CHAPTER 10

VIBRATION, MECHANICAL SHOCK, AND IMPACT

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10.1 INTRODUCTION

Time-varying forces and accelerations occur in daily life, and are commonly experienced, for example, in an elevator and in aircraft, railway trains, and automobiles. All of these situations involve motion of the whole body transmitted through a seat, or from the floor in the case of a standing person, where the human response is commonly related to the relative motion of body parts, organs, and tissues. The vibration, shocks, and impacts become of consequence when activities are impaired (e.g., writing and drinking on a train, or motion sickness), or health is threatened (e.g., a motor vehicle crash). Equally important are exposures involving a localized part of the body, such as the hand and arm (e.g., when operating a hand tool), or the head (e.g., impacts causing skull fracture or concussion).

In this chapter, methods for characterizing human response to vibration, shock, and impact are considered in order to prescribe appropriate countermeasures. The methods involve data from experiments on humans, animals, and cadavers, and predictions using biodynamic models and manikins. Criteria for estimating the occurrence of health effects and injury are summarized, together with methods for mitigating the effects of potentially harmful exposures. There is an extensive literature on the effects of vibration, shocks, and impacts on humans (Griffin, 1990; von Gierke et al., 2002; von Gierke, 1997; Nahum et al., 1993).

10.1.1 Definitions and Characterization of Vibration, Mechanical Shock, and Impact

Vibration and Mechanical Shock. *Vibration* is a time-varying disturbance of a mechanical, or biological, system from an equilibrium condition for which the long-term average of the motion will tend to zero, and on which may be superimposed either translations or rotations, or both. The mechanical forces may be distributed, or concentrated over a small area of the body, and may be applied at an angle to the surface (e.g., tangential or normal). Vibration may contain random or deterministic components, or both; they may also vary with time (i.e., be nonstationary). Deterministic vibration may contain harmonically related components, or pure tones (with sinusoidal time dependence), and may form "shocks." A mechanical *shock* is a nonperiodic disturbance characterized by suddenness and severity with, for the human body, the maximum forces being reached within a few tenths of a second, and a total duration of up to about a second. An *impact* occurs when the body, or body part, collides with an object. When considering injury potential, the shape of the object in contact with, or impacting, the body is important, as is the posture. In addition, for hand tools, both the compressive (grip) and thrust (feed) forces employed to perform the manual task need to be considered.

Although vibration, shock, and impact may be expressed by the *displacement* of a reference point from its equilibrium position (after subtracting translational and rotational motion), they are more commonly described by the *velocity* or *acceleration,* which are the first and second time derivatives of the displacement.

Vibration Magnitude. The magnitude of vibration is characterized by second, and higher even-order mean values, as the net motion expressed by a simple, long-term time average will be zero. For an acceleration that varies with time t, as $a(t)$, the higher-order mean values are calculated from:

$$a_{\mathrm{RM}} = \left[\frac{1}{T}\int_0^T [a(t)]^m \, dt\right]^{1/r} \tag{10.1}$$

where the integration is performed for a time T, and m and r are constants describing the moment and root of the function. By far the most common metric used to express the magnitude of whole-body or hand-transmitted vibration is the *root mean square* (RMS) acceleration a_{RMS}, which is obtained from Eq. (10.1) with $m = r = 2$; i.e.,

$$a_{\mathrm{RMS}} = \left[\frac{1}{T}\int_0^T [a(t)]^2 \, dt\right]^{1/2} \tag{10.2}$$

Other metrics used to express the magnitude of vibration and shock include the *root mean quad* (RMQ) acceleration a_{RMQ}, with $m = r = 4$ (and higher even orders, such as the *root mean sex* (RMX) acceleration a_{RMX}, with $m = r = 6$).

The RMS value of a continuous random vibration with a gaussian distribution corresponds to the magnitude of the 68th percentile of the amplitudes in the waveform. The higher-order means correspond more closely to the peak values of the waveform, with the RMQ corresponding to the 81st percentile and the RMX to the 88th percentile of this amplitude distribution. The relationships between these metrics depend on the amplitude distribution of the waveform, wherein they find application to characterize the magnitude of shocks entering, and objects impacting, the body. This can be inferred from the following example, where the RMS value corresponds to 0.707 of the amplitude of a sinusoidal waveform, while the RMQ value corresponds to 0.7825 of the amplitude.

EXAMPLE 10.1 *Calculate the RMS and RMQ accelerations of a pure-tone (single-frequency) vibration of amplitude A and angular frequency ω.*

Answer: *The time history (i.e., waveform) of a pure-tone vibration of amplitude A can be expressed as* $a(t) = A \sin \omega t$, *so that, from Eq. 10.(2):*

$$a_{\mathrm{RMS}} = \left[\frac{1}{T}\int_0^T [A \sin(\omega t)]^2 \, dt\right]^{1/2} \qquad \text{or} \qquad a_{\mathrm{RMS}} = \left[\frac{A^2}{2T}\int_0^T [1 - \cos(2\omega t)] \, dt\right]^{1/2}$$

Let us integrate for one period of the waveform, so that $T = 2\pi/\omega$ *(any complete number of periods will give the same result). Then:*

$$a_{\mathrm{RMS}} = \frac{A}{\sqrt{2}} = 0.707A$$

From Eq. (10.1):

$$a_{\mathrm{RMQ}} = \left[\frac{1}{T}\int_0^T [A \sin(\omega t)]^4 \, dt\right]^{1/4} \quad \text{or} \quad a_{\mathrm{RMQ}} = \left[\frac{A^4}{4T}\int_0^T [1 - \cos(2\omega t)]^2 \, dt\right]^{1/4}$$

$$a_{\mathrm{RMQ}} = \left[\frac{A^4}{4T}\int_0^T \left[\frac{3}{2} - 2\cos(2\omega t) - \frac{1}{2}\cos(4\omega t)\right] dt\right]^{1/4}$$

Again, integrating for one period of the waveform:

$$a_{\mathrm{RMQ}} = \left[\frac{3A^4}{8}\right]^{1/4} = 0.7825A$$

Equinoxious Frequency Contours. Human response to vibration, shock, and impact depends on the frequency content of the stimulus, as well as the magnitude. This may be conveniently introduced electronically, by filtering the time history of the stimulus signal, and has led to the specification of vibration magnitudes at different frequencies with an equal probability of causing a given human response or injury, so defining an *equinoxious frequency contour*. The concept, while almost universally employed, is strictly only applicable to linear systems. The biomechanic and biodynamic responses of the human body to external forces and accelerations commonly depend nonlinearly on the magnitude of the stimulus, and so any equinoxious frequency contour can be expected to apply only to a limited range of vibration, shock, or impact magnitudes.

Equinoxious frequency contours may be estimated from epidemiological studies of health effects, or from the response of human subjects, animals, cadavers, or biodynamic models to the stimuli of interest. Human subjects cannot be subjected to injurious accelerations and forces for ethical reasons, and so little direct information is available from this source. Some information has been obtained from studies of accidents, though in most cases the input acceleration-time histories are poorly known.

Frequency Weighting. The inverse frequency contour (i.e., reciprocal) to an equinoxious contour should be applied to a stimulus containing many frequencies to produce an overall magnitude that appropriately combines the contributions from each frequency. The frequency weightings most commonly employed for whole-body and hand-transmitted vibration are shown in Fig. 10.1 (ISO 2631-1, 1997; ISO 5349-1, 2001). The range of frequencies is from 1 to 80 Hz for whole-body vibration, and from 8 to 1250 Hz for vibration entering the hand. A frequency weighting for shocks may also be derived from a biodynamic model (see "Dynamic Response Index (DRI)" in Sec. 10.3.1).

Vibration Exposure. Health disturbances and injuries are related to the magnitude of the stimulus, its frequency content, and its duration. A generalized expression for exposure may be written

$$E(a_w, T)_{m,r} = \left[\int_0^T [F(a_w(t))]^m \, dt\right]^{1/r} \tag{10.3}$$

where $E(a_w, T)_{m,r}$ is the exposure occurring during a time T to a stimulus function that has been frequency weighted to equate the hazard at different frequencies, $F(a_w(t))$. In general, $F(a_w(t))$ may be expected to be a nonlinear function of the frequency-weighted acceleration-time history $a_w(t)$.

Within this family of exposure functions, usually only those with *even* integer values of m are of interest. A commonly used function is the so-called *energy-equivalent* vibration exposure for which $F(a_w(t)) = a_w(t)$ and $m = r = 2$:

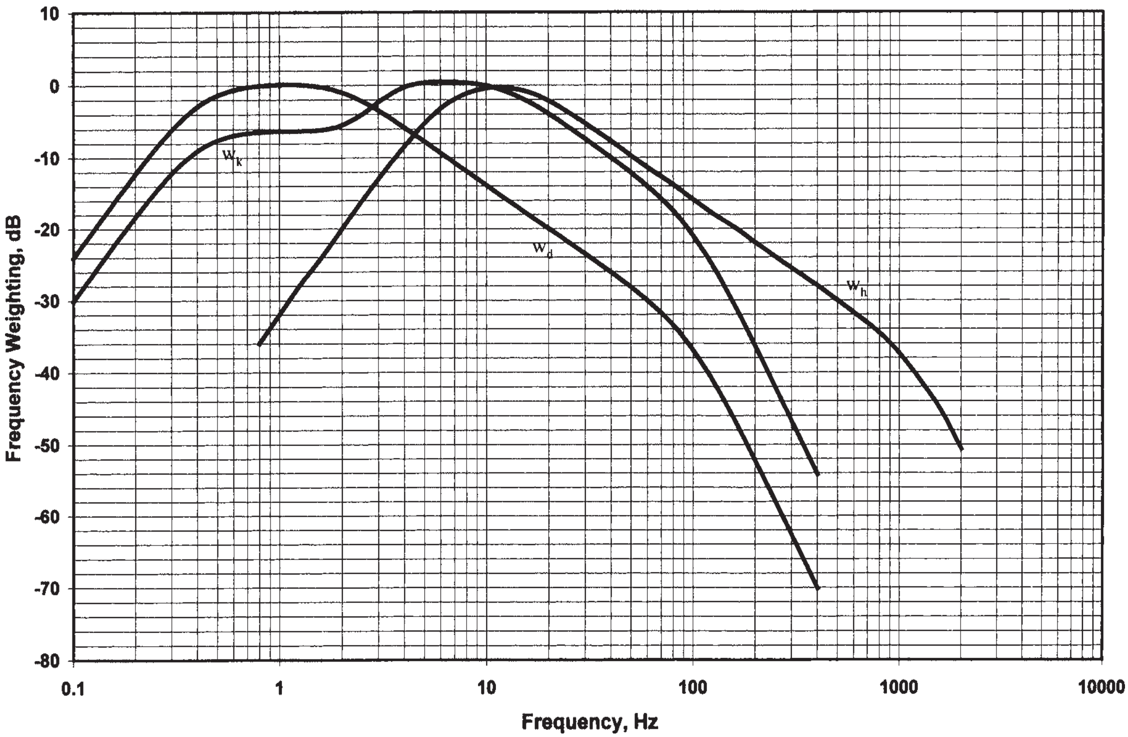

FIGURE 10.1 Frequency weightings for whole-body (W_k and W_d) and hand-transmitted (W_h) vibration. W_k and W_d are for the z direction and x and y directions, respectively, and are applicable to seated and standing persons (see Fig. 10.2). W_h is for all directions of vibration entering the hand. The filters are applied to acceleration-time histories $a(t)$. (*ISO 2631-1*, *1997*; *ISO 5349-1*, *2001*.)

$$E(a_w, T)_{2,2} = \left[\int_0^T [F(a_w(t))]^2 \, dt \right]^{1/2} \tag{10.4}$$

For an exposure continuing throughout a working day, $T = T_{(8)} = 28{,}800$ s, and Eq. (10.4) can be written [using Eq. (10.2)]:

$$E(a_w, T)_{2,2} = T_{(8)}^{1/2} \left[\frac{1}{T_{(8)}} \int_0^{T_{(8)}} [F(a_w(t))]^2 \, dt \right]^{1/2} = T_{(8)}^{1/2} a_{\mathrm{RMS}(8)} \tag{10.5}$$

where $a_{\mathrm{RMS}(8)}$ is the 8-hour, energy-equivalent, frequency-weighted RMS acceleration.

