HUMAN INJURY MECHANISMS AND IMPACT TOLERANCE

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This paper is a brief review of the complex subject of human injury mechanisms and impact tolerance. Automotive accident-related injury patterns are briefly described, and the status of knowledge in the biomechanics of trauma of the head, neck, chest, abdomen, and extremities is discussed.

•EVERY year over 6 million people are injured and 200,000 are killed in road accidents around the world. Of these injuries and deaths, an estimated 25 percent occur in the United States annually, where accidents are the third highest cause of fatality after cardiovascular disease and cancer. Furthermore, data for deaths of persons under 45 years old show that 20,000 are caused by cardiovascular disease, 25,000 by cancer, and 38,000 by road accidents. Clearly, the young people in the country run a relatively substantial risk of death due to road accidents.

This awesome toll can be reduced by prevention of accidents through better education of road users and more efficient traffic control on one hand and the prevention of injuries and improved emergency treatment to accident victims on the other. To design automobiles and their occupant restraint systems such that accident injuries are reduced to a minimum requires a clear idea about the epidemiology of injuries and the biomechanics of injury causation.

Many studies have been conducted to delineate the anatomical distribution of injuries in road accidents, but the numbers reported vary widely from one study to another, especially if the reports originate in different countries. The reasons are twofold: (a) Traffic type and distribution vary greatly from region to region, and (b) different definitions are used for injury levels and fatalities. For example, in some studies, a fatality is reported only if the victim dies on the spot; in others, those dying within 30 days are counted as fatalities. Although it is difficult to give exact numbers to the frequency of injuries, it can be said that head and chest injuries are the most critical followed by the abdomen and then the extremities. A rough estimate of injury distribution is shown in Figure 1 (6, 11, 13, 36).

Head and neck injuries are the most frequent but are not of a critical nature as often as thoracic injuries. The introduction of improved windshields and collapsible steering columns seems to have reduced the incidence of serious head and chest injuries (16). Seat belts and shoulder harnesses have also helped prevent injuries to the head and upper torso. However, lap-belt-related abdominal trauma is known to occur and has caused some concern since it is hard to diagnose and manage. The extremities get injured quite frequently, but the injuries are not life threatening and may be minimized by improved design of car interiors.

The following sections of this paper will briefly discuss the status of knowledge in the biomechanics of trauma of the human body in terms of head injury, neck injury, chest injury, abdominal injury, and injury to the extremities.

HEAD INJURY

The automotive crash environment encompasses a wide range of impulse durations and directions. Thus, a valid head injury criterion must provide appropriate mechanisms that realistically account for the frequently observed, but poorly documented, relations of head impact tolerance and impulse duration and direction. Head injuries may be produced by direct impact that involves short durations and high accelerations or by

inertial loading that has associated large angular motions and longer time periods. Brain injury may be produced in both cases, but skull fractures and cracks are the result of head impacts only. Therefore, automobile interiors have to be designed to avoid head impacts or to reduce their severity, and the structural design should be manipulated to minimize decelerations. Head injury research as such has not been directly restraint related but focuses more on determination of the limits of tolerance to both linear and angular acceleration.

The relative contribution of linear and angular accelerations in head injury has been a matter of heated debates. Holbourn (14) contended that rotational acceleration was the main cause of all head injuries; Gurdjian, Hodgson, Thomas, and Patrick (12) emphasized linear accelerations. However, experiments done by McElhaney, Stalnaker, and Roberts (24) and Ommaya (31) indicate that either mechanism acting singly or in conjunction with the other may result in brain injury. The type of injury produced may differ according to the type of loading; e.g., contrecoup (opposite the point of impact) lesions are observed primarily in cases of direct impacts when linear accelerations are very high and diffuse brain injuries occur more often as a result of rotation of the brain relative to the skull.

A detailed analysis of brain injury in humans and its relation to the associated loading mechanism is difficult to perform since details of injuries become available only when there is an autopsy if the victim dies. Otherwise, only clinical information, which is subjective and at times incomplete, is available. Moreover, animal modeling is difficult since it is not possible to determine onset of headaches, losses of memory or cognitive functions in animals. In spite of all these difficulties, many researchers have attempted to come up with models that predict tolerance limits.

