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Human body dynamics: impact, occupational, and athletic aspects

EDITED BY
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Preface

Daily living entails a risk of injury from occupational tasks, from car crashes, and from taking part in sport. The study of human body dynamics reveals the mechanisms of injuries sustained in these situations and the measures that can be taken to avoid them. Moreover, in the area of athletics and sport, human body dynamics also plays an active part in the analysis of factors that contribute to the athlete's performance, and in the design of protective clothing and equipment. This book brings together the analyses of various types of human body dynamics involved in impact situations, occupational tasks, and athletics and sporting events, by invoking the disciplines of engineering mechanics, analytical mechanics, dynamics of structures, vibration, and impact mechanics.

The book is divided into three sections: impact; occupational tasks and environment; and athletics and sport. The first section deals with theoretical models of head, thorax, and spinal injury, to elucidate the mechanisms of injury and describe appropriate restraint and protective systems, as well as vehicle design for crashworthiness. The second section presents computerized human body models first to determine the relationships between applied forces and resulting displacements in simulated jumping, work motions in space, and response to impulsive forces; secondly to analyse slow moves in the handling of heavy materials in order to understand the mechanism of low-back injury and how it can best be prevented; and thirdly to determine maximal levels for load-lifting, pushing, and pulling. Experimental studies of the response of the body to vibrations from industrial machinery and vehicles are also described, together with suggestions for improving the design of, among other things, hand-held power tools, tractor suspension systems, and the interiors of ambulances. The last section deals with athletics and sport. Theoretical mechanics models are developed for the following: (i) to study impulse and momentum considerations in kicking and heading a soccer ball, and kinetic-to-potential energy conversion in maximizing pole-vaulting height; (ii) to analyse discus throw, javelin throw, and shot put, in order to demonstrate the influence on performance of disc spin-speed and moment of inertia, javelin inclination and throw angle and geometry, and the shot angle; (iii) to help in the design of hockey helmets, sticks, and jogging shoes; (iv) to describe the mechanics of the impact of golf-club head and ball, to maximize the velocity imparted to a golfball; (v) to calculate energy balance in assessing rope strength when arresting the fall of rock-climbers; and (vi) to estimate the energy of karate blows and kicks, for fracture of mediums.

The book is designed as a biomechanics course text as well as a reference source in body dynamics of impact, occupational manoeuvres, and athletics, for physicians, bioengineers, and physical educators. The text employs mechanics rigor at the requisite level for elucidating the mechanisms and applications, and also emphasizes the results of the analyses in setting out the causes of injury, design considerations for protective measures and devices, and factors for maximizing performance in sport and athletics.

I wish to thank the contributing authors for their painstaking efforts and Professor Collins for his editorial assistance.

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Hamilton, Ontario

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Contents

Section I Impact

1. Head injury mechanisms—characterizations and clinical evaluation

SUNDER H. ADVANI, AYUB K. OMMAYA, AND WEN-JEI YANG

1.1. Introduction

Head injury poses a grave health problem. The National Safety Council of the United States estimates that 6 per cent of all accidents involve the head but this figure is as high as 67 per cent in automobile accidents. In the United States such accidents cause about three million head injuries annually; 50 000 of these are fatal. The head injury literature is vast and varied since several disciplines are involved. These include neurophysiology, neurosurgery, neuropathology, kinesiology, engineering mechanics and computer modelling. With the objective of bringing these fields into an interdisciplinary focus, the *1966 Head Injury Conference Proceedings* [1] represent a milestone in terms of discipline cross-fertilization. This chapter essentially presents subsequent work which has emerged from this vantage point and is oriented towards the biomechanics of head injury stemming from a mechanical impact environment. Related work can also be found in a comprehensive review by Goldsmith [2] and studies by McElhaney, Stalnaker, and Roberts [3].

In order to understand the head injury mechanisms it is necessary that the mechanical input be related to the resultant pathophysiological responses. Dynamic loading to the head can generally be classified in two categories:

(i) Direct contact between the head and colliding object. In an automotive crash environment the head may impact the windscreen, dashboard, airbag, side door, head rest, etc.
(ii) Inertial impact such as loading transmitted to the head by torso motion via the head–neck junction.