A second function, used for exposure to whole-body vibration, is the *vibration dose value*, VDV, for which $F(a_w(t)) = a_w(t)$ and $m = r = 4$. The function is thus:

$$\mathrm{VDV} = E(a_w, T)_{4,4} = \left[\int_0^T [a_w(t)]^4 \, dt \right]^{1/4} \tag{10.6}$$

which is more influenced by the large amplitudes in a fluctuating vibration than the energy-equivalent exposure. A related function, the *severity index* for which $F(a_w(t)) = a_w(t)$, $m = 2.5$, and $r = 1$, is sometimes used for the assessment of head impact, though it cannot be applied to continuous acceleration-time histories owing to the value of m.

10.1.2 Human Response to Vibration, Mechanical Shock, and Impact

Mechanical damage can occur at large vibration magnitudes, which are usually associated with exposure to shocks, and to objects impacting the body (e.g., bone fracture, brain injury, organ hemorrhage, and tearing or crushing of soft tissues). At moderate magnitudes there can be physiological effects leading to chronic injury, such as to the spine, and disorders affecting the hands. At all magnitudes above the threshold for perception there can be behavioral responses ranging from discomfort to interference with tasks involving visual or manual activities.

Injury from Vibration. *Whole-Body Vibration.* Small animals (e.g., mice and dogs) have been killed by intense vibration lasting only a few minutes (see Griffin, 1990). The internal injuries observed on postmortem examination (commonly heart and lung damage, and gastro-intestinal bleeding) are consistent with the organs beating against each other and the rib cage, and suggest a *resonance* motion of the heart, and lungs, on their suspensions. In man, these organ suspension resonances are at frequencies between 3 and 8 Hz.

Chronic exposure to whole-body vibration may result in an increased risk of low back pain, sciatic pain, and prolapsed or herniated lumbar disks compared to control groups not exposed to vibration. These injuries occur predominantly in crane operators, tractor drivers, and drivers in the transportation industry (Bovenzi and Hulshof, 1998). However, it is difficult to differentiate between the roles of whole-body vibration and ergonomic risk factors, such as posture, in the development of these disorders.

Hand-Transmitted Vibration. Chronic injuries may be produced when the hand is exposed to vibration. Symptoms of numbness or paresthesia in the fingers or hands are common. Reduced grip strength and muscle weakness may also be experienced, and episodic finger blanching, often called colloquially "white fingers," "white hand," or "dead hand," may occur in occupational groups (e.g., operators of pneumatic drills, grinders, chipping hammers, riveting guns, and chain saws). The blood vessel, nerve, and muscle disorders associated with regular use of hand-held power tools are termed the *hand-arm vibration syndrome* (HAVS) (Pelmear et al., 1998). An exposure-response relationship has been derived for the onset of finger blanching (Brammer, 1986). Attention has also recently been drawn to the influence of vibration on task performance and on the manual control of objects (Martin et al., 2001).

Repeated flexing of the wrist can injure the tendons, tendon sheaths, muscles, ligaments, joints and nerves of the hand and forearm. These *repetitive strain injuries* commonly occur in occupations involving repeated hand-wrist deviations (e.g., keyboard and computer operators), and frequently involve nerve compression at the wrist (e.g., *carpal tunnel syndrome*) (Cherniack, 1999).

Injury from Shock and Impact. Physiological responses to shocks and objects impacting the body include those discussed for whole-body vibration. For small contact areas, the injuries are often related to the elastic and tensile limits of tissue (Haut, 1993; von Gierke et al., 2002). The responses are critically dependent on the magnitude, direction, and time history of the acceleration and forces entering the body, the posture, and on the nature of any body supports or restraints (e.g., seat belt or helmet).

Vertical Shocks. Exposure to single shocks applied to a seated person directed from the seat pan toward the head ("headward") has been studied in connection with the development of aircraft ejection seats, from which the conditions for spinal injury and vertebral fractures have been documented (Anon., 1950; Eiband, 1959). Exposure to intense *repeated* vertical shocks is experienced in some off-the-road vehicles and high-performance military aircraft, where spinal injury has also been reported. A headward shock with acceleration in excess of $g = 9.81$ m/s^2 (the acceleration of gravity) is likely to be accompanied by a downward ("tailward") impact, when the mass of the torso returns to being supported by the seat.

Horizontal Shocks. Exposure to rapid decelerations in the horizontal direction has been extensively studied in connection with motor vehicle and aircraft crashes ("spineward" deceleration). Accident statistics indicate that serious injuries to the occupants of motor vehicles involved in frontal collisions are most commonly to the head, neck, and torso, including the abdomen (AGARD-AR-330, 1997).

Injuries to the head usually involve diffuse or focal brain lesions either with or, commonly, without skull fracture. The former consists of brain swelling, concussion, and *diffuse axonal injury*, that is, mechanical disruption of nerve fibers; the latter consists of localized internal bleeding and contusions (coup and contrecoup).

The most common neck injury is caused by rearward flexion and forward extension ("whiplash"), which may result in dislocation or fracture of the cervical vertebrae, and compression of the spinal cord.

10.2 *PHYSICAL MEASUREMENTS*

The complexity of a living organism, and its ability to modify its mechanical properties (e.g., in response to mechanical or physiological demands or muscle tension), necessitates the careful design of experiments. There is a large variability in response between individuals. Also, the direct attachment of vibration and shock sensors to soft tissues produces a mechanical load that influences tissue motion. With appropriate measurement methods and instrumentation (ISO 8041, 1990), mechanical responses to vibration can be determined for tissues, body segments, and the whole body.

10.2.1 Methods and Apparatus

Tissue Vibration. Noncontact methods are preferred for measuring the surface motion of tissues. Laser vibrometers are commercially available with sufficient bandwidth and resolution for most studies. A direct mass load on the skin, together with the skin's elasticity, forms a mechanical low-pass filter (see "Simple Lumped Models" in Sec. 10.3.1). If a device is to be mounted directly on the skin, it must be of low weight (e.g., <3 g) and possess a comparatively large attachment area (e.g., >5 cm^2), in order for vibration to be recorded without attenuation of the motion at 80 Hz. An upper frequency limit of 200 Hz is theoretically achievable (-3dB) with a transducer and skin mount weighing 3 g and an attachment area of 1.8 cm^2.

The measurement of hand-transmitted vibration requires a bandwidth extending to at least 1250 Hz, which is unobtainable by localized, skin-mounted transducers. Attempts to strap the transducer to a bony prominence (e.g., by a "watch strap" at the wrist) have demonstrated that the upper frequency limit for this method of measurement is about 200 Hz (Boileau et al., 1992). A distributed sensor, such as a pressure-sensitive lightweight film, would permit the motion of a large skin area to be monitored with acceptable mass loading, and could respond to vibration at higher frequencies.

Interface between Body and Vibrating Surface. Devices have been developed to measure the motion of the interface between the skin and a source of vibration in contact with the body, such as a vehicle seat pan or tool handle. The former consists of a flexible disk of rubberlike material thickened at the center, where an accelerometer is mounted. The dimensions have been standardized (ISO 7096, 1982). Attempts have been made to design transducer mounts for the palm of the hand to measure the vibration entering the hand from tool handles (see, for example, ISO 10819, 1997), and also the static compressive force (i.e., the combined grip and thrust forces exerted by the hand on the tool handle). The frequency response of one such device extends to more than 1 kHz (Gillmeister et al., 2001).

Structures in Contact with the Body. The vibration of structures in contact with the body, such as seats and tool handles, is conveniently measured by accelerometers rigidly attached to a structural element. Accelerometers are designed to respond to the vibration in a given direction and may be grouped to record simultaneously accelerations in three orthogonal directions. They are commercially available in a wide range of sizes and sensitivities, and may be attached by screws or adhesives to the structure of interest.

Orientation of Sensors. Tissue-, interface-, and structure-mounted sensors may be aligned as closely as possible with the axes of a *biodynamic coordinate system* (see, for example, Fig. 10.2), even though these may have inaccessible origins that are anatomical sites within the body. In practice, sensors are commonly oriented to record the component accelerations defined by the *basicentric coordinate systems* shown in Fig. 10.2 (ISO 2631–1, 1997; ISO 5349–1, 2001), which have origins at the interface between the body and the vibrating surface. The location of accelerometers to record the handle vibration of specific power tools is described in an international standard (ISO 5349-2, 2001).

Errors in Shock and Impact Measurement. Care must be taken when accelerometers employing piezoelectric elements are used to measure large-magnitude shocks and impacts, as they are subject to internal crystalline changes that result in dc shifts in the output voltage. Results containing such shifts should be considered erroneous. This limitation of piezoelectric transducers may be overcome by mounting the sensor on a mechanical low-pass filter (see "Simple Lumped Models" in Sec. 10.3.1), which is, in turn, attached to the structure of interest. Such filters possess a resilient element that serves to reduce the transmission of vibration at high frequencies. The filter cutoff frequency is selected to be above the maximum vibration frequency of interest but below the internal mechanical resonance frequency of the accelerometer.

Data Recording. The signal produced by vibration, shock, and impact sensors is first conditioned to remove bias voltages or other signals required for the device to function, and then amplified and buffered for output to a data recording system. The output may be stored on high-quality magnetic tape (e.g., a DAT recorder), or by a digital data acquisition system. The latter should possess low-pass, antialiasing filters (with cutoff frequency typically one-half the sampling frequency), and an analog-to-digital (A/D) converter with sufficient dynamic range (commonly 12 or 16 bits). The data acquisition system should be capable of recording time histories at sampling frequencies of at least 2500 Hz for hand-transmitted vibration, or 160 Hz for whole-body vibration, *per component and measurement site* (e.g., palm and wrist, or seat surface and seat back).

10.2.2 Small-Amplitude Response of the Human Body

Tissue Properties. The properties of human tissues when the body is considered a linear, passive mechanical system are summarized in Table 10.1 (von Gierke et al., 2002; Goldstein et al., 1993). The values shown for soft tissues are typical of muscle tissue, while those for bone depend on the structure of the specific bone. Cortical bone is the dominant constituent of the long bones (e.g., femur, tibia), while trabecular bone, which is more elastic and energy absorbent, is the dominant constituent of the vertebrae. The shear viscosity and bulk elasticity of soft tissue are from a model for the response in vivo of a human thigh to the vibration of a small-diameter piston (von Gierke et al., 1952)

The nonlinear mechanical properties of biological tissues have been studied extensively in vitro, including deviations from Hooke's law (Fung, 1993; Haut, 1993).

Mechanical Impedance of Muscle Tissue. The (input) *mechanical impedance* is the complex ratio between the dynamic force applied to the body and the velocity at the interface where vibration enters the body. The real and imaginary parts of the mechanical impedance of human muscle in vivo are shown as a function of frequency in Fig. 10.3 (von Gierke et al., 1952). In this diagram the measured resistance (open circles) and reactance (diamonds) are compared with the predictions of a model, from which some tissue properties may be derived (see Table 10.1). It should be noted that the mechanical stiffness and resistance of soft tissues approximately triple in magnitude when the

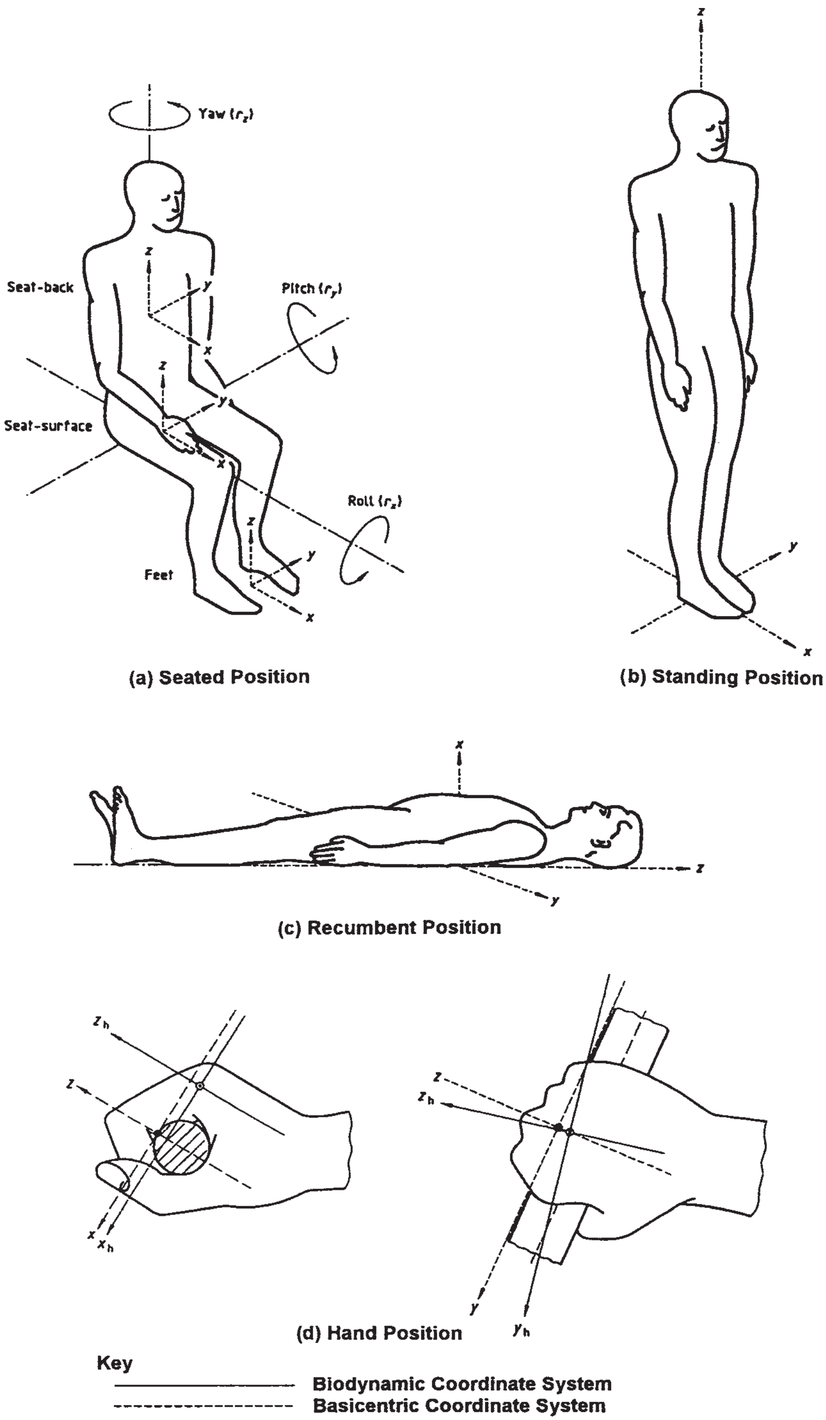

FIGURE 10.2 Basicentric coordinate axes for translational (x, y, and z) and rotational (r_x, r_y, and r_z) whole-body vibration, and basicentric (filled circles and dashed lines) and biodynamic (open circles and continuous lines) axes for hand-transmitted vibration. The biodynamic coordinates axes for the hand are x_h, y_h, and z_h. (*ISO 2631-1, 1997; ISO 5349-1, 2001*.)

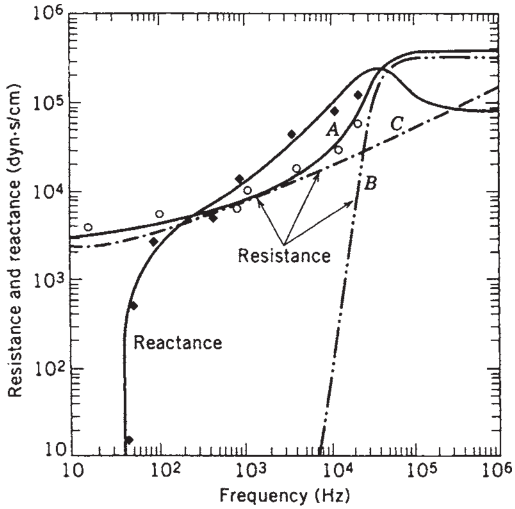

FIGURE 10.3 Mechanical resistance and reactance of soft thigh tissue (2 cm in diameter) in vivo from 10 Hz to 1 MHz. The measured values (open circles—resistance; diamonds—reactance) are compared with the calculated resistance and reactance of a 2-cm-diameter sphere vibrating in a viscous, elastic compressible medium with properties similar to soft human tissue (continuous lines, curves A). The resistance is also shown for the sphere vibrating in a frictionless compressible fluid (acoustic compression wave, curve B) and an incompressible viscous fluid (curve C). (*von Gierke et al., 1952.*)

TABLE 10.1 Typical Physical Properties of Human Tissues at Frequencies Less than 100 kHz

Property	Soft tissues	Bone (wet)	Bone (dry)
Density, kg/m^3	$1–1.2 \times 10^3$	$1.9–2.3 \times 10^3$	1.9×10^3
Young's modulus, Pa	7.5×10^3	$1.6–2.3 \times 10^{10}$	1.8×10^{10}
Shear modulus,* Pa	2.5×10^3†	$2.9–3.4 \times 10^9$	7.1×10^9
Bulk modulus, Pa	2.6×10^9†		1.3×10^{10}
Shear viscosity, Pa · s	15†		
Sound velocity, m/s	$1.5–1.6 \times 10^3$	3.4×10^3	
Acoustic impedance, Pa · s/m	1.7×10^6	6×10^6	6×10^6
Tensile strength, Pa			
Cortical bone		$1.3–1.6 \times 10^8$	1.8×10^8
Compressive strength, Pa			
Cortical bone		$1.5–2.1 \times 10^8$	
Trabecular bone (vertebrae)		$0.4–7.7 \times 10^6$	
Shear strength, Pa			
Cortical bone		$7.0–8.1 \times 10^7$	

*Lamé constant.
†From soft tissue model (von Gierke et al., 1952).
Source: After von Gierke et al., 2002; and Goldstein et al., 1993.

static compression of the surface increases by a factor of three. The relationship, however, is not linear.

Apparent Mass of Seated Persons. The *apparent mass* is often used to describe the response of the body at the point of stimulation rather than the mechanical impedance, and is the complex ratio between the dynamic force applied to the body and the acceleration at the interface where vibration enters the body. It is commonly expressed as a function of frequency, and is equal to the static weight of a subject in the limiting case of zero frequency when the legs are supported to move in unison with the torso. The influence of posture, muscle tension, and stimulus magnitude on the apparent mass of seated persons, in the vertical direction, is shown for four subjects in Fig. 10.4 (Fairley et al., 1989).

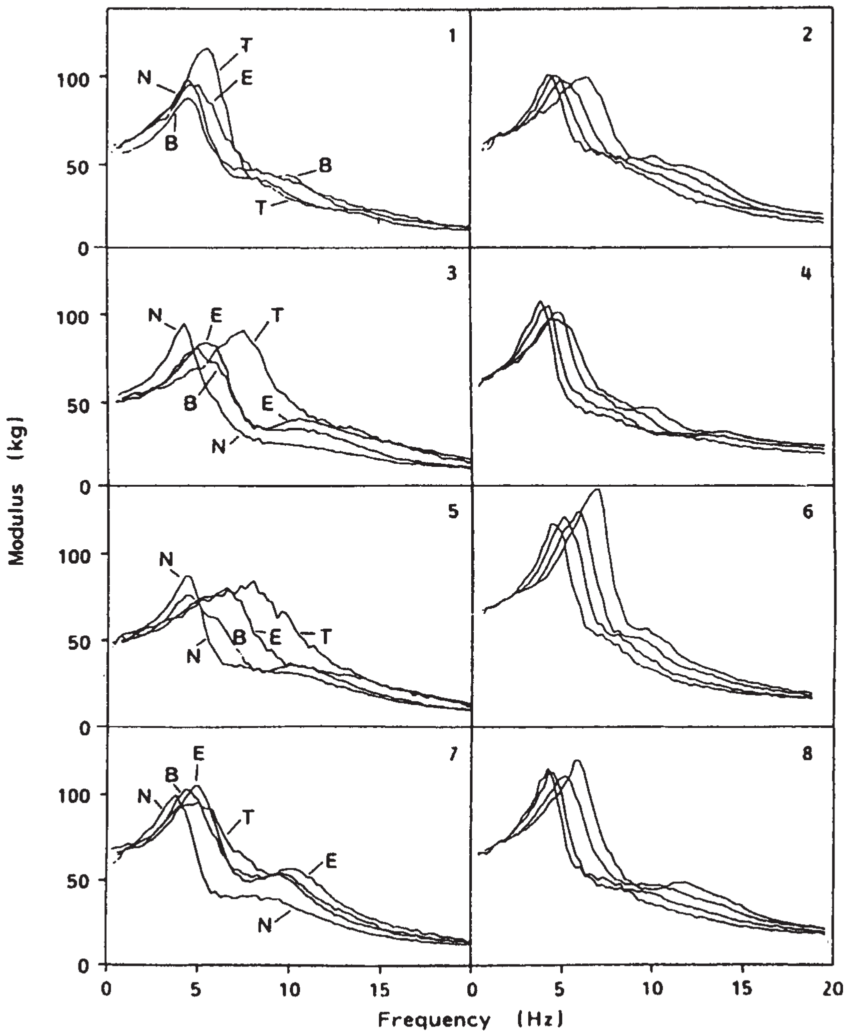

FIGURE 10.4 Effect of posture (N—"normal"; E—erect; B—with backrest), muscle tension (T—tensed muscles), and stimulus magnitude (0.25, 0.5, 1.0, and 2.0 m/s^2) on the apparent mass of a seated person for four subjects (see text for explanation). (*Fairley et al., 1989*.)

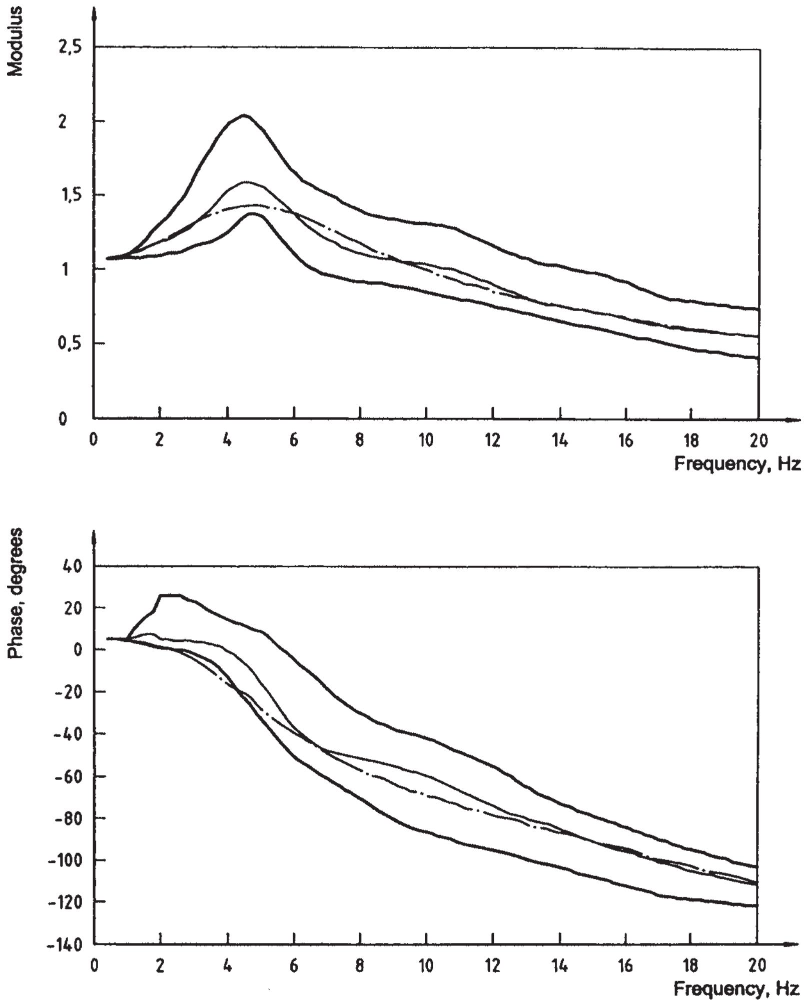

FIGURE 10.5 Idealized values for the modulus and phase of the seat-to-head transmissibility of seated persons subjected to vertical vibration. The envelopes of the maximum and minimum mean values of studies included in the analysis are shown by thick continuous lines, and the mean of all data sets is shown by the thin line. The response of a biodynamic model (see text and Fig. 10.8) is plotted as a dash-dot line. (*ISO 5982, 2001*.)

The column of graphs to the left of the diagram shows the modulus of the apparent mass measured with a comfortable, "normal," upright posture and muscle tension (labeled N), with this posture but an erect torso and the shoulders held back (E), with all muscles in the upper body tensed (T), and, finally, with the subject leaning backward to rest against a rigid backrest (B). The largest variation in apparent mass between these conditions was associated with tensing the back muscles, which clearly increased the frequency of the characteristic peak in the response (at around 5 Hz). In some subjects the frequency of this peak could be changed by a factor of 2 by muscle tension. A variation in the apparent mass could also be induced by changing the stimulus magnitude, as is shown for four RMS accelerations (0.25, 0.5, 1.0, and 2.0 ms^{-2}) to the right of Fig. 10.4. Within this range of stimulus magnitudes, the frequency of the characteristic peak in the apparent mass was found to decrease with increasing stimulus magnitude, for each subject.

Seat-to-Head Transmissibility. The *transmissibility* expresses the response of one part of a mechanical system (e.g., the head or hand) to steady-state forced vibration of another part of the system (e.g., the buttocks), and is commonly expressed as a function of frequency. A synthesis of measured values for the seat-to-head transmissibility of seated persons has been performed for vibration in the vertical direction, to define the *idealized* transmissibility. The idealized transmissibility attempts to account for the sometimes large and unexplained variations in the results from different experimental studies conducted under nominally equivalent conditions. The results of one such analysis are shown by the continuous lines in Fig. 10.5 (ISO 5982, 2001). It can be seen by comparing Figs. 10.4 and 10.5 that the characteristic peak of the apparent mass remains in the modulus of the idealized transmissibility.

Mechanical Impedance of the Hand-Arm System. The idealized mechanical input impedance of the hand-arm system when the hand is gripping a cylindrical or oval handle has been derived for the three directions of the basicentric coordinate system shown in Fig. 10.2 (ISO 10068, 1998). The transmissibility of vibration through the hand-arm system has not been measured with noncontact or lightweight transducers satisfying the conditions described in Sec. 10.2.1. However, it has been demonstrated that the transmissibility from the palm to the wrist when a hand grips a vibrating handle is unity at frequencies up to 150 Hz (Boileau et al., 1992).

10.3 MODELS AND HUMAN SURROGATES

Knowledge of tolerable limits for human exposure to vibration, shock, and impact is essential for maintaining health and performance in the many environments in which man is subjected to dynamic forces and accelerations. As already noted, humans cannot be subjected to injurious stimuli for ethical reasons, and so little direct information is available from this source. In these circumstances, the simulation of human response to potentially life-threatening dynamic forces and accelerations is desirable, and is commonly undertaken using *biodynamic models*, and *anthropometric* or *anthropomorphic manikins*. They are also used in the development of vehicle seats and, in the case of hand-arm models, of powered hand-held tools.

10.3.1 Biodynamic Models

Simple Lumped Models. At frequencies up to several hundred hertz, the biodynamic response of the human body can be represented theoretically by point masses, springs, and dampers, which constitute the elements of *lumped* biodynamic models. The simplest one-dimensional model consists of a mass supported by a spring and damper, as sketched in Fig. 10.6, where the system is excited at its base. The equation of motion of a mass m when a spring with stiffness k and damper with resistance proportional to velocity, c, are base driven with a displacement $x_0(t)$ is:

$$ma_1(t) + c(\dot{x}_1(t) - \dot{x}_0(t)) + k(x_1(t) - x_0(t)) = 0 \qquad (10.7)$$

where the displacement of the mass is $x_1(t)$, its acceleration is $a_1(t)$, and differentiation with respect to time is shown by dots.

For this simple mechanical system, the apparent mass may be expressed as a function of frequency by (Griffin, 1990):

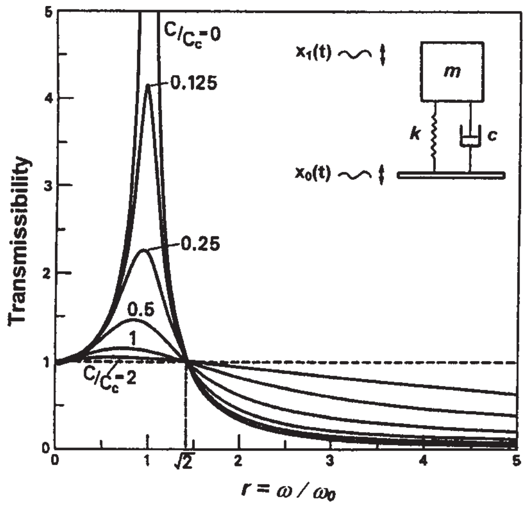

FIGURE 10.6 Single-degree-of-freedom, lumped-parameter biodynamic model. The mass m is supported by a spring with stiffness k and viscous damper with resistance c. The transmissibility of motion to the mass is shown as a function of the frequency ratio r (= ω/ω_0) when the base is subjected to a displacement $x_0(t)$. (*After Griffin, 1990.*)

$$M(\omega) = \frac{m(k + i\omega c)}{k - \omega^2 m + i\omega c} \tag{10.8}$$

where ω is the angular frequency (= $2\pi f$), $i = (-1)^{1/2}$, and the transmissibility from the base to the mass is

$$M(\omega) = \frac{m(k + i\omega c)}{k - \omega^2 m + i\omega c} \tag{10.9}$$

The modulus of the transmissibility may then be written

$$|H(\omega)| = \left[\frac{1 + (2\xi r)^2}{(1 - r^2)^2 + (2\xi r)^2}\right]^{1/2} \tag{10.10}$$

where r is the ratio of the angular excitation frequency to the angular *resonance frequency* of the system, ω/ω_0, and

$$\omega_0 = 2\pi f_0 = \left(\frac{k}{m}\right)^{1/2} \tag{10.11}$$

In Eq. (10.10), the damping is expressed in terms of the damping ratio $\xi = c/c_c$, where c_c is the critical viscous damping coefficient [= $2(mk)^{1/2}$]. The transmissibility of the system is plotted as a function of the ratio of the angular excitation frequency to the natural (resonance) frequency in Fig. 10.6. It can be seen from the diagram that, at excitation frequencies less than the resonance frequency (i.e., r <<1), the motion of the mass is the same as that of the base. At frequencies greater than the resonance frequency, however, the motion of the mass becomes progressively less than that of the base. At angular excitation frequencies close to the resonance frequency ω_0, the motion of the mass exceeds that of the base. This response is that of a *low-pass mechanical filter*.

Dynamic Response Index (DRI). The response of the spine to shocks in the vertical (headward) direction has long been modeled in connection with the development of ejection seats for escape from high-performance aircraft. The simple model shown in Fig. 10.6 has been used to simulate the maximum stress within the vertebral column by calculating the maximum dynamic deflection of the spring, $|x_1(t) - x_0(t)|_{max}$, 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 $(\omega_0)^2|x_1(t) - x_0(t)|_{max}/g$, where the natural frequency is 52.9 rad/s (i.e., f_0 = 8.42 Hz), the damping ratio is 0.224, and g is the acceleration of gravity (9.81 ms^{-2}). The success of the model has led to its adoption for specifying ejection seat performance and its extension to a metric for exposure to repeated shocks and for ride comfort in high-speed boats (Allen, 1978; Brinkley, 1990; Payne 1976).

Whole-Body Apparent Mass for Vertical (z-Direction) Vibration. The apparent mass of the seated human body may be described by a variant of the simple biodynamic model of Fig. 10.6. A satisfactory prediction is obtained with the addition of a second mass, m_0, on the seat platform to represent the mass of the torso and legs that does not move relative to the seat (i.e., to the base of the

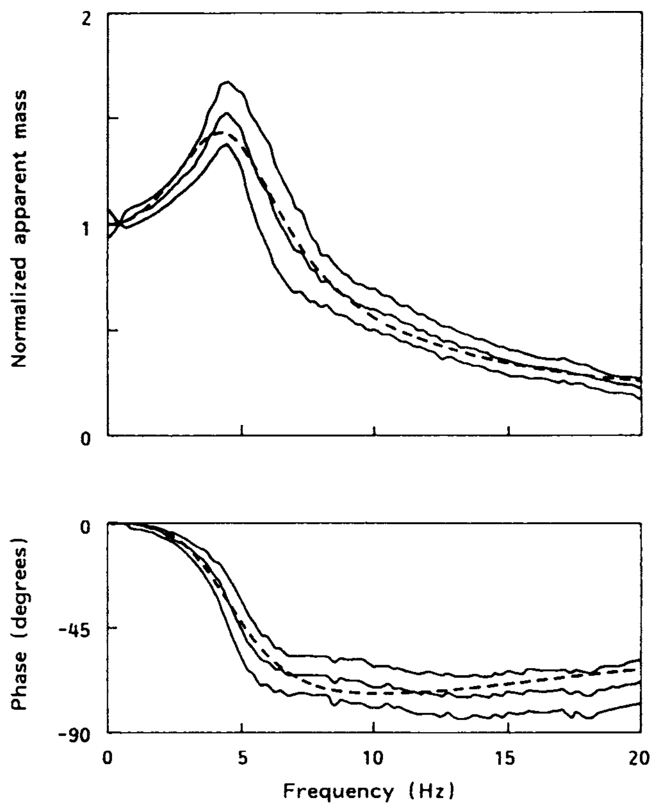

FIGURE 10.7 Comparison between predicted and observed apparent mass. The mean, normalized apparent masses of 60 subjects, ± 1 standard deviation, are shown by the continuous lines, and the predictions of a single-degree-of-freedom, lumped parameter biodynamic model by the dashed line. (*Fairley et al., 1989*.)

model). The apparent mass measured on 60 subjects (24 men, 24 women, and 12 children) can be well represented by this model when the legs are supported to move in phase with the seat, and the data are normalized for the different weights of individuals using the value of apparent mass recorded at 0.5 Hz (Fairley and Griffin, 1989). A comparison of the observed values for the magnitude, and phase, of the normalized apparent mass (continuous lines) and the predicted values (dashed lines) is shown in Fig. 10.7. To obtain this agreement, the natural angular frequency of the model is 31.4 rad/s (i.e., f_0 = 5 Hz), and the damping ratio is 0.475. These values differ considerably from those of the DRI model (see above), reflecting the stimulus magnitude dependent, nonlinear response of the human body to vibration.

Whole-Body Impedance, Apparent Mass, and Transmissibility for Vertical (z-direction) Vibration. A model has been developed to predict idealized values for the input impedance, apparent mass, *and* transmissibility of the seated human body when subjected to vertical vibration. The model, shown in Fig. 10.8, comprises elements forming three of the models sketched in Fig. 10.6 (ISO 5982, 2001). It contains elements similar to those used to predict apparent mass on the left-hand side of the diagram with, in addition, two basic models in series to the right of the diagram (i.e., one on top of the other), in order to represent "head" motion (i.e., motion of mass m_2). The model predictions for the transmissibility from the seat to the head are shown by the dash-dot lines in Fig. 10.5, and should be compared with the target mean values (the thin continuous lines). While the model meets the considerable challenge of predicting three biodynamic responses with one set of model values, the parameters of the model do not possess precise anatomical correlates, a common failing of simple lumped models, and the predictions are applicable only to a limited range of stimulus magnitudes.

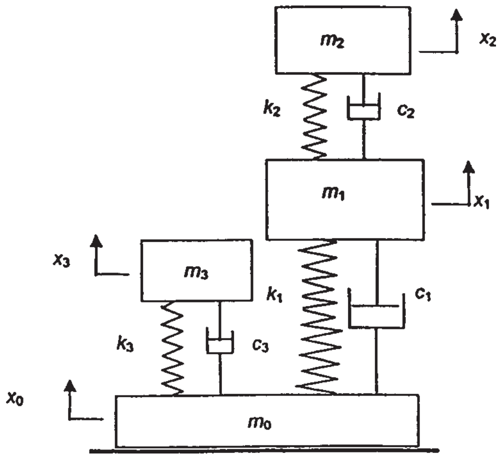

FIGURE 10.8 Three-degree-of-freedom, lumped-parameter biodynamic model of the seated human body for estimating mechanical impedance, apparent mass, and transmissibility, for vertical vibration. The model is driven at the base (X_0), and the transmissibility is calculated to the "head," mass m_2. (*ISO 5982, 2001.*)

Impedance Model of the Hand-Arm System. A 4-degree-of-freedom model consisting of four of the models sketched in Fig. 10.6, arranged in series (i.e., on top of each other), has been used to predict idealized values for the input impedance to the hand (ISO 10068, 1998). As is the case for the lumped whole-body models, the parameters of the hand-arm model do not possess direct anatomical correlates.

Rigid, Multielement, and Finite-Element (FE) Models. More complex models have been developed to incorporate more realistic descriptions of individual body parts and to predict the motion of one body part relative to another. One of the first such models consisted of a seven-segment numerical model for describing the motion of a vehicle occupant in a collision. The segments consisted of the head and neck, upper torso, lower torso, thighs, legs, upper arms, and forearms and hands. Of the many multielement models that have since been developed, the *Articulated Total Body* (ATB) model and the *MAthematical DYnamical MOdel* (MADYMO) are the most widely used. Each employs a combination of rigid bodies, joints, springs, and dampers to represent the human, or in some cases manikin, and sets up and solves numerically the equations of motion for the system. The applied forces and torques on these components are established by using different routines for contact with external surfaces and for joint resistance, the effect of gravity, and restraining the body (e.g., by a seat belt).

ATB and MADYMO Models. The ATB and MADYMO computer models commonly employ 15 ellipsoidal segments with masses and moments of inertia appropriate for the body parts they represent, as determined from anthropomorphic data for adults and children. The connections between these segments are flexible, and possess elastic and resistive properties based on measurements on human joints. The environment to be simulated may include contact surfaces (e.g., seats and vehicle dashboard), seat belts, and air bags. The models may include advanced harness systems and wind forces to model seat ejection from aircraft and can also describe aircraft crash. An example of the use of these models to predict human response in crashlike situations is shown in Fig. 10.9 (von Gierke, 1997). In this diagram, the motion of an unrestrained child standing on the front seat of an automobile is shown in response to panic breaking. The output of the model has been calculated at the onset of breaking and after 500 and 600 ms. It can be seen from the diagram that the model predicts the child will slide down and forward along the seat until its head impacts the dashboard of the vehicle.

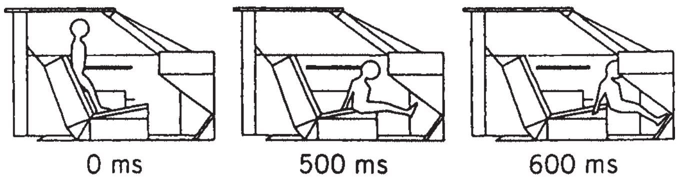

FIGURE 10.9 Articulated total body (ATB) model prediction for the response of an unrestrained standing child to panic braking in an automobile at 500 and 600 ms after commencing braking, (*von Gierke, 1997.*)

Finite-Element (FE) Models. The detailed response of selected parts of the body and the environment being simulated (e.g., head and neck, spine, and vehicle seating) have been modeled by using finite elements (FEs). In the MADYMO model, the FEs can interact with the multibody model elements. Examples of human body subsystems that have been modeled with FEs include the spine, to predict the injury potential of vertebral compression and torsional loads, and the head and neck, to predict rotation of the head and neck loads during rapid horizontal deceleration (see Fig. 10.12) (RTO-MP-22, 1999).

10.3.2 Anthropometric Manikins

Mechanically constructed manikins, or dummies, are used extensively in motor vehicle crash testing and for evaluating aircraft escape systems and seating. Several have been developed for these purposes (AGARD AR-330, 1997), some of which are commercially available.

Hybid III Manikin. The Hybrid III manikin, shown in Fig. 10.10, was originally developed by General Motors for motor vehicle crash testing, and has since become the de facto standard for

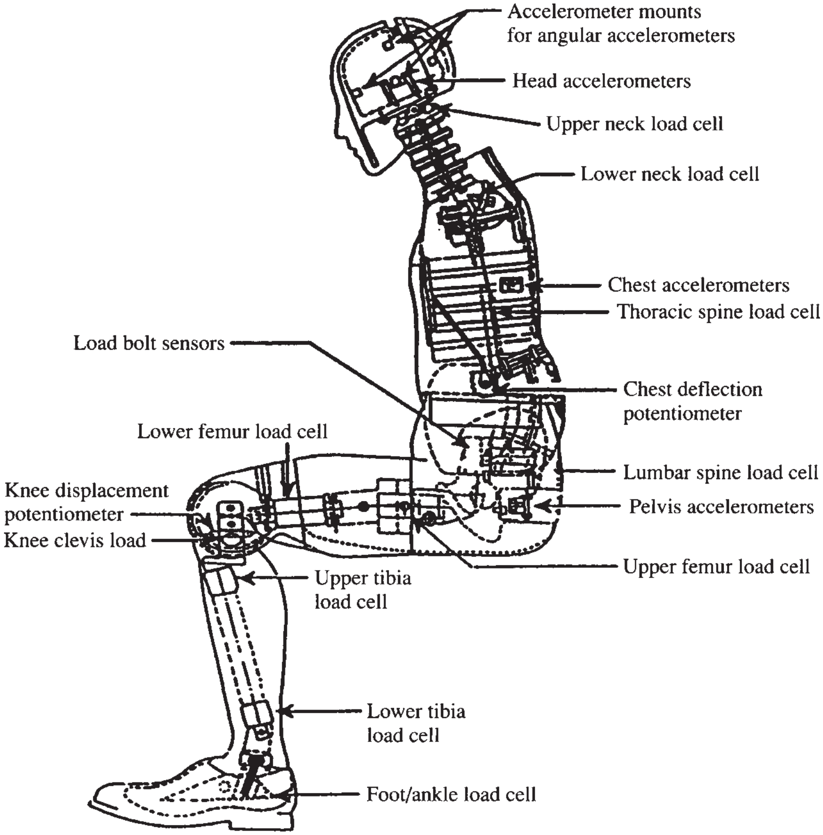

FIGURE 10.10 Hybrid III anthropometric dummy designed for use in motor-vehicle frontal crash tests, showing elements of construction and sensors. (*AGARD-AR-330, 1997.*)

simulating the response of motor vehicle occupants to frontal collisions and for tests of occupant safety restraint systems. The manikin approximates the size, shape, and mass of the 50th percentile North American adult male, and consists of metal parts to provide structural strength and define the overall geometry. This "skeleton" is covered with foam and an external vinyl skin to produce the desired shape. The manikin possesses a rubber lumbar spine, curved to mimic a sitting posture. The head, neck, chest, and leg responses are designed to simulate the following human responses during rapid deceleration: head acceleration resulting from forehead and side-of-the-head impacts; fore-and-aft, and lateral, bending of the neck; deflection of the chest to distributed forces on the sternum; and impacts to the knee (Mertz, 1993). The instrumentation required to record these responses together with local axial compressional loads is shown in Fig. 10.10. Hybrid III dummies are now available for small (fifth percentile) adult females, and large (95th percentile) adult males, as well as for infants and children. A related *side impact dummy* (SID) has been developed for the U.S. National Highway Traffic Safety Administration (NHTSA), for crash tests involving impacts on the sides of motor vehicles.

ADAM. ADAM (Advanced Dynamic Anthropomorphic Manikin) is a fully instrumented manikin primarily used in the development of aircraft ejection systems. Its overall design is conceptually similar to that of the Hybrid III dummy (see Fig. 10.10), in that ADAM replicates human body segments, surface contours, and weight. In addition to a metal skeleton, the manikin posseses a sandwich skin construction of sheet vinyl separated by foamed vinyl to mimic the response of human soft tissue. ADAM also attempts to replicate human joint motion and the response of the spine to vertical accelerations for both small-amplitude vibration and large impacts. The spine consists of a mechanical spring-damper system, which is mounted within the torso.

10.3.3 Biodynamic Fidelity of Human Surrogates

Animals, cadavers, manikins, and computer models have been used to predict human responses to potentially injurious or life-threatening stimuli. To evaluate the biofidelity of the surrogate or model, it is necessary to identify the response characteristics that are most relevant. For biodynamic responses, the time histories of the acceleration, velocity, displacement, and forces provide the most meaningful comparisons, though point-by-point comparisons can be misleading if the system response to the stimulus of interest is extremely nonlinear. In these circumstances, evaluating peak values in the time history, impulses calculated from the acceleration or contact forces, or energy absorption may be more appropriate.

Manikins and Computer Models. The current state of the art is illustrated in Figs. 10.11 and 10.12, where several biodynamic parameters of the response of the human head and neck to rapid horizontal

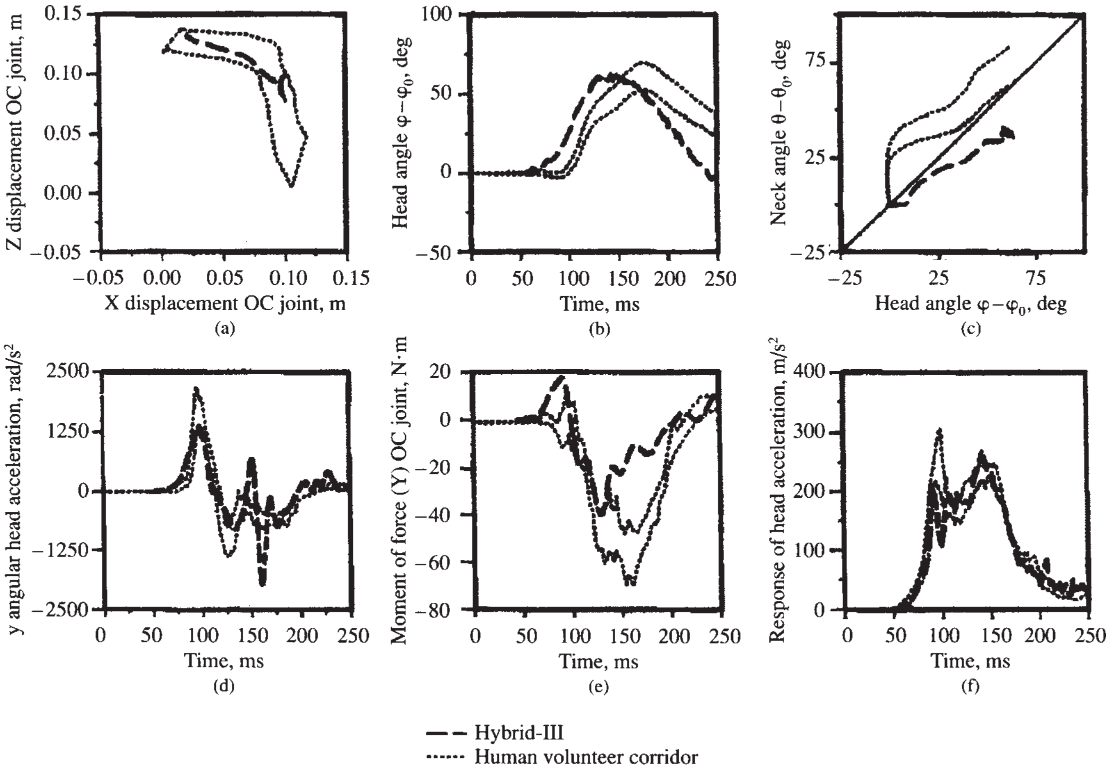

FIGURE 10.11 Response of human volunteers and the Hybrid III head and neck to 15 *g* spineward deceleration. The range of responses from human subjects is shown by the dotted lines, and the response of the manikin by the dashed lines (see text for explanation of the motions plotted). (*RTO-MP-20, 1999*.)

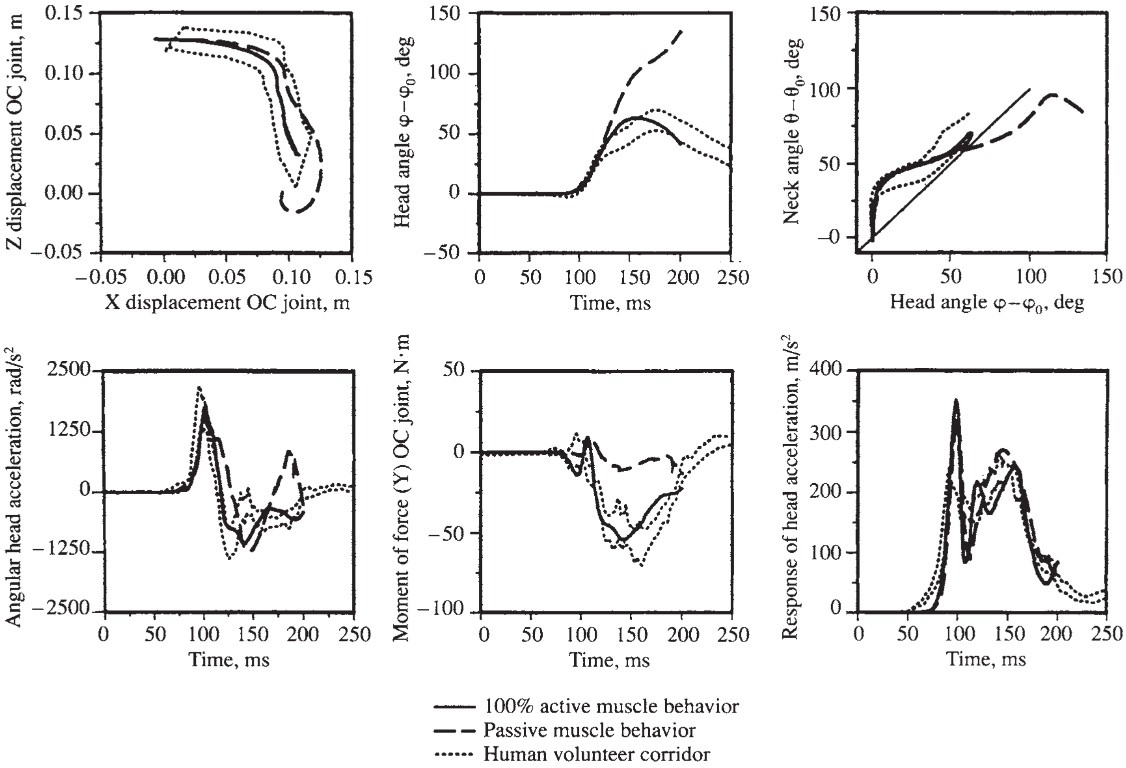

FIGURE 10.12 Response of human volunteers and the 3D MADYMO model of the head and neck to spine ward decelerations. The range of responses from human subjects is shown by the dotted lines, and the response of the model with passive and active muscles is shown by the dashed and continuous lines, respectively (see text for explanation of the motions plotted). (*RTO-MP-20, 1999*.)

decelerations are compared. The comparisons are between the responses of human volunteers, the limits of which are shown by dotted lines, with those of the Hybrid III manikin at the same deceleration (Fig. 10.11), and of a three-dimensional head and neck for the MADYMO computer model (Fig. 10.12) (RTO-MP-20, 1999).

While the Hybrid III manikin can reproduce some human responses, the head and neck system does not introduce appropriate head rotation lag (see neck angle versus head angle in Fig. 10.11*c*) or torque at the occipital condyles joint (see moment of force OC joint versus time (Fig. 10.11*e*)]. Note, however, that the linear acceleration of the center of gravity of the head is well reproduced by the manikin, except for the peak acceleration at 100 ms [see response of head acceleration versus time (Fig. 10.11*f*)]. In contrast, the three-dimensional head and neck model for MADYMO can be seen to reproduce most human responses when active muscle behavior is included (the continuous lines in Fig. 10.12). In this computer model, the neck muscles are represented by simple cords between anatomical attachment points on the head and the base of the neck.

Animals. Employing the results of experiments with animals to predict biodynamic responses in humans introduces uncertainties associated with interspecies differences. Of particular concern is the difference in size of body parts and organs, which influences resonance frequencies. For this reason, most animal research on the limit of exposure to rapid horizontal deceleration and to vertical acceleration, which commonly involves shock and impact, has employed mammals of roughly similar size and mass to man (i.e., pigs and chimpanzees). Research on pathophysiological mechanisms is less subject to concerns with different body size, and more with the biological equivalence of the systems being studied.

Cadavers. Human cadavers have also been used for experiments involving potentially injurious stimuli, and in particular to relate skull fracture to frontal head impact. Such studies resulted in the formulation of the Wayne State concussion tolerance curve, which has been widely used to define survivable head impacts in motor vehicle collisions (SAE J885, 1986). Cadavers lack appropriate mechanical properties for tissues and muscle tension. The latter is important for obtaining realistic human responses, as can be seen from Fig. 10.12 by comparing the results with and without active muscle behavior.

10.4 COUNTERMEASURES

Reducing the effects of vibration, shock, and impact on humans is effected in several ways: (1) by isolation, to reduce the transmission of dynamic forces and accelerations to the body; (2) by personal protective equipment, to distribute the dynamic forces over as large a surface area as possible; and (3) by redesign, to reduce the source's vibration intensity. These subjects are considered after summarizing the occurrence of health effects from exposure to vibration, shock, and impact, which establishes the performance required of ameliorative measures.

10.4.1 Occurrence of Health Effects and Injury

There is extensive literature on the occurrence of health effects and injury from exposure to vibration, shock, and impact, which has been reviewed recently (von Gierke et al., 2002) and serves as the basis for the present discussion. Estimates of exposures necessary for common human responses and health effects are summarized in Table 10.2 for healthy adults; the interested reader is directed to the references given in the table for more complete information, or to the recent review article cited. Included in Table 10.2 are the metric used for assessing the exposure, the frequency weighting of the stimulus, and a *representative* value for the response or health effect under consideration. As already noted, there are large variations between individuals in response, and susceptibility, to vibration, shock, and impact.

Vibration Perception. The perception of vibration depends on the body site and on the stimulus frequency. The thresholds in Table 10.2 are typical of those for healthy adults, and are expressed as instantaneous RMS accelerations [i.e., with $T \approx 1$ s in Eq. (10.2)]. The value for whole-body vibration is given in terms of a frequency-weighted acceleration, and so is applicable to vibration at frequencies from 1 to 80 Hz. The values for hand-transmitted vibration are for sinusoidal stimuli applied to the fingertips of males (M) and females (F) at the specified frequencies.

Thresholds for Health Effects. Thresholds for the onset of health effects have been estimated for regular, near-daily exposure to hand-transmitted and to whole-body vibration. The metrics employed, however, differ. For hand-transmitted vibration, the assessment is in terms of the magnitude of the 8-hour, energy-equivalent, frequency-weighted, RMS, vector acceleration sum, $a_{\mathrm{WAS}(8)}$. This metric employs values of the RMS component accelerations averaged over 8 hours [i.e., T = 28,800 s in Eq. 10.2] which have been frequency-weighted using W_h in Fig. 10.1. The components are determined for each of the directions of the basicentric, or biodynamic, coordinate system shown in Fig. 10.2, $a_{\mathrm{X,RMS}(8)}$, $a_{\mathrm{Y,RMS}(8)}$, and $a_{\mathrm{Z,RMS}(8)}$, respectively. Thus:

$$a_{\mathrm{WAS}(8)} = [a^2_{X,\mathrm{RMS}(8)} + a^2_{Y,\mathrm{RMS}(8)} + a^2_{Z,\mathrm{RMS}(8)}]^{1/2} \tag{10.12}$$

A reduction in the metric will occur for a given acceleration magnitude if the duration of the exposure is reduced, as is illustrated by the following example (see also "Vibration Exposure") in Sec. 10.1.1).

TABLE 10.2 Estimates of Health and Injury Criteria for Healthy Adults.

Human response	Metric	Frequency or weighting	Value	Source
Perception (population mean, threshold):				
Whole Body (vertical)	RMS accel.	W_k	0.015 m/s²	ISO 2631-1, 1997
Fingertip (selected frequencies)	RMS accel.	4 Hz	0.0075 m/s²	ISO/FDIS 13091-2, 2002
	RMS accel.	31 Hz	M—0.10 m/s²	ISO/FDIS 13091-2, 2002
	RMS accel.	31 Hz	F—0.12 m/s²	ISO/FDIS 13091-2, 2002
	RMS accel.	125 Hz	M—0.25 m/s²	ISO/FDIS 13091-2, 2002
	RMS accel.	125 Hz	F—0.32 m/s²	ISO/FDIS 13091-2, 2002
Health effects (threshold):				
Hand (HAVS)	$a_{WAS(8)}$	W_h	1 m/s²	ISO 5349-1, 2001
Whole body				
Z direction	VDV	W_k	8.5 m/s$^{1.75}$	ISO 2631-1, 1997
X and Y directions	VDV	W_d	6.1 m/s$^{1.75}$	ISO 2631-1, 1997
Risk of injury (5–10% probability):				
Hand (HAVS)	$a_{WAS(8)}$	W_h	3 m/s²	ISO 5349-1, 2001
Whole-body vibration, Z (direction)	VDV	W_k	17 m/s$^{1.75}$	ISO 2631-1, 1997
Shocks to the body (headward):*				
1 shock/day ($n_q = 1$)	$E(\text{DRI}_q, n_q)$	DRI model	$\text{DRI}_{max} = 14$	After Allen, 1978, and Payne, 1976
10 shocks/day ($n_q = 10$)	$E(\text{DRI}_q, n_q)$	DRI model	$\text{DRI}_{max} = 8$	Allen, 1978
100 shocks/day ($n_q = 100$)	$E(\text{DRI}_q, n_q)$	DRI model	$\text{DRI}_{max} = 6$	Allen, 1978
Survivable single shock or impact:				
To body (headward)*	DRI	DRI model	20	After Allen, 1978
To head (manikin)	HIC	None	1000	NHTSA FMVSS 208†

M-male
F-female
*When body, is restrained.
†U.S. National Highway Traffic Safety Administration Federal Motor Vehicle Safety Standard 208.

EXAMPLE 10.2 *A worker uses a percussive rock drill for 3 hours daily. Measurement of the handle vibration indicates a frequency-weighted RMS acceleration of 18 m/s² along the drill axis, and a frequency-weighted RMS acceleration of 5 m/s² perpendicular to this axis. Estimate the 8-hour daily exposure.*

Answer: *From the (limited) available information, we obtain the following approximate values:*

$$a_{Z,\text{RMS}(3)} = 18 \qquad a_{X,\text{RMS}(3)} = a_{Y,\text{RMS}(3)} = 5$$

So

$$a_{\text{WAS}(3)} = (5^2 + 5^2 + 18^2)^{1/2} = 19.3$$

Now an exposure for 3 hours ($T_{(3)}$) can be expressed as an equivalent 8-hour exposure using Eq. (10.5):

$$a_{\text{WAS}(8)} = a_{\text{WAS}(3)}\left(\frac{T_{(3)}}{T_{(8)}}\right)^{1/2}$$

$$a_{\text{WAS}(8)} = 19.3\left(\frac{10{,}800}{28{,}800}\right)^{1/2}$$

so that

$$a_{\text{WAS}(8)} = 11.8 \text{ m/s}^2$$

The assessment of whole-body vibration employs the VDV averaged over 8 hours [i.e., T = 28,800 s in Eq. (10.6)], with frequency weighting W_k for vertical vibration and frequency weighting W_d for horizontal vibration. It is believed that the higher-power metrics, as recommended here, better represent the hazard presented by motion containing transient events, particularly when these become small-magnitude shocks or impacts. The most appropriate metric, however, remains a subject for research.

Risk of Injury. Exposures to vibration magnitudes between the threshold for perception and that for health effects commonly occur in daily life. Near-daily exposures to values of $a_{WAS(8)}$ and VDV in excess of those estimated to result in 5 to 10% injury in Table 10.2 occur in numerous occupations (involving some 8 million persons in the United States) and lead to the symptoms described in Sec. 10.1.2.

There have been several attempts to relate shocks and impacts to risk of injury, as they can be life threatening (Brinkley et al., 1990; Eiband, 1959; SAE J885, 1986). For headward accelerations, the method proposed by Allen is included here (Allen, 1978). This is based on the DRI biodynamic model (see Sec. 10.3.1), which is extended to include multiple shocks or impacts. The assessment of the risk of spinal injury depends on the number of shocks or impacts entering the buttocks of a seated person, as listed in Table 10.2. For exposures consisting of multiple headward shocks or impacts of differing magnitudes (e.g., as in off-the-road vehicles), if there are n_q shocks or impacts of magnitude DRI_q, where q = 1, 2, 3, …, Q, then the exposure is believed to be acceptable if

$$E(\mathrm{DRI}_q, n_q) = \sum_{q=1}^{Q}\left[\frac{\mathrm{DRI}_q}{(\mathrm{DRI}_{\max})_q}\right] \leqslant 1 \tag{10.13}$$

In this expression, the denominator is the maximum allowable DRI corresponding to the *observed* number of shocks or impacts n_q with magnitude DRI_q and is derived from Table 10.2.

Survivable Shocks and Impacts. Estimates for *survivable* exposures to single impacts and shocks are given for both headward acceleration and spineward deceleration in Table 10.2. For headward acceleration, experience with nonfatal ejections from military aircraft suggests that a DRI of 18 is associated with a 5 to 10 percent probability of spinal injury among healthy, *young* males who are restrained in their seats. For spineward deceleration, the estimate is based on the *head injury criterion* (HIC), which was developed from the severity index (see Sec. 10.1.1). The metric is defined as:

$$\mathrm{HIC} = \left\{(t_2 - t_1)\left[\frac{1}{(t_2 - t_1)}\int_{t_1}^{t_2} a(t)\,dt\right]^{2.5}\right\}_{\max} \tag{10.14}$$

where t_1 and t_2 are the times between which the HIC attains its maximum value and $a(t)$ is measured at the location of the center of gravity of the head. The HIC is applied to instrumented crash test dummies; its ability to rank order, by magnitude, the severity of injuries in humans has been questioned. The time interval $(t_2 - t_1)$ is commonly chosen to be 36 ms, a value prescribed by the NHTSA, though a shorter interval (e.g., 15 ms) has been suggested as more appropriate (SAE J885, 1986). The value listed in Table 10.2 is the maximum allowable by the NHTSA in frontal collisions. The assessment is related to the Wayne State concussion curve (see Sec. 10.3.3) and is applicable to head injury in vehicle and aircraft crash. Survivable injuries to the neck and chest, and fracture limits for the pelvis, patella, and femur in frontal and side impacts have also been proposed for manikins (Mertz, 1993).

10.4.2 Protection against Whole-Body Vibration

Vibration Isolation. Excessive whole-body vibration is most commonly encountered in transportation systems, where it predominantly affects seated persons. In consequence, an effective remedial measure is to reduce the vertical component of vibration transmitted through seats (and, where applicable, vehicle suspension systems), by means of low-pass mechanical filters. The transmissibility of a vibration-isolated seat or vehicle suspension (neglecting tire flexure) can be modeled in the vertical direction by the mechanical system shown in Fig. 10.6. For a vibration-isolated seat, the spring and damper represent mechanical elements supporting the seat pan and person, which are represented by a single mass *m*. For a vehicle suspension, the mechanical elements support the passenger compartment. With this model, it is evident that the vibration transmitted to the body will be reduced at frequencies above the resonance frequency of the mechanical system defined by the spring, damper, and total supported mass [i.e., for $r > \sqrt{2}$ in Fig. 10.6 and Eq. (10.10)]. The frequency weightings of Fig. 10.1 suggest that effective vibration isolation for humans will require a resonance frequency of 1 Hz or less. The low resonance frequency can be achieved by using a soft coiled spring, or an air spring, and a viscous damper. Vehicle suspensions with these properties are commonly employed. So-called *suspension seats* are commercially available, but are limited to applications in which the vertical displacement of the seat pan that results from the spring deflection is acceptable. A situation can be created in which the ability of a driver to control a vehicle is impaired by the position, or motion, of the person sitting on the vibration-isolated seat relative to the (nonisolated) controls.

Active Vibration Reduction. An *active vibration control system* consists of a hydraulic or electrodynamic actuator, vibration sensor, and electronic controller designed to maintain the seat pan stationary irrespective of the motion of the seat support. Such a control system must be capable of reproducing the vehicle motion at the seat support, which will commonly possess large displacement at low frequencies, and supply a phase-inverted version to the seat pan to counteract the vehicle motion in real time. This imposes a challenging performance requirement for the control system and vibration actuator. Also, the control system must possess safety interlocks to ensure it does not erroneously generate harmful vibration at the seat pan. While active control systems have been employed commercially to adjust the static stiffness or damping of vehicle suspensions, to improve the ride comfort on different road surfaces, there are currently no active seat suspensions.

10.4.3 Protection against Hand-Transmitted Vibration

Vibration-Isolated Tool Handles. Vibration isolation systems have been applied to a range of powered hand tools, often with dramatic consequences. For example, the introduction of vibration-isolated handles to gasoline-powered chain saws has significantly reduced the incidence of HAVS among professional saw operators. Unfortunately, such systems are not provided for the handles of all consumer-grade chain saws. The principle is the same as that described for whole-body vibration isolation, but in this case the angular resonance frequency can be ~350 rad/s (i.e., $f_0 \approx 55$ Hz) and still effectively reduce chain-saw vibration. The higher resonance frequency results in a static deflection of the saw tip relative to the handles that, with skill, does not impede the utility of the tool.

Tool Redesign. Some hand and power tools have been redesigned to reduce the vibration at the handles. Many are now commercially available (Linqvist, 1986). The most effective designs counteract the dynamic imbalance forces at the source—for example, a two-cylinder chain saw with 180° opposed cylinders and synchronous firing. A second example is a pneumatic chisel in which the compressed air drives both a cylindrical piston into the chisel (and workpiece) and an opposing counterbalancing piston; both are returned to their original positions by springs. A third is a rotary grinder in which the rotational imbalance introduced by the grinding wheel and motor is removed by a *dynamic balancer*. The dynamic balancer consists of a cylindrical enclosure, attached to the motor spindle, containing small ball bearings that self-adjust with axial rotation of the cylinder to positions on the walls that result in the least radial vibration—the desired condition.

Gloves. There have been attempts to apply the principle of vibration isolation to gloves, and so called antivibration gloves are commercially available. However, none has yet demonstrated a capability to reduce vibration at the frequencies most commonly responsible for HAVS, namely 200 Hz and below (an equinoxious frequency contour for HAVS is the inverse of frequency weighting W_h in Fig. 10.1). Performance requirements for antivibration gloves are defined by an international standard (ISO 10819, 1997). No glove has satisfied the modest transmissibility requirements, namely <1 at vibration frequencies from 31.5 to 200 Hz, and <0.6 at frequencies from 200 to 1000 Hz. An extremely soft spring is needed for the vibration isolation system because of the small dynamic mass of the hand if the resonance frequency is to remain low [see Eq. (10.11)]. An air spring formed from several air bladders sewn into the palm, fingers, and thumb of the glove appears most likely to fulfill these requirements (Reynolds, 2001).

10.4.4 Protection against Mechanical Shocks and Impacts

Protection against potentially injurious shocks and impacts is obtained by distributing the dynamic forces over as large a surface area of the body as possible and transferring the residual forces preferably to the pelvic region of the skeleton (though not through the vertebral column). Modifying the impact-time history to involve smaller peak forces lasting for a longer time is usually beneficial. Progressive crumpling of the passenger cabin floor and any forward structural members while absorbing horizontal crash forces, as well as extension of a seat's rear legs, are all used for this purpose.

Seat Belts and Harnesses. For seated persons, lap belts, or combined lap and shoulder harnesses, are used to distribute shock loads, and are routinely installed in automobiles and passenger aircraft. In addition, the belts hold the body against the seat, which serves to strengthen the restrained areas. Combined lap and shoulder harnesses are preferable to lap belts alone for forward-facing passengers, as the latter permit the upper body to flail in the event of a spineward deceleration, such as occurs in motor vehicle and many aircraft crashes. Harnesses with broader belt materials can be expected to produce lower pressures on the body and consequently less soft tissue injury. For headward accelerations the static deformation of the seat cushion is important, with the goal being to spread the load uniformly and comfortably over as large an area of the buttocks and thighs as possible.

A significant factor in human shock tolerance appears to be the acceleration-time history of the body immediately before the transient event. A *dynamic preload* imposed immediately before and/or during the shock, and in the same direction as the impending shock forces (e.g., vehicle braking before crash), has been found experimentally to reduce body accelerations (Hearon et al., 1982).

Air Bags. Although originally conceived as an alternative to seat belts and harnesses, air bags are now recognized to provide most benefit when used with passive restraints, which define the position of the body. The device used in automobiles consists, in principle, of one or more crash sensors (accelerometers) that detect rapid decelerations, and a controller that processes the data and initiates a pyrotechnic reaction to generate gas. The gas inflates a porous fabric bag between the decelerating vehicle structure and the occupant within about 25 to 50 ms, to distribute the shock and impact forces over a large surface of the body.

An example of the use of the MADYMO model to simulate air bag inflation and occupant response to the frontal collision of an automobile is shown in Fig. 10.13. In this diagram, the response of a person wearing a shoulder and lap seat belt has been calculated at 25-ms time intervals following the initiation of air bag inflation. The forward rotation of the head is clearly visible and is arrested before it impacts the chest. Also, the flailing of the arms can be seen.

Air bags can cause injury if they impact someone positioned close to the point of inflation, which may occur in the event of unrestrained, or improperly restrained, vehicle occupants (e.g., small children; see Fig. 10.9).

Helmets. Impact-reducing helmets surround the head with a rigid shell to distribute the dynamic forces over as large an area of the skull as possible. The shell is supported by energy absorbing material formed to the shape of the head, to reduce transmission of the impact to the skull. The shell of a helmet must be as stiff as possible consistent with weight considerations, and must not deflect sufficiently on impact for it to contact the head. The supporting webbing and energy absorbing foam

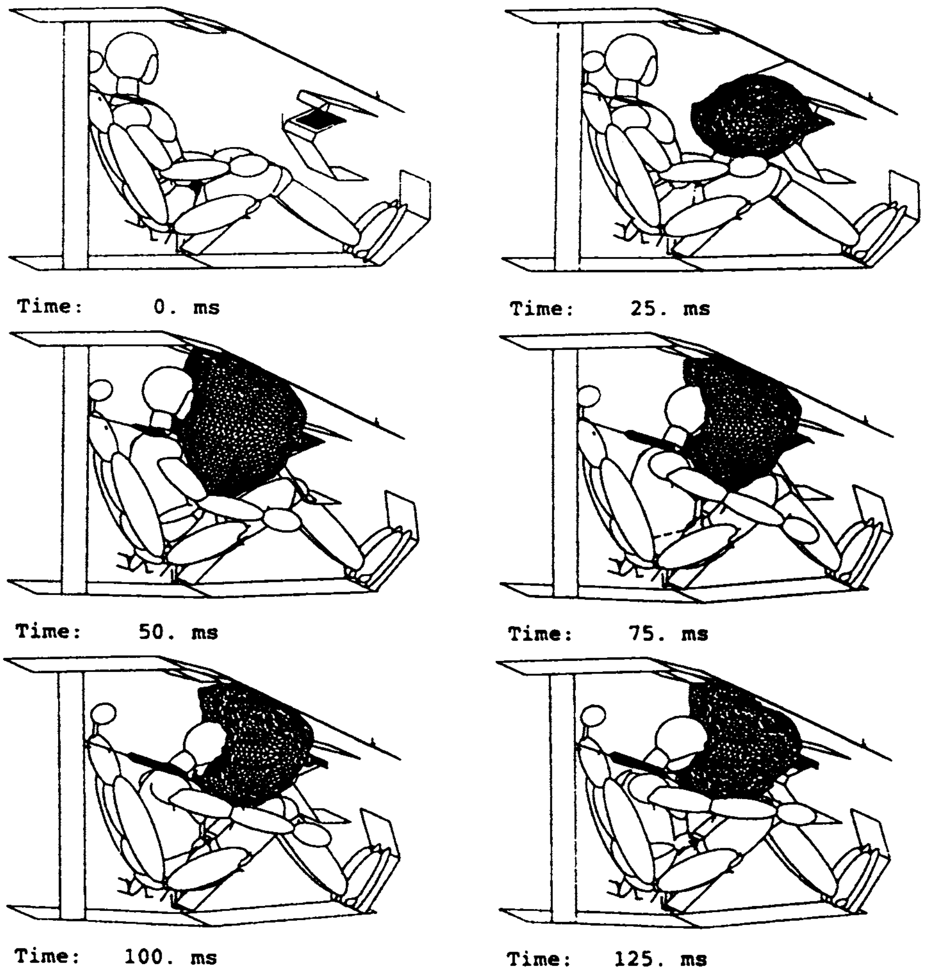

FIGURE 10.13 MADYMO simulation of the response of a person wearing a shoulder and lap seat belt to the inflation of an air bag in a frontal motor vehicle collision. The model predicts the body position every 25 ms after the collision. Note the time for the air bag to inflate (between 25 and 50 ms), the rotation of the head, the flailing of the arms, and the bending of the floor. (*AGARD-AR-330, 1996*.)

plastic must maintain the separation between the shell and the skull, and not permit shell rotation on the head, to avoid the edges of the helmet impacting the neck or face.

Most practical helmet designs are a compromise between impact protection and other considerations (e.g., bulk and weight, visibility, comfort, ability to communicate). Their efficacy has been demonstrated repeatedly by experiment and accident statistics.

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