One method of presenting experimental data on the tolerance of regions of the body to acceleration is the tolerance curve shown in Figure 2 (41). For the automotive crash situation, the time regime of interest is from 1 to 300 msec, the two regions on the left of Figure 2. Such a representation leaves much to be desired when a wide variety of acceleration-time profiles are dealt with. This difficulty has led to the development of head injury criteria as shown in Figure 3 that for the most part are either weighted-impulse criteria (severity index and head injury criterion) or simple, single degree-of-freedom mechanical models (J-tolerance index, revised brain model, and effective displacement index) that were fitted to the two left time regions of the tolerance curve of Figure 2. The maximum strain criterion model is unique in that it was developed from mechanical impedance experiments on human cadaver heads and experimental lower primates. The weighted-impulse human injury criterion is currently the method used for head injury evaluation by the National Highway Traffic Safety Administration (NHTSA).

All of these models predict the severity of an impact by considering linear acceleration and impact duration. None of them considers angular accelerations, nor do most of them simulate the structural properties of the head. In the past few years, models have been proposed that incorporate rotational accelerations as well (1, 39). These models, as shown schematically in Figure 4, are still in the conceptual stage, since there is still not sufficient information that separates the effects of rotational and angular accelerations or of impact duration. The experimental techniques that were used in the past were not generally sophisticated enough to make such an analysis. Only recently have investigators made attempts to experiment by using instrumentation that will permit the complete linear and angular motion to be determined. It will be some time before sufficient research will have been done to allow a complete determination of the interplay between linear motion, angular motion, and time duration in assessing head injury potential.

In the meantime, automobiles will continue to be produced, and designers must try to optimize their safety. Dummy-based head injury criterion (HIC) measurements are just relative indicators of restraint system performance since they measure only linear acceleration. Until better evaluation techniques are developed, the design of restraint systems and automobile structures should be such that both angular and linear accelerations of the head during the crash are reduced and the head does not contact any part of the interior. Although these design considerations are complicated, they require a systems

Figure 1. Approximate anatomical distribution of injuries due to accidents.

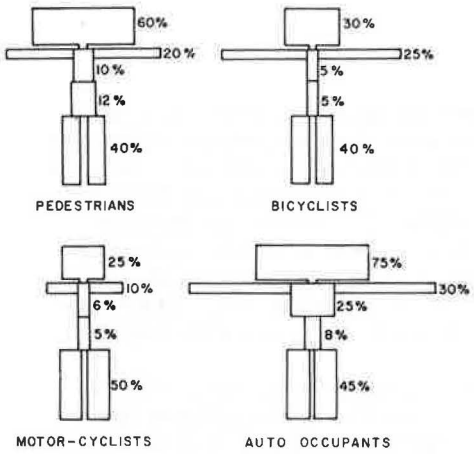


Figure 2. Human tolerance curves for +Gx acceleration.

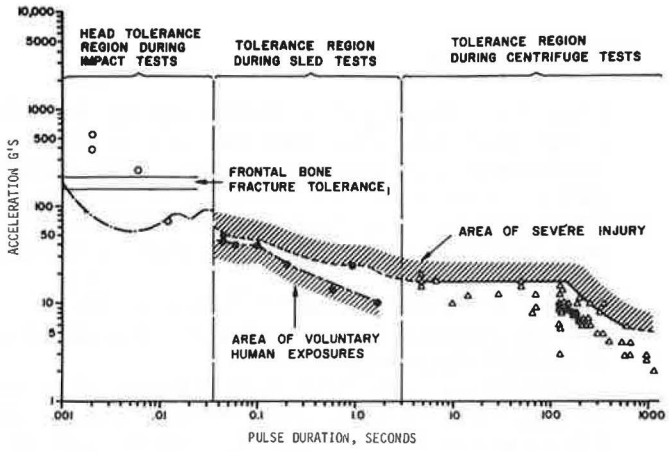
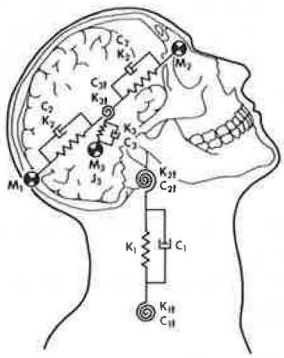


Figure 3. Head injury criteria.

SI	HIC	JTI	RBM	EDI	MSC
SEVERITY INDEX (GADD)	HEAD INJURY CRITERION (VERSACE & NHTSA)	J-TOLERANCE INDEX (SLATTENSCHEK)	REVISED BRAIN MODEL (FAN)	EFFECTIVE DISPLACEMENT INDEX (BRINN)	MAXIMUM STRAIN CRITERION (STALNAKER)
Weighted Impulse of a(t) $SI = \int_0^T [a(t)]^{2.5}\,dt$ Time in seconds Acc. in g-units	Weighted Impulse of a(t) Let $\bar{a}_{12} = \frac{\int_{t_1}^{t_2} a(t)\,dt}{(t_2 - t_1)}$ $HIC = \left\{ \max_{\text{all } t_1, t_2} \left[\bar{a}_{12}^{2.5} (t_2 - t_1) \right] \right\}$ $0 \le t_1 < t_2 \le T$	a(t) k m c X(t) $\omega_n = \sqrt{k/m}$ (rad/sec) $\beta = c/c_c$ $\omega_n = 635.$ $\beta = 1.0$	a(t) k m c X(t) $\omega_n = 175.$ $\beta = 0.4$	a(t) k m c X(t) $\omega_n = 482.$ $\beta = 0.707$	a(t) m_1 k m_2 c X(t) $m_1 = 0.6$ (lbs) $m_2 = 10.0$ (lbs) $c = 2.0$ (lb sec/in) $k = 50000.$ (lb/in)
$SI_{tol} = 1500$	$HIC_{tol} = 1000$	$J = \frac{X_{max}}{0.0925}$, $\frac{in}{in}$ $J_{tol} = 1.0$	$T < 20ms$: $\dot{X}_{tol} = 135.3 \frac{in}{sec}$ $T > 20ms$: $X_{tol} = 1.25$ in	X_{tol} = A-P RES. HUMAN 0.15 in 0.18 in DUMMY 0.17 in 0.2 in	$\varepsilon = X_{max}/L$ HUMAN: L = 5.75 in (A-P) $\varepsilon_{tol} = 0.0061$ in/in

Figure 4. Schematic lumped parameter head injury model for linear and angular two-dimensional motion.



approach so that the correct trade-offs can be made among structural integrity of the vehicle, its crush profile, occupant restraints, and interior packaging.

NECK INJURY

From the standpoint of accidental injury, the neck does not appear to react to impact in the same manner as other body regions (41), in that some low-velocity impacts often produce injury as severe as, or even more severe than, high-velocity impacts. Neck injuries can occur in many ways, but the most common cause of fractures and dislocations of the cervical spine itself is the automobile accident (3). A common form of neck injury associated with automobile accidents is the so-called whiplash injury due to indirect impact to the unsupported head-neck region of the body. At present, over 200 papers concerning whiplash types of injuries have been published; yet, to date, its precise definition, nature, measurement, diagnosis, and treatment are still subjects of medical disagreement.

In contrast to the large body of literature describing neck injuries, there are few definitive studies that attempt to quantify the loading conditions and magnitudes that can cause neck injury in humans. Mertz and Patrick (28) have performed crash sled tests on human cadavers and on a human volunteer in which the inertial forces and moments acting on the neck due to the head have been calculated. Their work did not address itself directly to injury mechanisms in the neck, however. Gadd, Nahum, and Culver (8) conducted static and dynamic bending tests on dissected unembalmed segments of human cervical spines and static bending tests on four intact cadaver necks. In both studies, only the marginal ligamentous injury stage was reached. The most comprehensive study to date on the mechanical properties of the cervical spine is the work of Sonoda (43), in which the strength of the human cervical spine was determined for compression loading, tensile loading, and torsional loading.

Experimental impacts to the cervical spines of monkeys have been studied by Gosch, Gooding, and Schneider (10). In this study, direct impacts were delivered to the vertex of the animal whose neck was extended, flexed, or aligned along the loading axis. Both bone destruction and ligamentous damage were produced, and it was found that rotation was necessary in addition to extension or flexion to produce dislocations. The presence of muscular tone at the time of injury was also found to have a notable influence on the ability to produce cervical lesions.

In most cases of severe cervical spine injury, the automotive occupant is propelled into head contact with the surrounding passenger compartment. The position of the head and neck, the impact site, and the direction of cervical spine loading determine the resulting cervical fracture. The head and neck area is either flexed (forward inclination), neutral, extended (rearward inclination), laterally flexed or rotated, and the cervical spine is subjected to bending, compression, tension, shear, or torque. Impacts about the face and frontal regions tend to produce bending in extension, and flexion results from parietal (top) or occipital (rear) head contact. When the impact is off center, a lateral flexion or rotary component may also be imparted to the head and neck. For the purposes of classification of common types of automotive accident-related cervical spine injury, the following three groups of conditions are useful:

1. Head and neck extended, cervical spine subjected to tension (extension-tension fractures);
2. Head and neck extended, cervical spine subjected to compression (extension-compression fractures); and
3. Head and neck flexed, cervical spine subjected to compression (flexion-compression fractures).

These basic groups are further modified by lateral bending and rotation. In some instances, the head is not injured, the cervical fracture being the result of direct trauma. A summary of a study of 50 clinical cervical spine fractures (37) grouped as above is shown in Figure 5.

Mechanical models of the neck have been developed by Melvin, McElhaney, and Roberts (26) and by Culver, Neathery, and Mertz (5), and mathematical models have been proposed by Bowman and Robbins (2) and McKenzie and Williams (25). Both types of models are meant to simulate the response of the neck to load. Definitive tolerance information is needed, however, before evaluations of injury potential based on such models can be made.

CHEST INJURY

The human chest (or thorax) is a ribbed shell that contains the following important organs: heart, lungs, trachea, esophagus, great blood vessels, and nerves. The size and shape of the thorax depend on the age and sex of the individual, but roughly it may be described as a truncated cone with its depth less than its breadth (aspect ratio < 1). The chest cage is semirigid in structure and not only provides protection to the internal organs but also facilitates mechanics of respiration.

Thoracic injuries may be divided into two types: injuries to the endothoracic organs and injuries to the thoracic cage. Injuries to the endothoracic organs include atrial and ventricular ruptures, aortic ruptures, damage to the electrical conducting system and the cardiac muscle, pneumothorax, hemothorax, pulmonary contusions, and rupture of the bronchi. Of these, the most frequent and most serious is the rupture of the thoracic aorta. The cardiac injuries are probably caused by the impingement of the heart between the spinal column and the sternum. Also, there is an increased possibility of cardiac rupture if the heart is full of blood. Aortic tears usually occur immediately above the heart or in the descending aorta at the isthmus. The tears are usually transverse to the vessel axis, and the exact mechanism of failure is not yet understood.

Several parameters have been suggested for evaluating injuries to the thoracic cage, in particular acceleration, force, displacement, or some combination of these. Chest impact studies have been conducted by a number of researchers (20, 21, 32, 33) by using both embalmed and unembalmed cadavers for their studies and human volunteers for quasistatic chest load-deflection studies. Chest-impact studies at the Highway Safety Research Institute (HSRI) (44) have used rhesus monkeys for evaluating injury tolerance, unembalmed cadavers for skeletal trauma, and human volunteers for static load-deflection tests. The results from these tests indicate that rib fractures do not occur at chest deflections of less than 2 in. (5 cm) for front or side impacts. For young people, this limit seems to be higher. As this deflection limit does not change appreciably from quasistatic deflection rates to dynamic impact velocities of 30 ft/sec (9 m/s), it would appear that rib fractures primarily depend on the extent of chest deflection and not on impact forces.

There are some problems associated with the use of the cadaver chest for obtaining tolerance information, and careful consideration must be used when chest impact data are interpreted. Recent studies indicate that the effects of muscle tension and air-filled lungs can contribute significantly to the load-carrying ability of the thorax. The effects of tensing of thoracic muscles are shown in Figure 6 (21, 44). The data bands cover the range for all the data gathered by various investigators, and, therefore, the wide range is due to both anatomical differences among volunteers and different testing procedures. It is worth nothing that the maximum stiffness of the tensed volunteers is about twice that of the maximum stiffness of the relaxed volunteers and almost eight times their minimum stiffness. When these curves are compared with those obtained for embalmed and unembalmed cadavers (Figure 7) (44), the stiffness of the chests of unembalmed cadavers is found to lie in the lower range of relaxed volunteers' chest stiffnesses. This is probably due to the lack of muscle tone and lung inflation. Figures 8 (33, 45) and 9 (44) show response corridors for front and side chest impacts to cadavers without lung inflation. Here the forces are almost 10 times those recorded in quasistatic tests on cadavers. In the HSRI tests, in which the impactor was a 6-in.-wide (15-cm) flat disk, the force penetration trace showed a pattern where there was an initial load spike followed by a plateau. Whereas when the chest was impacted by a simulated arm rest, the force rises progressively to a peak. This is because the force increases succes-

Figure 5. Mechanism of cervical spine injury as related to level of cervical fracture or dislocaton.

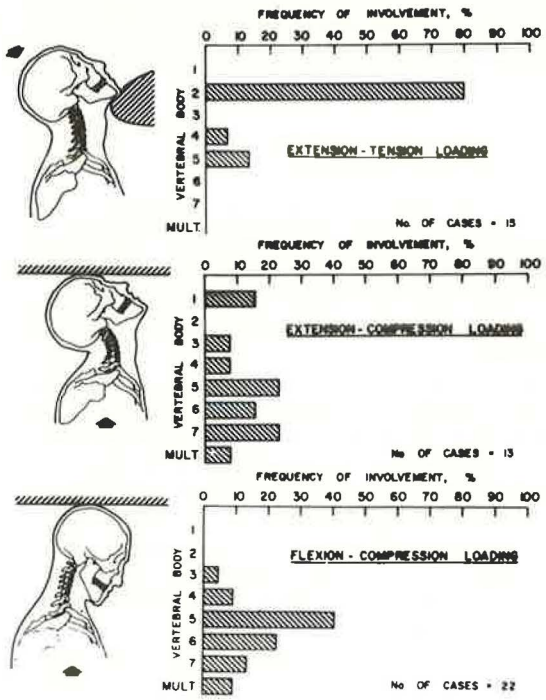


Figure 6. Range of force-penetration data for quasistatic front chest compression tests on human volunteers.

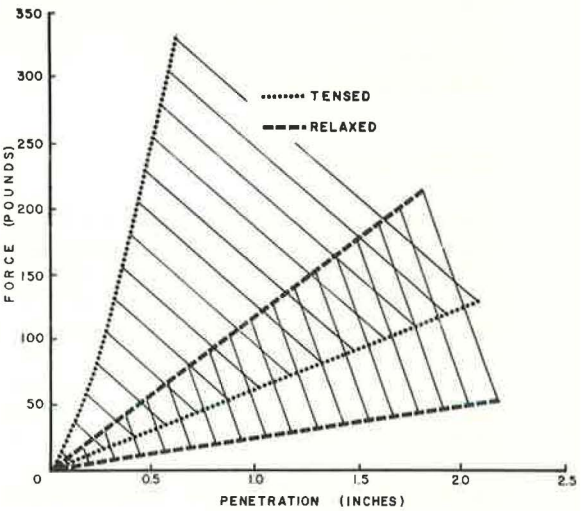


Figure 7. Range of force-penetration data for quasistatic front chest compression tests on human cadavers.

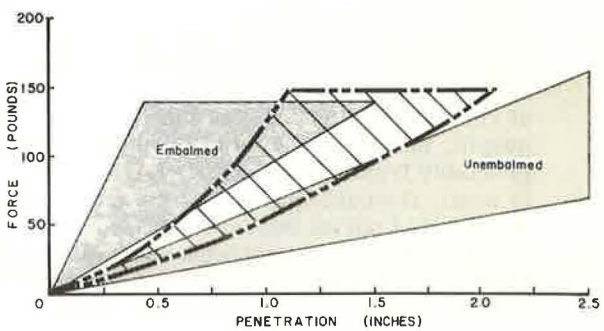
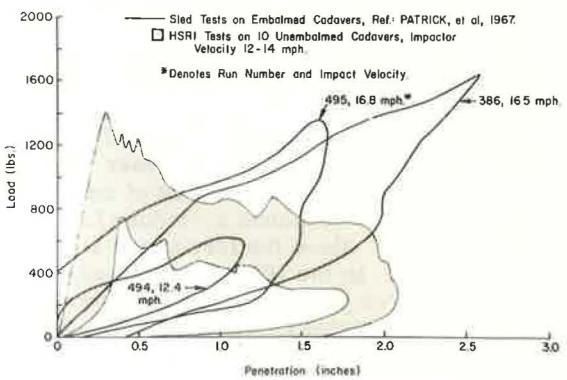


Figure 8. Dynamic load-penetration curves for human cadaver front chest impacts.



sively as more of the armrest comes in contact with the chest. This is unlike the occurrence with the flat disk, where total contact is made on impact.

Specification of thoracic impact tolerance is complicated further by the lack of biomechanical fidelity of current dummy chest structures. A deflection criterion for thoracic impact can be used directly only if the dummy chest responds to load in a manner similar to the living human thorax. Under certain well-controlled loading environments, an equivalent deflection response of a dummy chest might be usable, but it would not be useful in general applications with a variety of loading conditions. For frontal chest impact tolerance, a deflection of 1.75 in. (4.4 cm) has been suggested (44) if rib fracture is to be avoided. A tolerance value based on the American Medical Association abbreviated injury scale (AIS) level of 3 (severe, but not life threatening) would be in the 2.5 to 3.0 in. (6.4 to 7.6 cm) range for the average male (20, 44). This corresponds to approximately a 30 to 35 percent reduction in the chest depth. A similar value for the percentage reduction in chest width was found for side impact studies at HSRI (44) by using experimental lower primates. The corresponding deflection levels for an average male would be 2.65 in. (6.70 cm) for a nonfracture level and 3.72 in. (9.44 cm) for an AIS level 3 injury in side impact.

ABDOMINAL INJURY

Blunt abdominal trauma is a common cause of accidental injury and death, and motor vehicle accidents are the most frequent cause of nonpenetrating abdominal trauma. The sources of abdominal loading interior to the vehicle include steering wheel rims, lap belts, armrests, and protruding dashboard structures, knobs, and levers. Ejection of the vehicle occupants during a crash or pedestrian impact frequently produces severe abdominal trauma. The organs most frequently injured as a result of blunt abdominal trauma are the liver, kidneys, spleen, pancreas, and intestines. Diagnosis and localization of organ injury in the abdomen are difficult, and the serious threats of hemorrhage and infection require prompt surgical intervention when these organ injuries are present.

Much clinical literature has evolved over the years that documents the various forms of injuries produced by blunt abdominal trauma. In contrast, there are few quantitative data available on the loading conditions, force levels, and impact velocities that characterize typical accident situations. In particular, there are almost no quantitative data on the mechanical response of the critical abdominal organs to direct loading.

Injury to the liver due to blunt trauma can take many forms ranging from subcapsular hemotomas and superficial lacerations to the severe crushing and bursting types of injuries with stellate capsular lacerations and gross destruction and devitalization of the parenchyma (30). Bursting injuries are vastly more severe than the more common simple tears or lacerations of Glisson's capsule (23). In bursting injuries, hemorrhage is massive and the mortality rate is high regardless of the treatment instituted (7).

To simulate the trauma sustained by the liver in automobile accidents, Mays (23) dropped cadaver livers from various heights ranging from 8.5 to 91 ft (2.6 to 27 m). An important finding of this work, which was also reported by Glenn, Mujahed, and Grafe (9), is the necessity of maintaining the turgor of the liver at a level comparable to normal hemodynamic pressures so that realistic bursting injuries can be produced. Mays achieved this pressure by injecting the livers with saline solution before they were dropped and was able to produce bursting injuries as seen clinically by using energies on the order of 285 to 360 lbf-ft (386 to 488 J).

The kidneys in the adult are paired bean-shaped organs, measuring about 5 in. (13 cm) long, 3 in. (8 cm) wide, and 2 in. (5 cm) thick. They are buried in a mass of fat on each side of the vertebral column, behind all the other abdominal organs. The injury types sustained in blunt trauma (7) range from renal contusions in which there is minor disruption of the renal parenchyma to complete tears in which there is complete disruption of the organ.

Direct loading tests (27) of both liver and kidney demonstrated the sensitivity of these organs to rate of loading. The effects of loading rate were most pronounced in the liver.

Examination of Figure 10 shows that the onset of severe trauma (AIS-3) under dynamic loading occurs at a threshold stress level of approximately 45 psi (310 kPa) in the liver (27). The severity of the injury past the 3+ level is primarily related to the additional stress and strain produced in the specimen above the threshold stress level and is best characterized by the maximum average strain energy density produced in the material of the organ.

The stresses produced in the kidneys were higher than those in the livers; this indicates the effect of the tough, thick capsule of the kidney. Figure 11 shows that the dynamic stress levels necessary to cause injuries ranging from a 1 to 2 AIS rating to a 4 to 5 rating did not vary significantly and that the injury level was ordered more effectively according to the strain level (27). This effect may be attributed to the properties of the capsule.

The injury modes observed in the dynamic tests of both livers and kidneys were similar to those seen clinically. The AIS rating of these injuries correlated well with the mechanical input parameters and indicates the effectiveness of using a rating system to describe mechanical damage to tissue.

Besides liver and kidney injuries, the spleen, colon, and jejunum also get injured because of abdominal impacts. Seat belts are known to cause injuries (40) like a lacerated spleen or colon. This is especially true when the seat belts are not worn properly and the buckle is in front of the abdomen rather than the side.

Abdominal tolerance to injury is rather low, and therefore loading of this area must be avoided. Safety belts must remain below the iliac crests, and the pelvis should bear all the load. Seat belt designs that make it almost impossible for the wearer to keep them loose or twisted are necessary. Serious thought must be given to belt placement, automatic retraction, load limiting energy absorbing devices, and last but not least, comfort. Unless the belts are convenient and comfortable, occupants will always find ways to avoid using them.

EXTREMITIES

As mentioned earlier, both the upper and lower extremities are injured rather frequently in automobile accidents. Injuries to the upper extremities are not very serious and do not cause disability. They may be reduced by eliminating rigid edges in the interior of the car. However, leg injuries, though not life threatening, do cause disabilities and days lost from work. Thus, only the biomechanics of leg injury will be considered in this discussion.

In the automobile crash situation, the fractures of the upper leg are more serious than those of the lower leg. Lower leg fractures are common in pedestrians, and Kramer, Burow, and Heger (18) impacted tibia bones of more than 200 human cadavers frontally to obtain basic information for construction of safer vehicle fronts. The lowest force level recorded for fracture was 2,200 lbf (9786 N) for an impact with an 8.5-in.-diameter (28-cm) cylinder. Researchers have shown greater interest in the biomechanics of the upper leg, and there are numerous papers dealing with this subject. The hip and knee joints are critical areas of injury and more difficult to treat.

The most common form of leg loading to a vehicle occupant in an accident is a knee impact that may damage the knee, the femur, and the pelvis. A variety of studies have been conducted on the various aspects of knee impact (4, 15, 32, 38). Although the assessment of functional disability to the knee joint itself in cadaver knee impacts is difficult, one can establish fracture levels for the knee-femur-pelvis complex. The tolerable force level of axial load to the flexed knee has been established at 1,700 lbf (7562 N) by NHTSA for its standards activities. This level was based on embalmed cadaver data (33) and has recently been criticized for being too low (17). Studies presently underway at HSRI based on unembalmed cadavers indicate that the fracture load level for the unembalmed knee-femur complex may be much greater. This difference may be attributed to modification of the fracture characteristics of bone when embalmed.

Figure 9. Response envelopes for side impacts on unembalmed human cadavers.

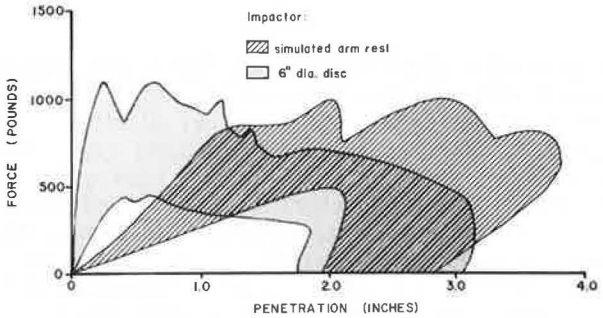


Figure 10. Dynamic stress-strain behavior of lower primate livers under direct impact.

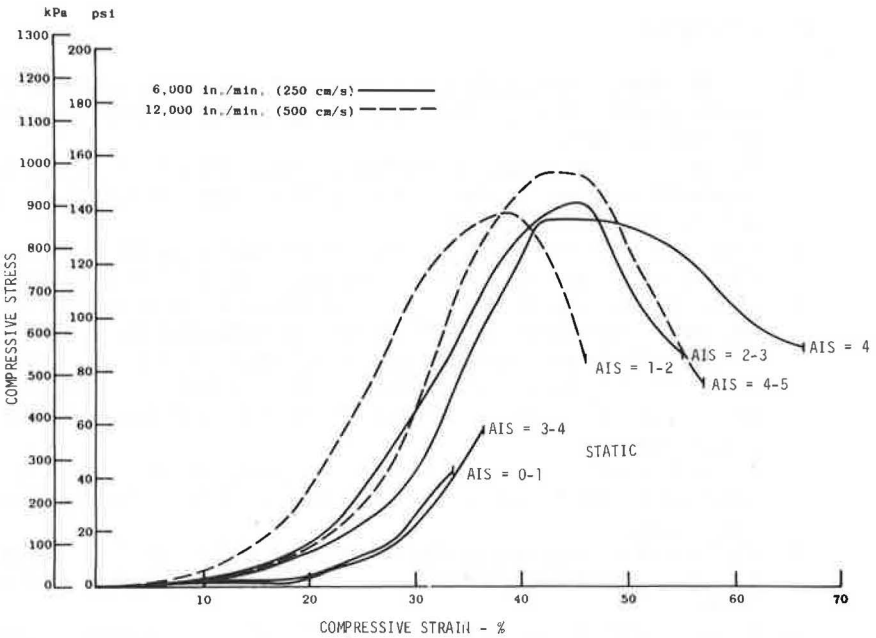
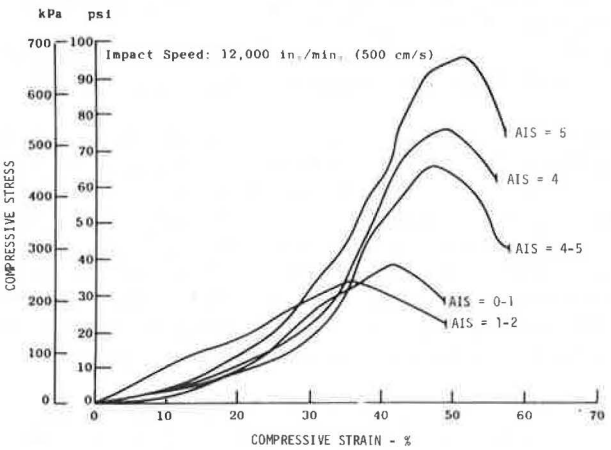


Figure 11. Dynamic stress-strain behavior of lower primate kidneys under direct impact.



SUMMARY

The increasing numbers of road accidents around the world have prompted serious research in human injury mechanisms and impact tolerance. However, lack of documentation, complicated human structures, difficulty in evaluating injury, inadequate instrumentation, and imperfect animal models make this job rather difficult. Although many researchers have spent a great deal of time studying head and chest injury, appropriate levels for tolerable impact forces, accelerations, or deflections have not yet been established, and the criteria being used are sometimes uncertain and disputable. Until the time when research provides more definitive information, the designer will have to rely on conservative estimates and values. Using these guidelines, one can design safer and more comfortable energy-absorbing restraint systems, windshields, and car interiors. These improvements in themselves will reduce the hazards of accident injury.

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