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HUMAN TOLERANCE TO IMPACT CONDITIONS AS RELATED TO MOTOR VEHICLE DESIGN

Foreword**—**This Document has not changed other than to put it into the new SAE Technical Standards Board Format.

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1. Scope—This report reviews current[1] quantitative data on human tolerance levels without recommending specific limits. Data developed on humans (including cadavers) are presented where available; however, in many cases animal data are provided where no suitable human results have been reported. This report confines itself, as much as possible, to information of direct use to the automotive designer and tester. Data of only academic interest are largely omitted; therefore, J885 should not be considered as a complete summary of all available biomechanical data.

Most of the data cited in this report applies to adult males since little information is available on women or children. The summary data provided in the tables should be considered in conjunction with the accompanying descriptive test. This material explains the manner in which the data were obtained and provides an insight as to their limitations.

1.1 Purpose—The purpose of this report is to assist the automotive safety designer and tester by providing them with quantitative data on the strength of the human body under impact loading conditions.

2. References

2.1 Applicable Publications—The following publications form a part of the specification to the extent specified herein.

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FMVSS 208

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3. *Definitions*

3.1 Abbreviated Injury Scale (AIS)—A numerical rating system for quantifying the severity of injuries to an accident victim. The rating scale is:

Code	Category
1	Minor
2	Moderate
3	Serious
4	Severe
5	Critical (survival uncertain)
6	Maximum (virtually unsurvivable)
9	Unknown

For further details see the Abbreviated Injury Scale 1980 Revision published by the American Association for Automotive Medicine.

3.2 Anterior—Front.

3.3 Anterior-Posterior (a-p)—Front to back; in humans, directed from the belly surface towards the back surface.

3.4 Articular—Pertaining to a joint.

3.5 Avulsion—Tearing away of a part.

3.6 Cancellous Bone—The spongy or lattice-like structure of a bone occurring towards its inner core.

3.7 Cartilage—Fibrous connective tissue.

3.8 Cervical—Pertaining to the neck.

3.9 Comminuted—Broken into small pieces.

3.10 Compact Bone—The dense structure of the bone which constitutes its outer portion.

3.11 Condyle—A rounded projection on a bone usually associated with a joint.

3.12 Contusion—Bruising from a direct impact.

3.13 Cricoid Cartilage—The ring shape fibrous tissue which encircles the airway passage near the top of the neck.

3.14 Distal—Remote; further away from the point of reference.

3.15 Esophagus—The passageway which carries food to the stomach.

3.16 Extension—Rearward bending when applied to the neck.

3.17 Femur—Thigh bone.

3.18 Fibula—The outer and smaller of the two bones of the lower leg.

3.19 Flexion—Bending; forward bending when applied to the neck.

3.20 Frontal Bone—The bone constituting the forehead and upper forward portion of the skull.

3.21 Functional Injury—A trauma which is not readily observable on visual examination but manifests itself as an impairment of normal usage or behavior.

3.22 Hemorrhage—Bleeding.

3.23 Hemothorax—A collection of blood in the sac surrounding the lungs.

3.24 Hyperextension—Extreme or excessive extension of a limb or part; backward overbending when applied to the neck.

3.25 Hyperflexion—Extreme or excessive flexion of a limb or part; forward overbending when applied to the neck.

3.26 Inferior—Below.

3.27 Inferior-Superior(i-s)—Below to above or lower to upper; from the trunk towards the head.

3.28 Injury—Physical disturbance, damage, or destruction to a biological structure which impairs or prevents its normal functioning.

3.29 Injury Level—A rating of a trauma's severity relative to its threat to life or degree of physical or functional impairment (cf: Abbreviated Injury Scale).

3.30 Injury Criterion—A numerical relationship between measurable engineering parameters and injury level.

3.31 In Situ—In its normal location in the body.

3.32 Intervertebral Disc—Circular pads of fibrous cartilage situated between adjacent vertebrae in the backbone.

3.33 In Vivo—Within the living body.

3.34 Laceration—A wound made by cutting or tearing.

3.35 Larynx (pl. Larynges)—The muscle/cartilage structure at the front of the neck.

3.36 Lesion—Any bodily disfunction or damage.

3.37 Ligament—A band of tissue that connects bone or supports viscera.

3.38 Loading—See Measurable Engineering Parameter.

3.39 Mandible—The bone of the lower jaw.

3.40 Maxilla—The bone which forms the central portion of the upper jaw.

3.41 **Measurable Engineering Parameter—**Physical behavior of a system detectable by instrumentation which describes the externally applied environment to, or the structural response or, the system or its surrogate. (Examples of measurable engineering parameters are force, acceleration, strain, and pressure).

3.42 **Mobilized—**Made free to move. (See Surgically Mobilized).

3.43 **Occipital Condyles—**Rounded prominences on each side of the base of the skull which articulate with the uppermost vertebra of the neck.

3.44 **Occiput—**The bone forming the rear and lower rear portion of the skull.

3.45 **Patella—**Knee cap.

3.46 **Pneumothorax—**An accumulation of air or gas in the sac surrounding the lungs.

3.47 **Posterior—**Rear.

3.48 **Process—**A prominence or projection on a bone.

3.49 **Rotation—**When applied to neck motion refers to the no gesture of the head.

3.50 **Sagittal—**A plane or section dividing the body into right and left portions.

3.51 **Spinous Process—**A projection of the rear on a vertebra.

3.52 **Sternum—**Breastbone.

3.53 **Subclavian Arteries—**Two of the four major blood vessels arising from the top of the heart; the subclavian arteries pass under the clavicles and supply blood to the upper body.

3.54 **Subdural Hematoma—**Bleeding between the two layers surrounding the brain.

3.55 **Superior—**Above.

3.56 **Supracondylar—**Situated above (superior to) a condyle.

3.57 **Surgically Mobilized—**Separated by surgery from its surrounding tissues but leaving connecting blood vessels intact.

3.58 **Suture—**A joint in which the opposed bone surfaces are closely united.

3.59 **Symphysis—**A line of union; a type of joint in which the opposing bones are firmly united by cartilage.

3.60 **Temporal Bones—**Two bones which make up the lower sides of the skull.

3.61 **Temporo-Parietal—**Side of the skull.

3.62 **Tendon—**A fibrous cord by which a muscle is attached to a bone.

3.63 **Thorax—**Chest.

3.64 **Thyroid Cartilage—**A wishbone shaped stiff tissue located in the upper portion of the neck.

3.65 **Tibia—**The larger of the two long bones of the lower leg.

3.66 Tolerance Level—The magnitude of loading which produces a specific injury level.

3.67 Tolerance Specification—An impact level which is taken as the maximum (or minimum) allowable condition for design purposes.

3.68 Trachea—The windpipe.

3.69 Trauma—See Injury.

3.70 Vertebra—One of the thirty-three bones of the spinal column.

3.71 Zygoma—Cheekbone.

4. Introduction To Biomechanics

4.1 Test Subjects—Of necessity, human tolerance levels must be inferred by indirect means such as testing volunteers (below their injury level), cadavers, or anesthetized animals. Each of these subjects has advantages and shortcomings which influence the applicability of the resultant data.

4.1.1 HUMANS—Volunteers provide the primary source for determining the effects of muscle tone and pre-bracing on the dynamic behavior of humans. Volunteers can also provide some information on the upper boundary of the "no injury" tolerance level. However, true tolerance levels cannot be determined with volunteers since they cannot be tested into the injury range. A further disadvantage is that volunteers are usually young, robust males whose pain and injury tolerance is apt to be considerably higher than that of the general population. Finally, the muscle bracing which volunteers sometimes employ can have an important effect at low levels which cannot necessarily be extrapolated to higher levels.

4.1.2 CADAVERS—Cadavers are normally employed when testing is undertaken at severity levels which would be injurious to volunteers. Cadavers are logical candidates as test subjects since they retain geometric similarity to living humans and many of their structures retain a strength similarity as well. This latter aspect is highly dependent on the treatment of the cadaver and the time duration since death. Recognition of these factors has led to extensive changes in cadaver testing techniques in recent years in an effort to make such test results more representative of living human response. These changes include the use of unembalmed cadavers, inflation of the lungs, and pressurization of portions of the vascular system with dye solutions to assist in trauma diagonsis. It is generally accepted that the mechanical strength of most living human body tissues decreases with age. Consequently data obtained from tests of cadavers of the elderly are likely to be conservative relative to the general population. Other potential shortcomings of cadaver testing revolve around their lack of muscle tone, and differences in some body properties from those of the living.

4.1.3 ANIMALS—Animal testing is generally employed to study the mechanisms of trauma since animals provide the only functioning physiological systems which can be subjected to severe impacts. They also provide the only known bridge for examining the relationships between living and dead subjects. Thus animals may provide the only possibility for evaluating the usefulness of cadaver testing. Unfortunately, the results of animal tests cannot, as yet, be quantitatively scaled with confidence to determine human tolerance levels due to the size, shape, and structual differences between animals and humans.

4.2 Application of Biomechanical Data

4.2.1 HUMAN SURROGATES—The behavior of the human surrogate is an important consideration when the biomechanical data of Section 5 are applied to automotive testing. To be of value, the surrogate must be sufficiently human-like so that its performance will be indicative of human behavior under similar circumstances. The problems of achieving such correlations are discussed in Section 6 of this report.

4.2.2 DETERMINATION OF TOLERANCE LEVELS—A comprehensive discussion of the factors involved in the determination of human tolerance levels is beyond the scope of this report. Indeed, such specifications are beyond the state-of-the-art in biomechanics except perhaps for a few academic situations. There are several difficulties which prevent a ready establishment of human tolerance levels. First, there are differences in judgement as to the specific degree of injury severity that should serve as the tolerance level. Second, large differences exist in the tolerances of different individuals. It is not unusual for bone fracture tests on a sample of adult cadavers to show a three-to-one load variation. Presumably, variations of at least this magnitude exist in the living population. Finally, most tolerance levels are sensitive to modest changes in the direction, shape and stiffness of the loading source. The above considerations indicate that complete and precise definitions of human tolerance levels will require large amounts of data based on controlled statistical samples. Only in this way can the influence of age, size, sex, and weight be comprehensively assessed and only in this way can mean loads and statistical measures of scatter be linked to specific tolerance levels.

In the interim, it is necessary to employ various tolerance measures in the development and evaluation of safety features. Probably the most widely used of such measures is the tolerance specification. This is an impact level taken somewhat arbitrarily as a boundary condition for design purposes. The tolerance specification should not be confused with the tolerance level which is the magnitude of loading that produces a specific degree of injury. As explained above, complete definitions of tolerance levels properly should be statistical measures relating probabilities of injury and degrees of injury to impact histories.

4.3 Biomechanical Materials—The body is composed of hard and soft materials which can occur together in composite body structures such as the rib cage and vertebral column. The presence of soft tissue as a bone connector allows for large structural deflections.

4.3.1 BONE—In addition to being non-homogeneous and anisotropic, a given bone often varies in shape from individual to individual. Therefore, it is not generally convenient to employ conventional stress analysis techniques for estimating the strength of a bone from its material properties. To overcome this difficulty, bones are generally tested in situ to determine their load carrying capacity as a structure. As with any structure, a variety of failure modes is usually possible depending on the distribution and location of the applied forces and, for impact situations, time duration effects may be important as well. Accordingly it is important to understand the mode of load application (that is, torsion, bending, shear) for the situations presented in Section 5; these tolerance levels should only be applied under similar conditions.

The bones of the skull and knee cap are uniquely sensitive to punch through (bearing load) failures. This is due to their anatomical construction as well as to their physical prominence and lack of soft tissue covering to provide a padding effect. The bones of the skull and knee cap are of a sandwich construction. Their innermost and outermost layers are shells of compact bone which embrace a middle zone of porous bone between them. Excessive bearing loads can punch through the outer shell at force levels that would not cause failure of the overall bone structure. Examples of this bearing load effect are given in Section 5.

4.3.2 SOFT TISSUES—The development of injury criteria for soft tissue trauma is an extraordinarily complex subject which is only in its early development stage. Progress in this field is likely to be slow for the following reasons:

a. A wide variety of possible injury mechanisms exist.
b. Small differences in the location or level of injury can have vastly different consequences to the injured person.
c. The capability to analyze and model the organs is very limited.

4.3.2.1 *Skin*—Skin has been studied more than any other soft tissue insofar as automotive collision trauma is concerned. The state-of-the-art in assessing skin injuries is summarized in SAE Information Report J202. One test procedure is to expose a synthetic skin material to a standardized impact test and then to evaluate the injury level either by subjective observation or measurement of the resultant damage to the synthetic material. This appears to be a practical method to evaluate skin injury levels when the multiplicity of skin injury mechanisms are considered. Skin trauma includes:

a. Avulsion (tearing away).
b. Contusion (bruising from direct impact).
c. Laceration (cutting).
d. Puncturing.
e. Splitting.
f. Abrasion.

4.3.2.2 *Internal Soft Tissues*—Internal soft tissues are vulnerable to all of the above types of trauma except abrasion. In addition, they can be injured by excessive displacement which may detach an organ from its vascular or ligamentous connections. In the brain, rapid displacement may result in injury due to cavitation. Unfortunately, little quantitative data exist on force, penetration, or displacement levels that are injurious to soft tissues. No synthetic internal organs are currently in common use for impact testing.

5. ***Data***—The human body can be subjected to a broad variety of trauma caused by a number of injury mechanisms; certain of these predominate for each zone of the body. Therefore, this information report discusses each body zone separately. They are considered in body order from head to legs.

5.1 Fracture Loads for the Cranium

5.1.1 FRACTURE MODES—The bones of the cranium are not homogeneous throughout their thickness, but are of a sandwich construction. The sandwich consists of a core of cancellous (low density) bone between two layers of compact (high density) bone. This arrangement allows two different types of bone fractures to occur, each arising from a different failure mode.

5.1.1.1 *Linear Fractures (failure of the structure as a whole due to bending stresses)*—When the impact is well distributed, the skull will be bent inwardly at the site of the blow, and outwardly in some regions remote from the blow. The tensile stresses (or strain energy densities according to one theory) arising in the latter regions can precipitate a crack. The crack usually originates at some point of stress concentration and propagates towards the site of the blow along an essentially direct path. The bone on each side of the crack remains in alignment in this type of fracture. A linear fracture is not life threatening per se since it does not in itself precipitate brain injury. However a linear skull fracture is a cause for concern since the integrity of the skull has been lost.

5.1.1.2 *Depressed Fractures (localized failure of a cranial bone due to concentrated forces)*—If the impact force is sufficiently concentrated, it may break through the structure locally even though the magnitude of the force might be insufficient to over-stress the bone structure as a whole. A contact area of approximately 2 in^2 (13 cm^2) is considered here to represent the transition between distributed and concentrated loading. As the contact area diminishes below this threshold size, depressed fractures are likely to occur as a result of localized stresses at the impact site. This produces a cave-in mode of failure. If the contact area is further diminished, to less than approximately 3/4 in^2 (5 cm^2) the depressed fracture takes the form of a clean punch-through with a hole size which matches the size of the struck object. This behavior is thought to be due to two concurrent failures: compression of the core of cancellous bone and shearing of the compact bone. Both the cave-in and punch-through types of depressed fractures result in an inward displacement of the bone which can lead to mechanical impingement against the brain and allow the entry of foreign bodies. Depressed fractures are potentially life threatening injuries.

5.1.2 FRACTURE DATA—Table 1 presents a summary of fracture load data for the cranium. Some of the anomalies found there can be explained by considering the impactor shape and the cranial bone properties; however, most are likely due to the large variations that are inherent in any cadaver population. The likelihood of inconsistencies is compounded by the small number of test specimens employed in many of the test series cited. These considerations suggest the need for caution in the use of this table.

TABLE 1—DYNAMIC FRACTURE FORCES FOR CRANIAL BONES RIGID SURFACE IMPACTS

Type of Impact Surface	Fracture Forces(1) Mean lb	Fracture Forces(1) Mean N	Fracture Forces(1) Range lb	Fracture Forces(1) Range N	Sample Size	Fresh or Embalmed	Ref
Frontal Bone							
flat plate(2)	1430	6360	880–2650	3910–11 790	12	fr	1
flat plate	1440	6400	1220–1770	5420–7870	6	em	2
longitudinal surface of cylinder							
1 in (2.5 cm) rad. aligned transversely	1260	5600	950–1650	4220–7340	7	em	3
1 in (2.5 cm) rad. aligned sagittally	1600	7120	940–2000	4180–8900	5	em	2
5/16 in (0.79 cm) rad. aligned transversely	1230	5470	700–1730	3110–7700	5	em	3
sphere, 8 in (20.3 cm) rad.	1180	5250	830–1530	3690–6810	5	em	2
small area flat surfaces							
1 1/8 in (2.9 cm) dia	1130	5030	848–1600	3770–7120	5	em	4
1 1/8 in (2.9 cm) dia	1390	6180	980–1990	4360–8850	5	fr	4
1 1/8 in (2.9 cm) dia	1310	5830	930–2220	4140–9880	7	em	5
0.61 in (1.55 cm) dia	1710	7610	920–2200	4090–9790	5	em	6
0.43 in (1.09 cm) dia	1030	4580	470–2000	2090–8900	5	em	6
small area rounded surface							
0.67 in (1.70 cm) dia(2)	1000	4450	620–1820	2760–8100	6	fr	1
Temporo-Parietal Bones							
flat plate (2)	1140	5070	770–1760	3430–7830	13	fr	1
flat plate	1910	8500	1050–3360	4670–14 950	7	em	2
small area flat surfaces							
1 1/8 in (2.9 cm) dia	846	3760	550–1330	2450–5920	7	fr	5
1 1/8 in (2.9 cm) dia	702	3120	302–1330	1340–5920	8	em	5
0.61 in (1.55 cm) dia	1290	5740	500–2200	2220–9790	10	em	6
0.43 in (1.09 cm) dia	780	3470	140–1500	620–6670	10	em	6
small area rounded surface							
0.67 in (1.70 cm) dia(2)	766	3410	400–1100	1780–4890	7	fr	1
Occiput Bone							
small area rounded surface							
0.67 in (1.70 cm) dia(2)	1440	6410	1150–2150	5120–9560	5	fr	1
Padded Surface Impacts							
Frontal Bone							
rubber padded rigid surface							
flat sheet, 90 durometer	2530	11 260	1200–3400	5340–15 100	7	em	2
1 in (2.5 cm) rad. cyl. 90 durometer aligned sagittally	1650	7340	1100–1960	4890–8720	6	em	2

1. These forces produced a variety of fracture patterns: see the source material referenced for a description of the fractures produced by each surface.
2. Static tests.

5.2 Brain Injury—The brain can be injured by processes other than the fracture/impingement mechanism described above. Excessive acceleration, by itself, can cause brain injury through a variety of effects, none of which are completely understood. Relative motion between the brain and the skull can induce a wide range of debilitating effects; the periphery of the brain can be contused, the blood vessels leading from the brain to the skull can be ruptured, internal brain matter can be sheared by relative motion between its parts, and the brain stem can be distorted by extrusion through the opening at the base of the skull. Finally, excessive tensile stresses can occur independent of any large brain displacement. This usually takes place opposite the impact site and can disrupt a variety of brain functions depending on its location. Little is known about the effects of multiple or long duration impacts.

5.2.1 CONCUSSION—In 1966, the Committee of the Congress of Neurological Surgeons defined brain concussion as:

"A clinical syndrome characterized by immediate transient impairment of neural function such as alteration of consciousness, disturbances of vision, equilibrium, etc., due to mechanical forces."

Concussion is usually a fully reversible injury. It has been widely studied for a number of reasons:

a. It is by far the most prevalent brain trauma.
b. Concussion is usually the first functional impairment of the brain to occur as the severity of head impact increases.
c. It accompanies 80% of all linear skull fractures (however the vast majority of concussions occur without skull fracture).
d. It is reproducible in experiments with animals whereas other brain injuries are not.

For these reasons, early brain injury studies were based on analysis of concussion rather than the more complex injury mechanisms described previously.

5.2.2 HISTORICAL DEVELOPMENT—Three different aspects of the gross skull motion have been suggested as being correlated to concussion:

a. Rotational acceleration.
b. Translational acceleration.
c. Flexion-extension of the upper cervical cord during motion of the head-neck junction.

Only the first two phenomena have received quantitative appraisals and are discussed here.

5.2.2.1 Rotational Acceleration—Leaders in this field have been Holbourn (Ref. 7), Gurdjian, et al (Ref. 8), Ommaya, et al (Ref. 9A), Unterharnscheidt, et al (Ref. 9B), Gennarelli, et al (Ref. 10), and Hirsch, et al (Ref. 11). In more recent years, most of the research into the effects of rotation have been conducted on animal brains, in vivo or in isolation. Hirsch's group has attempted to establish injury criteria and tolerance levels for rotational acceleration. Based upon results of experiments with several types of monkeys, they have employed scaling laws to apply tolerance levels to the human.

5.2.2.2 Translational Acceleration—The first version of the Wayne State University Concussion Tolerance Curve (Figure 2) was proposed by Lissner, et al in 1960 (Ref. 12). The abscissa is the duration of the effective part of the pulse spanning the principal impact. The ordinate is effective acceleration which is the average a-p[2] acceleration of the skull measured at the occipital bone for the principal part of the impact of the forehead against plane, unyielding surfaces. The curve was derived from the following observations:

a. It was observed clinically that linear skull fracture is usually associated with unconsciousness or a mild concussion (Gurdjian, et al. (Ref. 13)).

2. Anterior-posterior or front to back.

b. The acceleration levels and pulse durations necessary to cause skull fractures in cadaver heads were measured in free fall impacts against a rigid surface. These results were considered to approximate the human tolerance level for concussion from the correlation noted in item (a) above. The fracture data provided points for the curve in the range of 0.001–0.006 s.
c. Experimental animals were concussed by air pressure pulses of varying magnitudes and durations applied directly to the membranous covering of the brain (Gurdjian, et al. (Ref. 14)).
d. The pressure pulses measured in the parietal and temporal regions of cadaver heads in drop tests (Lissner, et al (Ref. 12)) and Gurdjian, et al. (Ref. 15)) were compared with the animal data in item (c), and the corresponding cadaver acceleration measurements were used to provide data points for the concussion curve between 0.006 and 0.010 s.
e. The long-duration end of the curve, with the asymptotic value of 42 G, was obtained from whole body volunteer data reported by Stapp (Refs. 16, 17). Patrick, et al, considered this value to be too low, since other volunteers had survived frontal crash simulations exceeding 45 G. They recommended that the value of the asymptote be raised to 80 G for padded impacts that avoid concentrated loads (Ref. 18). The resulting curve (Figure 2) became the accepted version of the Wayne State Tolerance Curve and is the basis of most current head injury criteria including the original U.S. Federal Motor Vehicle Safety Standard head impact specification (FMVSS 201) (Ref. 19).

Figure 1

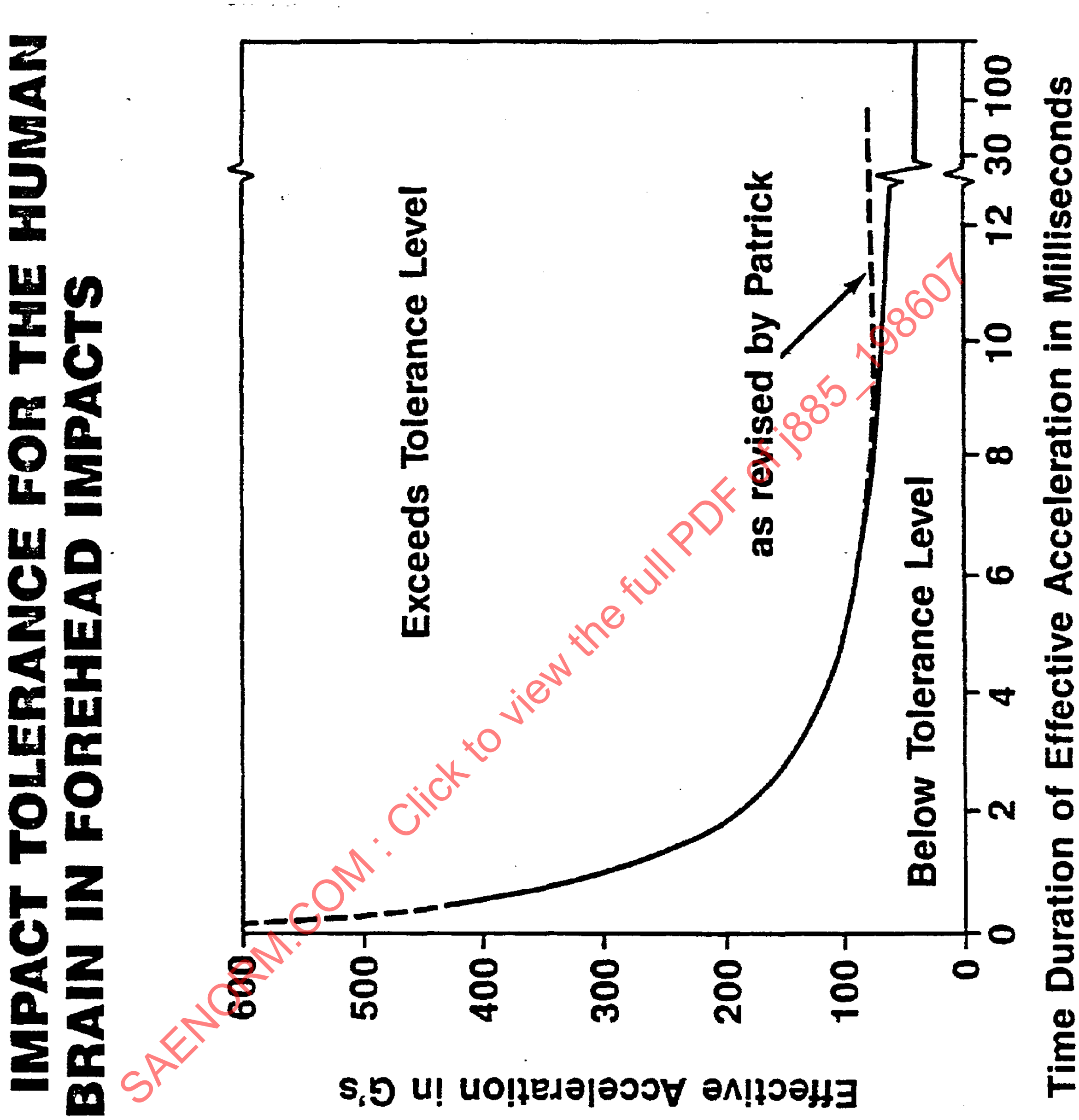


FIGURE 1—

Ono, et al, (Ref. 20) concluded that their results supported the Wayne State Concussion Tolerance Curve given in Figure 1. This conclusion was based on extensive microscopic examination of brain tissues and physiological measurements following translational and rotational impacts to the heads of sixty-three monkeys, and drops of fifteen human cadavers' skulls.

5.2.3 INJURY CRITERIA AND TOLERANCE LEVELS—The two principal criteria of brain injury are the Severity Index (SI) and the Head Injury Criterion (HIC). Numerous additional indices of brain injury have been proposed. Most of these are summarized in Ref. 21.

5.2.3.1 *Severity Index*—The WSU Tolerance Curve is difficult to apply to complex acceleration-time pulses because of uncertainties in determining the effective acceleration and time. To overcome this problem, Gadd (Refs. 22, 23) devised a weighted impulse criterion for establishing a Severity Index (SI):

$$SI = \int_O^T a^n dt \qquad \text{(Eq. 1)}$$

where:

a = acceleration in G's
n = weighting factor, 2.5 for head impacts
T = pulse duration
t = time in seconds

The weighting factor of 2.5 is primarily based on the slope of the straight-line approximation of the Wayne State Tolerance Curve plotted on log-log paper between 2.5 and 50 ms. A review of the mathematical derivation of the Severity Index by Versace (Ref. 26) details the relationship between the Wayne State Curve and the Severity Index. Gadd proposed a tolerance value of 1000 as the threshold of concussion for frontal impact (Ref. 22). This tolerance value was mandated in early versions of FMVSS 208; however, it specified that the Severity Index was to be calculated using the resultant acceleration measured at the CG of the head instead of the uniaxial acceleration measured on the occiput in the direction of the blow as was used by Gadd.

For distributed or non-contact blows to the head, Gadd (Ref. 24) has indicated that an SI value of 1500 would be an appropriate concussion tolerance level. Gadd cited the fact that Stapp experienced an acceleration pulse which equated to a true biaxial head exposure estimated to have reached a Severity Index of 1500. This occurred in a rocket sled run in which 45 G was measured on the seat. There was no brain injury in this exposure although retinal hemorrhages occurred.

Some success has been reported in employing the Severity Index concept to reduce brain injuries occurring to football players. Seventy-three to one hundred percent of the brain injury fatalities reported annually in this sport have been subdural hematomas. Beginning in 1970–1971, football helmets were designed to attenuate head impacts to an SI of less than 1500 in a simulation of a severe football head impact. The influence of this impact criterion is shown by a 50% reduction in fatality incidents (normalized) when comparing the post 1971 seasons to the preceding equivalent period (Ref. 25).

5.2.3.2 *Head Injury Criterion (HIC)*—Versace (Ref. 26) examined the relationship between the Wayne State Curve and the Severity Index. In response to this, a new parameter, the Head Injury Criterion (HIC) was defined by NHTSA as:

$$HIC = \left\{ (t_2 - t_1) \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} \right\} max \qquad \text{(Eq. 2)}$$

where t_1 and t_2 are the initial and final times (expressed in seconds) of the interval during which HIC attains a maximum value and a(t) is the resultant acceleration (expressed in G) measured at the head CG. The HIC replaced the SI in later versions of FMVSS 208 with a HIC value of 1000 being specified as the concussion tolerance level.

A point worth noting is a study by Hodgson and Thomas (Ref. 27) which concluded that the HIC interval (t_2 - t_1) must be less than 15 ms in duration in order to pose a concussion hazard even if the HIC value exceeds 1000. This finding resulted from an examination of events for which the concussive outcomes were known or could be inferred.

5.2.3.3 *Mathematical Models of the Brain*—Some researchers in brain injury have sought insights into its behavior by developing mathematical models of the brain and/or skull. Early investigators in this field employed continuum models. Much of this work is reviewed in Ref. 28. These simple models proved to be unsatisfactory and attention turned to finite element formulations. Ward and Thompson developed one of the more advanced models of this type (Ref. 29). More recently Ward et al proposed a brain injury criterion based on the intracranial pressure calculated from the model's response to input accelerations (Ref. 30).

5.2.3.4 *Lateral Tolerance of the Brain*—All of the preceding discussion is based on data obtained from head impacts in the a-p direction. Some lateral studies employing cadavers and primates have been reported by Stalnaker, et al (Ref. 31). They concluded that the threshold of irreversible closed skull brain injury to humans occurred when the translational head acceleration reached a peak of 76 G with a pulse duration of 20 ms.

Got, et al performed twenty-two lateral drop tests employing cadavers with pressurized brain arterial systems (Ref. 32). Seventeen of these specimens were helmeted, three were unhelmeted and struck padded surfaces, and two were unhelmeted and struck a rigid surface. Sixteen of these tests were considered to have produced useful results, with HIC values ranging from 900 to over 2000. For HIC values equal to or below 1500 (10 cases) two specimens exceeded an AIS injury level of 3; for HIC values equal to or below 1000 (two cases) one specimen received an AIS injury level of 0 and the other an AIS injury level of 2. It should be noted that the brain damage found with the use of brain pressurization procedures employing liquid dyes are typically arterial ruptures which are more serious injuries than the reversible concussion on which Figure 1 is based. The venous system was not pressurized reducing the chance of detecting a failure in its vessels. It should also be noted that neurophysiological damage cannot be detected in cadaver experiments.

Melvin et al (Ref. 33) investigated lateral impacts to unembalmed cadavers against rigid and deformable structures. They found, for head impacts against rigid walls, that brain damage of AIS 4 or greater began to occur at head impact speeds of 20 mph (33 km/h).

Nahum et al performed lateral impact tests to the heads of five cadavers using a padded impactor (Ref. 34). Arterial pressurization was employed with post test dissection showing subarachnoid hemorrhages on the lateral brain surfaces. HICs ranged from 1340 to 5246 with a mean of 2930.

5.3 Strength of Facial Bones—The principal facial bones are the mandible (lower jaw), maxilla (upper jaw), and the two zygomas (cheekbones). All are prominent and can be struck in a variety of locations and from a variety of directions. In addition, these bones can be loaded individually or collectively depending on the size, shape, and conformability of the impacted surface. The literature on facial bone fracture is limited, but the far greater part of it deals with the strength of the individual bones. The typical impactor employed was flat, circular, of 1 in^2 area (6 1/2 cm^2) and covered with little or no padding. Test impacts were usually delivered to the most prominent feature of the bone and essentially normal to it. A summary of facial bone fracture data is available in Table 2.

5.3.1 ZYGOMA—Four studies are available in which a 1 1/8 in (2.9 cm) diameter impactor was used to strike the zygoma. Three of these studies employed blows to the frontal portion of the zygoma (near its junction with the maxilla) while the fourth study addressed the mid-arch. The results of these four investigations were similar and their findings can be summarized by the results reported by Nahum, et al. (Ref. 35).

a. The minimal tolerance load was 200 lb (0.89 kN); their recommended level for a clinically significant fracture was 225 lb (1kN).
b. Embalming did not appear to affect results for the areas studied.
c. Thickness of the overlying soft tissue played an important role.

TABLE 2—DYNAMIC FRACTURE FORCES FOR FACIAL BONES

Bone	Impactor	Fracture Forces Mean lb	Fracture Forces Mean N	Fracture Forces Range lb	Fracture Forces Range N	Sample Size	Fresh or Embalmed	Ref
zygoma (1)	(2)	386	1717	138–780	614–3470	19	both	35
zygoma(1)	(3)	374	1665	208–640	925–2850	10	both	5
zygoma	(4)	283	1259	190–374	845–1665	5	embalmed	36
zygoma	(5)	516	2297	360–756	1600–3360	7	embalmed	36
zygomatic arch	(3)	345	1535	208–475	925–2110	17	both	5
maxilla	(3)	258	1148	140–445	623–1980	13	both	5
maxilla (lower)	(2)	(6)		175–210(7)	778–934	(6)	(6)	35
mandible (symphysis)	(2)	(6)		350–400(7)	1558–1780	(6)	(6)	35
mandible (midbody)	(2)	(6)		290–325(7)	1290–1445	(6)	(6)	35
mandible (center)	(3)	697	3100	425–925	1890–4110	9	both	5
mandible (lateral)	(8)	431	1918	184–765	818–3405	9	both	5

1. Impacted near the maxillary suture.
2. Flat rigid impactor, 1 1/8 in (2.9 cm) dia covered with MetNet pad. 0.2 in (0.5 cm) thick.
3. Same as footnote b except pad thickness was 0.10 in (0.25 cm).
4. Padded impactor, 1 1/8 in (2.9 cm) dia.
5. Patted impactor, 2 9/16 in (6.5 cm) dia.
6. Not reported.
7. Lower range of fracture values.
8. Flat rigid impactor 1 x 4 in (2 1/2 x 10 cm) covered with nickel foam pad 0.2 in (0.5 cm) thick.

In another zygoma study, Hodgson (Ref. 36) explored the effect of increasing the area of the impactor. He conducted paired tests of five cadavers; the zygoma on one side of the face was struck with a 1 1/8 in (2.9 cm) diameter impactor while the opposite zygoma was struck with a 2 9/16 in (6.5 cm) diameter impactor. The average fracture loads were 283 lb (1.26 kN) and 573 lb (2.55 kN) respectively.

5.3.2 MAXILLA—The maxilla is the weakest of the facial bones when the impact is directed to the thin bone covering the maxillary sinus. Schneider, et al (Ref. 5) reported that every one of the fractures in their maxilla study was "depressed and comminuted" due to breakage of this bone shell. They conducted thirteen impacts (producing eleven fractures) with a 1 1/8 in (2.9 cm) diameter flat impactor. Their average fracture force was 257 lb (1.15 kN) and their fracture range was 140–445 lb (0.62–1.98 kN). A previous study of the maxilla by Nahum, et al (Ref. 35) had reported a range of 175–210 lb (0.78–0.93 kN) as a "clinical fracture tolerance".

5.3.3 MANDIBLE—The size and shape of the mandible presents a wide range of impact possibilities. Schneider, et al (Ref. 5) noted an indeterminacy in delivering impacts to the center of the mandible. If the blow was directed towards the cranium, and the teeth were in contact, high forces could be sustained before failure occurred at the mandibular body or its symphysis. However, if the blow was directed towards the neck, the loading was carried primarily by the condylar processes (where the jaw articulates with the skull) which failed at lower loads. In this Schneider series, fractures occurred at all three locations; the fracture force levels for the nine specimens tested range from 425–925 lb (1.89–4.11 kN) with an average value of 639 lb (2.84 kN) for the six failures obtained. A previous study (Ref. 35) had found a "clinical fracture range" of 350–400 lb (1.56–1.78 kN) for impacts to the symphysis of the mandible.

Lateral impacts to the body of the mandible have been undertaken both with a 1 1/8 in (2.9 cm) diameter impactor and a 1 x 4 in (2 1/2 x 10 cm) rectangular impactor aligned along the body. The former study obtained "lower fracture values" of 290–325 lb (1.29–1.44 kN) (Ref. 35) while the latter study produced a fracture range of 184–765 lb (0.82–3.41 kN) and an average fracture load of 431 lb (1.92 kN) (Ref. 5).

5.3.4 FULL FACE—One study is available which indicates that the facial skeleton is remarkably strong when face contact occurs against a padded, deformable surface. Daniel and Patrick (Ref. 37) conducted 22 sled tests with lap-belted cadavers in an automobile body buck; head impact speeds ranged from 9–40 mph (4–18 m/s). Their test geometry was such that the cadaver heads typically struck the top of their padded instrument panel chin first; the head then rotated forward until full-face contact occurred. No facial bone injuries were found in this series. Head a-p accelerations were all below 60 G except for the single run at 40 mph (18 m/s); here the acceleration was 165 G.

5.4 Direct Impact to the Neck—The anterior portion (front) of the neck contains two stiff tissues which are delicate and vital. These stiff tissues, the thyroid and cricoid cartilages, are found at the upper end of the airway passage in the neck; hence their collapse can obstruct airflow. The thyroid cartilage is shaped like a wishbone with a relatively blunt apex. The apex (Adam's apple) faces anteriorly. The cricoid cartilage is immediately beneath the thyroid; it is ring shaped and completely encircles the trachea.

5.4.1 MELVIN DATA—The fragility of the thyroid and cricoid cartilages is illustrated by the data in Table 3. Melvin, et al., (Ref. 38), employed a Plastechon high-speed testing machine to conduct dynamic compression tests on eight excised, unembalmed human larynges. They found that incipient cracking occurred at a mean load of 40.6 lb (181 N) for the thyroid cartilage and 55.5 lb (247 N) for the cricoid when each was loaded separately. As a part of this program, both cartilages were also loaded simultaneously to very large deflections (one-half their original dimension) through the use of a 1 1/2 in (38 mm) diameter flat plate. For this situation, the force increased to a mean level of 110 lb (490 N). It should be noted that this 50 percent deflection represents a very serious fracture level at which total collapse of the larnyx was imminent.

5.4.2 GADD DATA—Another larynx study by Gadd, et al. (Ref. 39), tested unembalmed human subjects with the larynx in situ and obained somewhat higher loads than Melvin. The Gadd program employed an instrumented drop weight of 1 in^2 (6 1/2 cm^2) area and produced marginal fractures of either the thyroid or cricoid cartilage at dynamic loads of 90–100 lb (400–450 N).

5.5 Neck Injury Due to Head Inertia Loading—In automobile collisions, neck injuries can occur as a result of its bending from head inertial loading. When the torso is violently accelerated (or decelerated), large, potentially injurious neck loads and deflections are generated by the inertia of the head.

Neck bending can occur in any direction. In medical terminology, backward bending of the neck is called extension; forward bending of the neck is termed flexion, sidewards bending of the neck is called lateral flexion; the "no" gesture of the head is termed rotation.

TABLE 3—DYNAMIC FRACTURE LOADS FOR THE THYROID AND CRICOID CARTILAGES

Cartilage	Dynamic Fracture Loads Mean lb	Dynamic Fracture Loads Mean N	Dynamic Fracture Loads Range lb	Dynamic Fracture Loads Range N	Nature of Fracture	Ref
excised thyroid	40.6	180	14–85	62–377	incipient cracking	38
excised cricoid	55.5	248	35–68	156–302	incipient cracking	38
thyroid and cricoid loaded simultaneously	100	490	76–182	337–810	imminent total collapse	38
thyroid in situ	90–100	440–445			marginal fracture	39
cricoid in situ	90–100	400–445			marginal fracture	39

5.5.1 NECK STRUCTURE—The neck skeleton consists of seven cervical vertebrae. These vertebrae are generally referred to by number in order from top to bottom as C-1, C-2, etc. No two cervical vertebrae are identical; however, C-3 through C-7 are quite similar to one another. Adjacent vertebral bodies are separated by cartilagenous tissues called intervertebral discs. Vertebral articulations are stabilized by fibrous connecting tissues called ligaments. These ligaments also limit the degree of relative motion between the vertebrae.

Relative movement of the cervical vertebral column and the head is accomplished through muscle pairs which are attached to the skull, the individual vertebrae, and the torso through tendons. These pairs, which are symmetric on the right and left sides of the body, respond in various group actions to produce the desired movement of the head and neck. Muscle pairs which produce voluntary flexion are the ones which resist extension, and vice versa.

The muscles lying behind the head/neck are more massive than those lying in front; in addition, the former are located further from the head-neck pivot (the occipital condyles). Consequently, larger moments can be developed for resisting flexion than for extension. Also, a lower resultant muscle force level is required to produce the same magnitude of resisting bending moment in flexion than would be required in extension.

5.5.2 INJURY MECHANISMS

5.5.2.1 Hyperextension Injuries and Associated Mechanism—The rear-end collision accounts for most of the diagnosed neck injuries that occur to vehicle occupants. The resultant neck lesions are generally classified as hyperextension trauma and include symptoms such as localized neck pain, pain radiating to the shoulders, vague aches, discomfort, and vertigo due to strained muscles, damaged ligaments, injured articular joints, or fractures of various parts of the cervical vertebrae. The involvement of the cervical vertebrae, joints, connecting ligaments, and muscles in a rear-end collision environment can be qualitatively analyzed but are difficult to quantify.

If the head is turned to one side at the onset of a rear-end collision, the neck ligaments will be prestrained and less articulation of the neck will be required to produce high resistive forces. Consequently, there will be less time available for the neck muscles to respond to aid in accelerating the head, placing a greater burden on the ligaments. For this condition, the neck can be more susceptible to injuries.

5.5.2.2 *Hyperflexion Injuries and Associated Mechanisms*—Hyperflexion neck injuries to lap/shoulder belted occupants have not been reported with any degree of frequency in field accident studies. A study by Vazey and Holt (Ref. 40A) of fatalities to car occupants wearing lap/shoulder belts indicated that only two out of 136 fatalities were due to neck injuries. In these two cases, the occupant compartments were severely compromised. The involvement of the shoulder harness in producing these two neck injuries is doubtful. In contrast, Schmidt et al (Ref. 40B) conducted a series of one hundred simulated frontal collision sled tests using lap/shoulder belted cadavers as vehicle occupants. Sled impact speeds of 30, 40 and 50 km/h were used. The deceleration-time curve was trapezoidal with plateau deceleration levels ranging between 16.9 to 25.6 G's. The cadavers were unembalmed with an age distribution at time of death ranging from 12 to 83 years. While the most frequent cadaver damage observation was rib fracture, 46 of the 100 cadavers had neck damage with most of this damage being concentrated at the level of C-7 and T-1.

There are a number of possible explanations for the difference in frequency of field and experimental neck flexion damage. The neck structure of the unembalmed cadaver is flaccid. Its neck muscle cannot transmit any significant load even when the neck is hyperflexed. All the neck loads must be transmitted by the bony vertebrae, the intervertebral discs and the surrounding joint ligaments. In the living human, neck muscles can transmit loads, sharing the load distribution with other neck structures. The implication of this difference in the load carrying structures of the cadavers and the human neck is that while the cadaver will mimic the neck damage patterns of the human, the cadaver damage will occur at lower collision severity levels. On the other hand, published field accident data of neck injury may not represent the actual frequency of neck injuries since detailed autopsies of the neck are not routinely performed. Also, in the more severe frontal collision environments, the head of the belted occupant may impact a part of the forward interior. This is particularly true of the driver. In such cases, the head load causes a redistribution of the neck loading which could reduce the potential for neck injury.

Shear forces in the neck are important in flexion prior to the chin contacting the chest. For the vertebrae C-3 through C-7, there are bone-to-bone interlocking joint surfaces and ligaments to carry these shear forces as the neck is flexed. This is not the case for the upper neck joints (occipital condyles/C-1 and C-1/C-2). Here the ligaments must carry the shear loads. These upper joints are, therefore, the most likely to be injured by shear.

When the chin contacts the chest, a redistribution of the loading occurs. Chin-chest contact causes a lower force level to be developed in the posterior muscles for the same magnitude of resisting bending moment. In addition, the force on the chin has a component which is parallel to the shear force developed by the neck and aids in decelerating the head. Transfer of loading from the ligaments of the neck to the chin reduces the shear load transmitted between the head, C-1, and C-2 and reduces the probability of injury in these areas.

As the neck flexes, the front portions of the intervertebral discs are compressed. Lesions to the discs may result if the compressive forces become sufficient. Also, the anterior portions of the vertebral bodies may be fractured. The ligaments posterior to the articular surfaces can be torn during hyperflexion. In particular, the ligaments joining adjacent spinous processes are prime candidates for lesions since they undergo the greatest elongation. The ligamentous and muscle loads may fracture the spinous processes or parts of the vertebrae surrounding the spinal cord.

5.5.2.3 *Lateral Flexion Injuries and Associated Mechanisms*—Lateral flexion injuries occur less frequently than the other two types of neck injuries. Usually in a lateral (side impact) collision, severe lateral flexion of the neck does not occur. For a far side collision, the upper torso is accelerated but may be free to rotate towards the impacted side, minimizing the neck forces required to accelerate the head. For a near side collision, the torso is accelerated upright, but the head usually impacts the side door window or upper side structures minimizing the neck forces. If severe lateral flexion should occur, ligamentous injury and/or fractures of the articular processes of the vertebrae may be found at the C-5 to C-7 level.

5.5.3 NECK STRENGTH—To obtain measures of each injury mechanism which was discussed previously would be a difficult, if not an impossible task. First, volunteers cannot be exposed to injury-producing environments; second, relevant in vivo measurements cannot usually be made. Consequently, indirect approaches must be used to obtain data which can be related to the overall strength of the neck.

Three approaches have been used to obtain neck strength data. Static strength tests on necks have been conducted with volunteers resisting static loads applied to their heads. Dynamic tests have been conducted where volunteers have been subject to controlled, non-injurious acceleration environments. In these latter tests, the torso is restrained and the head is accelerated by the neck. A third approach utilizes human cadavers in a similar manner to the dynamic volunteer tests, except that the severity of exposure can be increased until physical damage to the neck structure is produced.

In dynamic neck tests, it is common practice to measure accelerations of the head and the angular position of the head relative to the torso; in static tests the usual measurement is the force applied to the head. Investigators have noted deficiencies in relating injury severity to these measurements. For the static tests, the applied head load does describe the force level the neck must resist, but does not directly define the resisting bending moment the neck must develop. The same is true of the accelerations measured in the dynamic tests. In an effort to minimize these deficiencies, Mertz and Patrick (Ref. 41) developed a method for calculating the resultant reactions developed between the top of the neck and the base of the skull (occipital condyles) for both the static and dynamic approaches. This method allows direct comparisons to be made of static and dynamic neck reactions.

The angular position of the head relative to the torso can be used as a measure of the severity of neck bending. However, it should be noted that the neck can be injured without exceeding its static angular range of motion. In addition, when the neck is flexed to an extreme of articulation, relative angular position becomes a poor measure of potential injury because a small increase in articulation, which is difficult to measure accurately, will produce large increases in neck loads. Measures of the neck loads may be a better indicator of injury potential.

5.5.3.1 Static Strength of the Neck—Mertz and Patrick (Refs. 41, 42) and Patrick and Chou (Ref. 43) have conducted tests on volunteers to determine the neck's reaction on the head for statically applied loads to the head. The principal results of these studies are summarized in Tables 4 and 5. Table 4 is a summary of the maximum static bending moments developed at the occipital condyles for various loading conditions. The maximum shear and axial forces that were observed are given in Table 5.

TABLE 4—MAXIMUM STATIC BENDING MOMENT DEVELOPED AT THE OCCIPITAL CONDYLES BY VARIOUS VOLUNTEERS FOR VARIOUS LOADING CONFIGURATIONS (AIS = 0) BENDING MOMENT DEVELOPED AT OCCIPITAL CONDYLES

ft-lb (N˙m)

Neck Position	Resist Flexion-0 deg	Resist Extension-180 deg	Resist Lateral Flexion-90 deg	Resist Flexion-45 deg	Resist Flexion-135 deg
Normal	37.0 (50.2)	15.0 (20.3)	35 (47.5)	40.5 (54.9)	24.0 (32.5)
Flexed[1]	30.0 (40.7)	28.0 (38.0)	45.5 (61.7)	46.0 (62.4)	40.0 (54.2)
Extended[2]	29.5 (40.0)	17.5[3] (23.7)	38.0 (51.5)	34.5 (46.8)	27.0 (36.6)

NOTE—These values are not necessarily upper bounds of tolerable neck bending moments. Tests could have been terminated for reasons other than reaching a limit on forces producing the resistive bending moment. For example, the moment arm could be progressively decreased as the neck bends due to increasing load. Thus, the magnitude of the applied load could increase, but the resistive bending moment could decrease.

1. Initial head position toward the applied load.
2. Initial head position away from the applied load.
3. Value taken from Ref 41.; all other values taken from Ref. 43.

TABLE 5—MAXIMUM STATIC FORCE REACTIONS DEVELOPED AT THE OCCIPITAL CONDYLES BY VARIOUS VOLUNTEERS FOR VARIOUS LOADING CONFIGURATIONS (AIS = 0)

Shear Force lb (N)	Shear Force lb (N)	Shear Force lb (N)	Shear Force lb (N)	Shear Force lb (N)	Axial Force lb (N)	Axial Force lb (N)
A-P	P-A	R-L L-R			Tension	Compression
0 deg	180 deg	90 deg	45 deg	135 deg		
190 (845) [1]	190 (845)[1]	90 (400)[2]	98 (436)[2]	96 (427)[2]	255 (1134)[1]	250 (1112)[1]

NOTE—These values are not necessarily upper bounds of tolerance load reactions between the head and the neck at the occipital condyles. Tests could have been terminated due to discomfort with the strapping used to apply the load to the head.

1. Values taken from Ref. 41
2. Values taken from Ref. 43

It should be noted that none of these are necessarily upper bounds of non-injury load reactions between the head and neck at the occipital condyles. Tests were usually terminated due to discomfort with the straps used to apply the load to the head. No injuries or neck pain occurred as a result of any of these loads. They are considered non-injurious neck reactions and correspond to an Abbreviated Injury Scale (AIS) rating of zero.

Gadd, et al. (Ref. 39), subjected human cadavers to static rearward and lateral neck bending loads. They noted that minor ligament injury occurred for 80 degrees of rearward neck bending and 60 degrees of lateral neck bending.

5.5.3.2 *Dynamic Strength of the Neck*—Mertz and Patrick (Ref. 41, 42) and Patrick and Chou (Ref. 43) have also conducted tests on volunteers and human cadavers to determine the neck's reaction on the head under dynamic conditions. The principal results of these studies are given in Table 6 for volunteers and in Table 7 for human cadavers. The bending moment for forward flexion includes the moment of the chin force taken with respect to the occipital condyles.

Mertz and Patrick (Refs. 41, 42) found that the resultant bending moment was an excellent indicator of neck strength. Based on their cadaver data, they suggested tolerance levels for the 50th percentile adult male. For flexion, a resultant bending moment of 140 ft-lb (190 N-m) was proposed as a lower bound for an injury tolerance level. This bending moment did not produce any discernible ligamentous damage to a human cadaver. For extension, an injury tolerance level of 42 ft-lb (57 N-m) was suggested. This level was associated with ligamentous damage to a human cadaver. However, it should be noted that the human cadaver was relatively old and, also, there may have been degeneration of the strength of the ligamentous tissue compared to living tissue. Based on these suggested bending moment tolerance levels, the neck appears to be at least three times stronger in resisting flexion than extension.

Ewing and Thomas (Ref. 44) have also conducted dynamic forward neck bending tests with instrumented volunteers. Their testing has been directed at obtaining neck response data, not tolerance data. However, for some of their more severe test conditions, they have calculated the maximum forward neck bending moments relative to the occipital condyles. Three of the volunteers developed maximum bending moments of 22.5 ft-lb (35 n·m), 33.2 ft-lb (45 n·m) and 36.9 ft-lb (50 n·m) without any pain. These values are consistent with the forward bending results of Mertz and Patrick (Ref. 41) given in Table 6 where neck pain, an AIS = 1 injury, was observed at 65 ft-lb (88.2 n·m).

Nyquist, et al. (Ref. 45), subjected an instrumented dummy (Hybrid III, Ref. 46) to simulated accident environments of lap/shoulder belted occupants. The dummy was instrumented to measure the resultant neck loadings at the interface between the head and neck. For each simulated accident condition, the type and severity of the expected neck injury were inferred from the field accident injury data. Nyquist measured a neck forward bending moment of 110 ft-lb (152 N·m) along with an a-p neck shear load of 670 lb (2.97 kN) and a neck axial tension load of 740 lb (3.29 kN) in a test condition associated with an AIS = 1 neck injury level. Environments which produced serious neck injury were not simulated in this study. The limitations of restaging field accidents are discussed in Section 6.2.2.

TABLE 6—TOLERABLE NECK REACTIONS CALCULATED AT THE OCCIPITAL CONDYLES FOR DYNAMIC VOLUNTEER TESTS

Loading Configuration	Ref	Neck Bending Moment ft-lb	Neck Bending Moment (N·m)	Neck Shear Force lb	Neck Shear Force (N)	Axial Force lb	Axial Force (N)	Head Angle Relative to Torso deg	AIS Rating	Comments
Forward bending 0 deg	41	65.0	(88.2)	177	(787)	-	-	70	1	Pain but no injury
Rearward bending 180 deg	43	22.5	(30.5)	52	(231)	56	(249)	68	0	No injury
Lateral bending 90 deg	43	33.3	(45.2)	178	(792)	-	-	43	0	No injury
Lateral bending 135 deg	43	13.3	(18.0)	70	(311)	80	(356)	-	0	No injury
Lateral bending 45 deg	43	23.0	(31.2)	99	(440)	37	(165)	-	0	No injury

NOTE—These values are not necessarily upper bounds of tolerance load reactions between the head and the neck at the occipital confyles. They are all tolerable loads.

TABLE 7—NECK REACTIONS CALCULATED AT THE OCCIPITAL CONDYLES FOR DYNAMIC HUMAN CADAVER TESTS

Loading Configuration	Ref.	Neck Bending Moment ft-lb	Neck Bending Moment (N·m)	Neck Shear Force lb	Neck Shear Force (N)	Axial Force	Head Angle Rrelative to Torso deg	AIS Rating	Comments
Forward bending 0 deg	41	140	(190)	357	(1588)	—	88	0	No damage
	41	130	(176)	437	(1944)	—	69	0	No damage
Rearward bending 180 deg	35	35	(47)	—	—	—	—	0	No damage
	35	42	(57)	—	—	—	—	3	Ligamentous damage

5.6 **Neck Injury Due to Head Loading—**The neck can be injured by loading of the head. During head loading, some or all of the head load is transmitted to the torso by the neck structure. The magnitude of the transmitted load is dependent on the location and direction of the head load, the inertia of the head, and the configuration of the cervical spine when the head load is applied. For example, if the neck is straight when a fore/aft or lateral head load is applied, then the neck may undergo significant bending prior to transmitting large neck loads to the torso. However, if a load is applied to the head colinear with the cervical spine and the neck is straight, large tensile or compressive neck loads may be transmitted to the torso with little neck distortion.

Hodgson and Thomas (Ref. 47) discussed the effect of neck configuration on the magnitude of axial compressive loads transmitted by the neck, and the location and type of neck injuries for impacts to the top of the head. They measured bone strains on the anterior surfaces of the third, fifth and seventh cervical vertebrae of human cadavers for various neck articulations. For a given applied head load, the anterior cervical body strains were the lowest when the vertebrae were aligned; i.e., neck straight. This implies that the neck behaved as a column and that the neck compressive load should be a good indicator of the potential for neck injury. When the neck is flexed, the cervical vertebrae are subjected to a combined axial compression and bending moment. For this condition, the axial compression load alone may not be a good indicator of the potential for neck injury.

Culver, et al (Ref. 48), subjected human cadavers to superior-inferior head impacts. The necks of the cadavers were not flexed and a padded impactor was used to preclude skull fractures. A summary of the peak applied head load and observed neck trauma is given in Table 8. The mean axial compressive peak head load producing neck trauma was 1620 lb (7.22 kN), and the range was 1060 lb (4.71 kN) to 1990 lb (8.85 kN).

Crown impact tests were conducted by Nusholtz, et al (Ref. 49). In these tests, the thickness of the padding covering the impactor surface was varied to give different force-time characteristics. A summary of the peak applied head loads and resulting neck damage is given in Table 9. The peak head loads which produced neck damage had a range of 405 lb (1.8 kN) to 2495 lb (11.1 kN) and a mean value of 1210 lb (5.4 kN). The authors concluded that the initial configuration of the cervical spine had a major influence on the load carrying capacity and damage patterns which were observed. Note that both Culver, et al., and Nusholtz, et al., measured applied HEAD loads. The axial compressive NECK loads corresponding to these applied HEAD loads could be smaller due to head mass inertial effects.

TABLE 8—COMPRESSIVE HEAD LOADS AND RESULTING NECK DAMAGE DESCRIPTIONS FOR SUPERIOR-INFERIOR HEAD IMPACTS TO CADAVERS WITH NECKS STRAIGHT (48)

Axial Compressive Head Load lb	Axial Compressive Head Load N	Neck Injury Description
1490	6620	No Fractures
1510	6700	No Fractures
1560	6950	No Fractures
1060	4710	Spinous processes of C4, C5, C6 fractured. Transverse process of C5 fractured. Body of C5 crushed on right side.
1360	6050	Tips of spinous processes of C3, C4, C5 fractured.
1580	7030	Spinous processes of C7 and T1 and both transverse processes of T1 fractured. Right transverse process of C7 crushed.
1620	7200	Body of C5 fractured.
1680	7450	C5–6 disk crushed. Spinous process of C2 fractured from body at arches. Tip of spinous process of C6 fractured.
1800	8000	Spinous processes of C1, T1, T2 fractured through arches. Tips of spinous processes of C2, C4, C7 fractured.
1900	8450	Complete fracture from body of C3 and C4 left transverse processes. Chip fractures of spinous processes of C5, C6, C7, T2.
1990	8850	C3–4, C4–5, C5–6 disks crushed. Transverse processes of C5 and T1 fractured, body of T2 severely crushed.

TABLE 9—COMPRESSIVE HEAD LOADS AND RESULTING NECK DAMAGED DESCRIPTIONS FOR SUPERIOR-INFERIOR HEAD IMPACTS TO CADAVERS WITH VARIOUS HEAD-NECK-TORSO ANGLES (49)

Axial Compressive Head Load lb	Axial Compressive Head Load N	Neck Damage Description
405	1800	Fracture of C6 spinous process. Fracture of C7 lamina, articular process and body. Rupture of anterior and posterior longitudinal lagaments and ligamentum flava between C6–C7 with disk involvement.
495	2200	Extension/compression type damage. (osteoporotic)
515	2300	Fractures of C5 and C6 bodies. Fracture of C5 spinous process. Fracture of C6 lamina. Rupture of anterior longitudinal ligament between C5–C6 with disc involvement.
740	3300	Fractures of C3–C7 spinous processes. Fracture of C2–C3 disc with displacement. Rupture of anterior longitudinal ligaments between C3–C4 and C5–C6 with disc involvement. Rupture of C2 posterior longitudinal ligament.
740	3300	Fractures of C3–C4 spinous processes. Fracture of C4 articular capsule. Rupture of longitudinal ligament between C3–C4 with disc involvement.
1280	5700	Fracture of C4 spinous process. Rupture of anterior longitudinal ligament between C2–C3 with disc involvement.
1350	6000	Fractures of C5 and C7 bodies. Rupture of anterior longitudinal ligament at C5. Fracture T2 body.
1395	6200	Fractures of C7 and T1 spinous processes. Rupture of anterior longitudinal ligament between C6–C7 with disc involvement.
1595	7100	Bilateral joint laxity between C4–C5. Fracture of T4 body. Rupture of inter - and supraspinous ligaments between T3–T4.
2315	10300	Rupture of supra - and inter spinous ligaments and ligamentum flavum between C7–T1 with disc involvement. Fracture of T1 body.
2495	11100	Rupture of anterior longitudinal ligament between C3–C4 with disc involvement. Fracture of T3 body.
630	2800	No damage

In an accident reconstruction program, Mertz et al. (Ref. 50), exposed an instrumented Hybrid III dummy to head impacts using a spring propelled, tackling dummy which has produced serious neck injury to football players. In these tests, the dummy was oriented so that the load was applied to the top of the head, loading the neck structure in compression with minimal head rotation. This configuration was chosen to produce the maximum value of neck compression force for the impact velocity used. The neck compressive load measured by the Hybrid III dummy should be representative of the upper bound of the maximum axial compressive load that an equivalent weight human would experience for the same impact velocity since the relatively soft stiffness of the tackling dummy bag should mask the effect of the relatively stiff neck response required to obtain a maximum compressive loading. Based on their results, they proposed a time-dependent, injury criterion for axial compressive neck loads, Figure 2. Exceeding the criterion implies that major neck injury (permanent impairment of a body function) is likely. However, being below the criterion does NOT imply that major neck injury will not occur if other neck loading modes are present.

5.7 Thorax—The human thorax (or chest) is a ribbed shell (rib cage) containing the following important organs: heart, lungs, trachea, esophagus, and major blood vessels. The size and shape of the thorax depends 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. The rib cage is a semi-rigid structure which provides protection to the internal organs and facilitates the mechanics of respiration.

5.7.1 THORACIC INJURIES—Thoracic injuries may be divided into two types: (a) injuries to the internal thoracic organs and (b) injuries to the rib cage. Injuries to the internal organs include pneumothorax[3], hemothorax [4], ruptures of the heart, ruptures of the arteries connected to the heart, injury to the cardiac muscle, lung contusion, bruising, or rupture. Of these, the most frequent and most serious is the rupture of the thoracic aorta which is the major artery attached to the heart. Cardiac injuries are thought to be caused by compression of the heart between the spinal column and the sternum (breast bone). There is an increased possibility of cardiac rupture if the heart is in that portion of its pumping cycle where it is full of blood. Tears of the aorta usually occur just beyond the aortic arch at its junction with the subclavian artery. The tears are usually transverse to the vessel axis. The exact mechanism of failure is not yet understood. Injuries to the rib cage include fractures of the ribs and sternum, and less often, dislocations and fractures of the thoracic vertebrae. Rib fractures become dangerous if the broken rib ends are displaced to the point where they can puncture internal organs or are numerous enough to inhibit adequate inspiration.

3. These denote, respectively, free air blood in the sac surrounding the lung tissue
4. These denote, respectively, free air blood in the sac surrounding the lung tissue

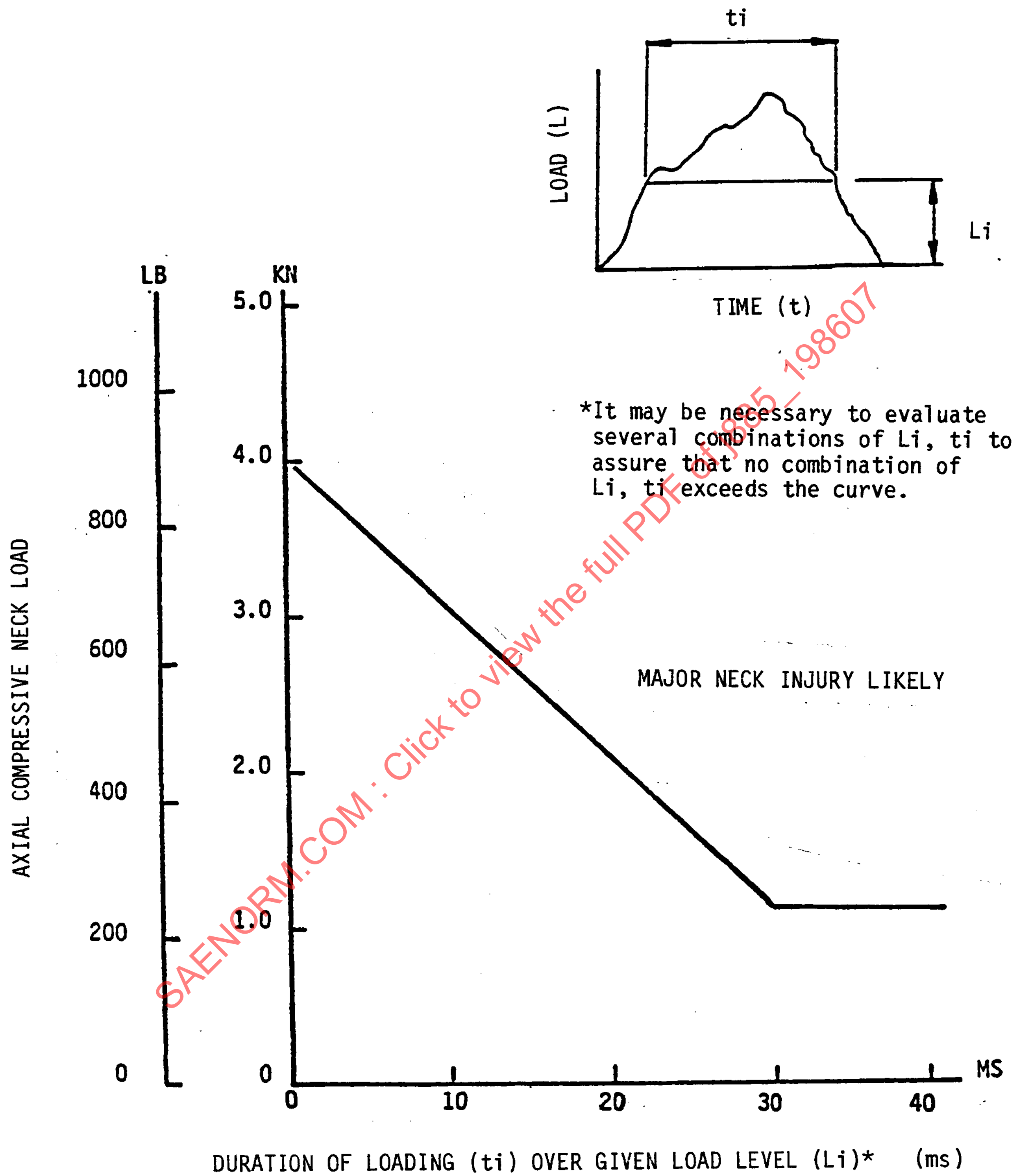


FIGURE 2—TIME DEPENDENT INJURY CRITERION FOR AXIAL COMPRESSIVE NECK LOADS, NECK STRAIGHT (REF. 48)

5.7.2 THORACIC INJURY CRITERIA—Several parameters have been suggested for monitoring the effect of a blow to the thorax based on thoracic acceleration, force, deflection, or some combination of these. Chest impact studies have been conducted by a number of researchers using both embalmed and unembalmed cadavers, experimental animals (monkeys, dogs, and pigs) and, for quasistatic chest loading, volunteers. There are difficulties associated with the use of any of the above test subjects in determining thoracic injury tolerance. With cadavers, corrections may be needed to account for their lack of muscle tone and lung inflation as compared to living subjects. Rib fractures are a commonly employed measure of thoracic injury with cadavers but this factor is highly age dependent. Data obtained with tests on animals must be scaled to account for size and shape differences of their thoracic cage as compared to the human; injury interpretation is complicated by anatomical differences between the human thoracic organs and those found in experimental animals.

The majority of experimental chest impact studies have involved frontal impacts employing either simple impactors or belt restraint systems. A lesser amount of data are available on lateral chest impacts. There are no data available at this time on oblique impacts to the chest.

5.7.2.1 Chest Deflection—Researchers have generally concluded that chest deflection is a response measure which shows good correlation with chest injury produced by blunt frontal impacts. Neathery, et al (Ref. 51) has been a major advocate of this approach. He analyzed test results on 24 cadavers that had received frontal blunt thoracic impacts delivered by a simple impactor. Employing regression analysis, Neathery related their chest traumas (using the 1971 AIS scale) to their chest deflection (normalized by chest depth) and ages at death.

Based on an injury level of AIS 3 (severe; not life-threatening), and a median driving age of 45 years old, Neathery recommended the following sternal deflection limits:

TABLE 10—

Occupant Size	Recommended Sternal Deflection Limit for AIS 3 mm	Recommended Sternal Deflection Limit for AIS 3 in
5th percentile female	60	2.36
50th percentile male	75	2.95
95th percentile male	90	3.54

The preceding recommendation for the 50th percentile male is consistent with observations made previously by Melvin, et al (Ref. 52). Melvin proposed a chest deflection limit of 1.75 in (44 mm) if rib fracture was to be avoided; he also concluded that a deflection range of 2.5–3.0 in (64–76 mm) would correspond to an AIS injury level of 3.

In a subsequent re-analysis of the data used by Neathery, Viano (Ref. 53) emphasized the distinction between skeletal and non-skeletal thoracic injuries. By separating these injury classes, Viano noted that internal injuries (which can be life-threatening) only began to appear at P/D ratios[5] of approximately 0.40. At this deflection level, the rib cage has lost its structural integrity due to multiple rib fractures. Viano's P/D limit of 0.40 is slightly greater than Neathery's recommended limit of 0.387; however, Viano's limit represents the onset of life-threatening injuries, (AIS = 4) whereas Neathery's limit represents an AIS level of 3 which is serious, but not life-threatening. It should also be noted that Viano's analysis did not correct for the effects of age levels as did Neathery's.

5. P/D = penetration divided by pre-impact chest depth

The primary disadvantage of the deflection criterion is the difficulty of performing the measurement, both on biological specimens as well as on test devices. A further complication is that a single deflection measurement is not generally representative of the complete thorax deformation behavior, unless the nature and location of the impact is well understood beforehand and the transducer positioned accordingly.

5.7.2.2 *Chest Acceleration*—The practical difficulties of the deflection criterion have led many researchers to conclude that acceleration measurements offer an attractive alternative. Stapp (Ref. 54) reports on numerous tests where volunteers were subjected to decelerative restraint environments. For a series of "rocket sled" tests where the volunteers were restrained by air force restraint harnesses, cardiovascular shock (drastic drop in blood pressure immediately post test) was noted in several tests where the peak sled deceleration ranged from 26–38.5 G with deceleration onsets of 896–1373 G/s. Unfortunately, these subjects were not instrumented with chest accelerometers.

Mertz and Gadd (Ref. 55) report that an instrumented stunt man experienced chest accelerations of 46 G's while impacting a thick foam mattress with his back after diving 57 ft (17.4 m) from a tower. Viano, et al. (Ref. 56), measured the chest acceleration of a performer who routinely dove from a height of 34.5 ft (10.5 m) into a shallow pool, impacting the water's surface with his belly. They measured thoracic spine and sternal accelerations of 25 G and 224 G, respectively, for a 15 ft (4.6 m) dive. The authors extrapolated these measured results to 68 G and 380 G, respectively, for his normal performance height.

FMVSS 208 currently specifies as acceptable any acceleration pulse which "... shall not exceed 60 G except for intervals whose cumulative duration is not more than 3 ms". Previously, MVSS 208 had applied a Severity Index to the chest acceleration pulse. This index was calculated in exactly the same manner as the head Severity Index discussed previously, and the limit of 1000 was the same as that for the head. Both the 60 G and the chest Severity Index limits are based on the resultant acceleration measured at the center of gravity of the dummy thorax.

Accelerations measured at a single point (as described above) cannot adequately represent the complete response of the thorax. For this reason, Robbins, et al (Ref. 57) employed a sophisticated approach to accelerometer usage for determining the overall response of the thorax. His group performed tests on animals and cadavers instrumented with ten accelerometers mounted at eight different locations on their rib cages and backbones. Test conditions included frontal and, subsequently, lateral impacts (Ref. 58). The instrumentation, which was later increased to twelve accelerometers, was chosen to be consistent with dummy instrumentation usage. Their intent is to determine a predictive function which will enable these accelerometer signals to be combined in a manner that is related to the thoracic injury. This approach requires a large number of tests and the use of a computer to generate regression models.

5.7.2.3 *Shoulder Belt Load*—In an analysis of data from 108 frontal tests with seat belted cadavers conducted at five research institutes, Eppinger (Ref. 59) formulated an equation which predicts the number of observed thoracic fractures (this includes fracture of the ribs, sternum, and clavicles) based on the maximum upper torso belt force, the cadaver weight, and the cadaver age at death. As an example of the application of this method, Eppinger chose to use the age and weight distributions of the U.S. automotive fatality population in a particular 30 mph (13.4 m/s) frontal crash with a particular belt restraint system. From this he determined that the total number of rib fractures to the target population would be minimized if shoulder belt forces could be limited to 1300–1500 lb. Eppinger employed a 12 in (305 m) stroke limit on his belt system which precluded a lower optimal force level.

Foret-Bruno, et al (Ref. 60), reported on the relationship between vehicle occupant thoracic injuries and shoulder belt loads estimated from frontal accidents in which the occupant was restrained by a unique energy absorbing lap/shoulder belt system which allowed the shoulder belt load to be approximated. No chest injuries were received by occupants less than 30 years old for shoulder belt loads under 1650 lbs. (7.30 kN). Beyond age 50, fractures began to occur at about 950 lbs. (4.20 kN) of belt load. The authors compared these results to Eppinger's analysis and concluded that cadavers could be expected to sustain three to five more rib fractures than the living crash victim under similar impact conditions.

The disadvantage of employing shoulder belt load as an injury criterion lies in its sensitivity to shoulder belt geometry. The force on the torso is not only a function of the belt loads, but also is dependent on the angles of the belts relative to the torso. These belt angles can be expected to vary among belt restraint systems since they are a function of such variables as anchorage locations, seat height, seat stiffness, and webbing properties. Belt angles can also be expected to change with the occupant's movement during the impact event.

5.7.2.4 *Lateral Loading*—Some recommended limits for side impact to the chest come from an animal and cadaver study by Stalnaker (Ref. 61). Two impact surfaces were considered in this investigation. A flat 6-in (15.2 cm) diameter surface was employed for blunt impacts and a simulated armrest was employed for concentrated impacts. Stalnaker concluded that a lateral chest deflection of 31 percent of chest width produced by the blunt surface would result in an AIS injury level of 3; the comparable non-fracture limit was found to be 22 percent.

Tarriere, et al. (Ref. 62), investigated the tolerance of the thorax to lateral impacts by dropping unembalmed cadavers, suspended horizontally, against a broad, flat load cell surface. