challenges in attempting to set *human* tolerance criteria, based on either the SI or HIC.[23] First, the ability of crash tests employing HICs computed from measurements on an anthropometric dummy to rank order impact conditions by severity has been questioned. Second, the original Wayne State Concussion Tolerance Curve was designed for *unrestrained* vehicle occupants, whereas the data employed to extend the relationship to head impact durations greater than 6 milliseconds, which commonly occur in vehicle crash tests, are for subjects with optimum body restraints. Third, the basis for the Wayne State Concussion Tolerance Curve, shown by the dashed line in Fig. 42.37, suggests that criteria based on it will represent impacts that may be survivable rather than tolerable in the sense used elsewhere in this chapter (i.e., boundary between no injury and some health effect). Despite these limitations, the severity index has been successfully applied to the reduction of brain injuries in football players, by employing football helmets that attenuate head impacts to SI < 1500, while the head injury criterion remains a cornerstone of occupant safety testing for automobiles and, more recently, for transport aircraft.

Motor Vehicle Regulations. According to NHTSA statistics from 1994 to 1996, chest injury has now become the most common serious injury in motor vehicle accidents in the U.S.A. In response to this situation, NHTSA has additionally set frontal crash test limits for a Hybrid III anthropometric dummy for impacts to the chest and to the knee.[38] See the NHTSA web site (www.nhtsa.dot.gov/cars/rules/crashworthy/).

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