Figure 1.1 presents an overview of the possible head injury loadings, mechanisms, and resulting clinical and pathological responses. A fundamental but controversial query still exists regarding the delineation of the translational and rotational components of the head centre of gravity acceleration vector and their relative contributions to head injury in various impacts. It is hoped that the following sections on anatomical, constitutive, modelling (analytical and experimental), and pathophysiological considerations will present the reader with a state-of-the-art appraisal. Selected literature is

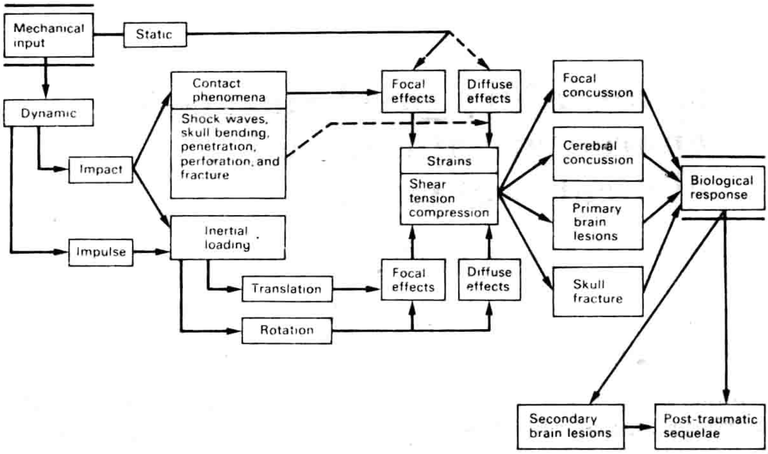

FIG. 1.1. Possible head injury loadings, mechanisms, and responses.

directly cited wherever possible. Additionally, pertinent references are listed regularly by the National Institutes of Health, USA.

1.2. Anatomical considerations

A comprehensive anatomical description of the human head is not undertaken here since several standard textbooks on human anatomy such as Gray [4] and Woodburn [5] provide excellent qualitative descriptions and agree well. Useful anatomical information can also be found in various atlases [6] and specialized articles. Detailed measurements and landmark co-ordinates for a typical 50th percentile human skull can be found in the work by Hubbard and McLeod [7]. A brief anatomical summary pertinent to the biomechanics of head impact is presented here.

The human head is a complex structural system which on a simplified basis is composed of the scalp, the skull, the dura, the pia–arachnoid complex, the brain, the blood vessels, and the cerebrospinal fluid. Considering the general structure of the head with regard to the central nervous system (CNS) one is impressed with the protective packaging. In general the CNS can be considered as a viscoelastic mass hydraulically shock mounted in a stiff container with external shock-attenuating layers. The external protective layer, or scalp, is composed of integument, subcutaneous fat, superficial facia, galea aponeurotica, and pericranium. Compared with other body skin the scalp is relatively thick, especially in the occipital region. The scalp is also richly supplied with blood vessels which course over the entire cranial area. As a composite tissue the scalp, along with variable quantities of hair, presents the first line of defence from physical trauma.

The eight bones comprising the skull are knit together in a structurally sound design approximating a spherical shell. The composition of the cranial bones is characterized by two lamellae or inner and outer tables between which lies a bed of dipolic tissue. In certain portions of the bones, particularly at their edges and junctures, the tables join internally into soild bone devoid of diploë. The sutures joining the cranial bones of an adult are comparatively strong in compression loading, although, depending on the variables of force application effecting the trauma, they may shear prior to or along with bone fracture.

The meninges impose the next and last barrier to access to the brain. The outermost membrane, the dura mater, offers the most significant structural protection of the three meninges. The disposition of the dura mater varies over the entire surface of the brain (and spinal cord). Elements of this tissue elaborate to form the falx and tentorium partitions among others. On the cranial surface the dura mater is a tough elastic tissue tightly connected to the internal surfaces of the cranial bones. In contrast, the subdural surface is relatively smooth and unattached, being equipped with a layer of mesothelium lining. Communicating via the arachnoid and parallel with the dura mater is the pia mater. This membrane, which is much finer than the dura, conforms to the cortical folds and is invested with the vascular network supplying the nervous tissue. Its strength is secondary to the transport of blood over the surface of the brain. The last meningeal membrane is the arachnoid which lies between and traverses the space between the dura mater and the pia mater. It is a delicate avascular membrane of little structural value in itself. Between the arachnoid and the pia mater lies the subarachnoid space filled with cerebrospinal fluid.

The brain is a soft structure composed of nerve cells, the grey matter, and their axons, the white matter, both of which are supported by the glia, all of these being derived from the ectoderm. It consists of two cerebral hemispheres, the basal ganglia, the cerebellum, and the brain stem. The cerebral hemispheres are divided into the frontal, temporal, parietal, and occipital lobes. Inside the brain are cavities, called ventricles, containing cerebrospinal fluid. The ventricles are in continuity with each other, either directly or indirectly, and the fourth ventricle empties into the cisterns at the base of the brain. The basal cisterns are in continuity with the subarachnoid spaces within the spinal canal and over the surface of the brain. The cerebrospinal fluid is largely secreted within the ventricles and is largely absorbed over the surface of the cerebral hemispheres.

1.3. Constitutive considerations

The mechanical properties of the tissues of the head have been extensively nvestigated [8]. In particular, considerable effort has been devoted to the characterization of the constitutive properties of human skull and brain tissue because of their relative importance.

Average values of the skull elastic moduli E and the ultimate strength σ_{ult} adapted from ref. 8 are presented in Table 1.1. The scatter in the data, although not reported here, is wide in range and typical for such tests. The variability of the data compared with that in literature is evident. For example, Wood [10] has reported an average tensile strength of 10 000 lbf in^{-2} at a strain rate of 0.01 s^{-1}. It is noteworthy, however, that the values in Table 1.1 represent lumped material properties of the structure in view of the skull compact bone–diploë sandwich configuration. Typical average compression curves for skull specimens (biopsy, autopsy, frozen and embalmed) tested in the compressive and tangential modes are illustrated in Fig. 1.2. Flexure characteristics of layered cranial bone have been reported by Hubbard, Melvin, and Barodawala [11].

TABLE 1.1
Average static mechanical properties of skull bones

	SAMPLE SOURCE			
	Biopsy	Frozen biopsy	Autopsy	Embalmed
E (radial compression) (lbf in^{-2})	5.23×10^4	22.99×10^4	11.84×10^4	38.10×10^4
E (tangential compression (lbf in^{-2})	59.40×10^4	62.6×10^4	37.8×10^4	80.8×10^4
E (tangential tension) (lbf in^{-2})	–	–	–	127×10^4
σ (ultimate radial compression) (lbf in^{-2})	–	21.1×10^3	14.21×10^3	13.9×10^3
σ (ultimate tangential compression) (lbf in^{-2})	11.87×10^3	11.81×10^3	7.34×10^3	13.12×10^3
σ (ultimate tangential tension) (lbf in^{-2})	–	–	–	6.3×10^3 [a]

[a] Value adapted from McElhaney *et al.*[9].

The rheological response of human brain tissue in pure shear has been reported by Shuck and Advani [12]. Mechanical shear strain failure levels of 0.035 rad have been identified at 10 Hz from dynamic torsion tests. Viscoelastic models characterizing brain tissue properties in the frequency range 5–350 Hz have also been formulated. A plot of the experimental storage G_1 and the loss modulus G_2 for human brain tissue in the yielded and unyielded states is illustrated in Fig. 1.3. Free-standing dynamic compression tests on brain tissue have been conducted by Estes and McElhaney [13]. The instantaneous dynamic elastic modulus at high strain rates is about 10 lbf in^{-2}. Creep and relaxation tests on brain tissue have been reported by Galford and McElhaney [14]. Compressibility and ultrasonic tests indicate that the bulk modulus of brain tissue is constant (305 000 lbf in^{-2}) over a wide frequency range [8].

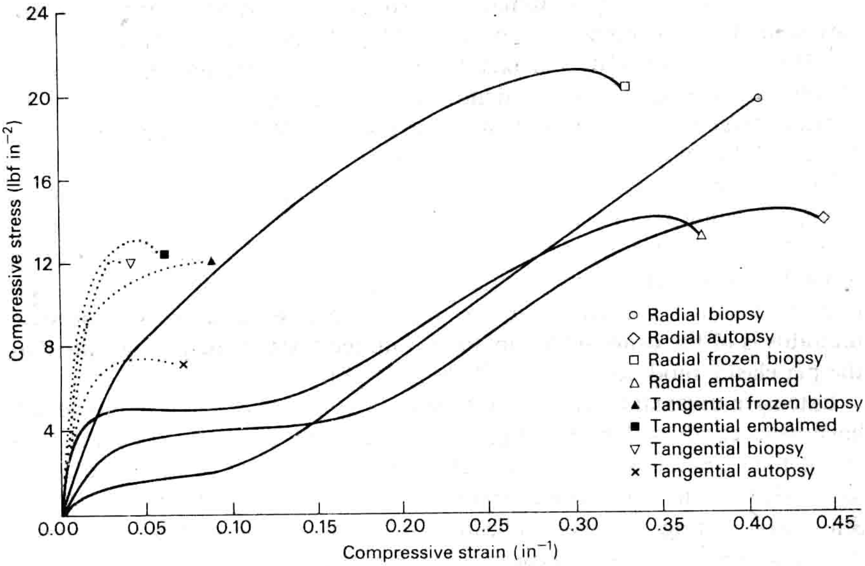

FIG. 1.2. Skull compressive stress–strain curve; composite curve (radial and tangential).

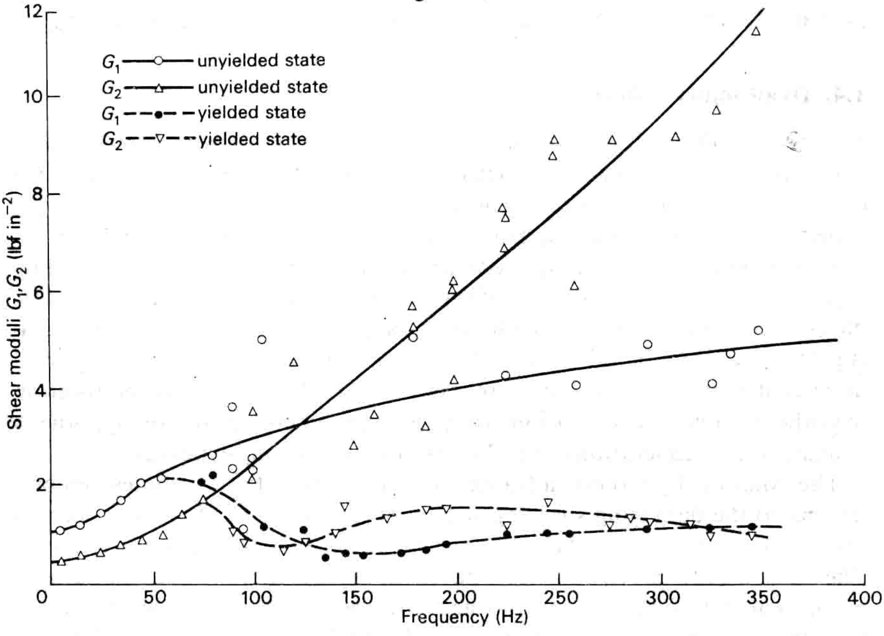

FIG. 1.3. Human brain tissue dynamic shear moduli (yielded and unyielded state). The mean values of G_1 and G_2 for white and grey matter were obtained at 98.6°F.

Free vibration testing of human dura strips have revealed that the average elastic modulus is represented by $E_1 = 4570$ lbf in^{-2}, $E_2 = 500$ lbf in^{-2} at 22 Hz ($E = E_1 + iE_2$) [8]. Uniaxial tensile tests on human dura at strain rates of 0.0666, 0.666, and 6.66 s^{-1} demonstrate little rate sensitivity effect. The initial average elastic modulus from these tests is 6000 lbf in^{-2} and the failure stress is 1100 lbf in^{-2}.

Free vibration tests on Rhesus monkey scalp specimens indicate that the average dynamic modulii are given by $E_1 = 210$ lbf in^{-2} and $E_2 = 74$ lbf in^{-2} at 20 Hz. Detailed results of relaxation and creep scalp tests have been reported by Galford and McElhaney [14]. Constant velocity free-standing compression tests at 65.0 s^{-1} indicate that the scalp thickness dynamic compressive modulus is of the order of 300 lbf in^{-2}. Limited tests of the pia indicate that the pia elastic modulus is about 20 lbf in^{-2} [8].

Extensive uniaxial tensile property investigations of cerebral blood vessels have been conducted by Chalupnik, Daly, and Merchant [15]. These tests reveal the non-linear stress–strain characteristics typical of most biological materials with the behaviour independent of strain rate from 0.001 to 50 s^{-1}. Sinusoidal testing also has indicated no frequency dependence of tensile properties up to 100 Hz. A typical relation between the Lagrangian stress (σ) and the extension ratio (λ) in the physiological stretch range is given by $\sigma = a\varepsilon^{b\lambda}$. No failure limits were identified in these studies since failure primarily occurred at the gripping ends of the vessels during these tests.

1.4. Head injury criteria

Two broad clinical classifications of cerebral trauma are generallv recognized. The first type is open head injury which occurs when the skull fractures and the dura mater is penetrated. Tearing or shearing of brain tissue in contact with the projectile or skull bone fragments usually occurs. Concomitant large intracranial pressures resulting from the volume occupied by the intruding object are also created. The second type of injury, namely closed head injury, includes cerebral concussion, contusions, lacerations, and haematoma due to rupture of cerebral blood vessels. Closed head injury mechanisms, as reported in current biomechanical literature, are categorized in terms of the rotation hypothesis (angular acceleration), the pressure gradient–cavitation hypothesis (translational acceleration), and the bending–stretching hypothesis.

The rotation hypothesis advanced by Holbourn [16] attributes cerebral trauma to the damaging shearing strains produced by head angular accelerations and the relative rotational displacement between the brain and cranium. The rationale for this hypothesis is based on the fact that the bulk modulus of brain tissue (305 000 lbf in^{-2}, constant with frequency) is considerably larger than the shear modulus (1 lbf in^{-2} at 5 Hz to 12.8 lbf in^{-2} at 350 Hz).

According to the pressure gradient–cavitation hypothesis large pressure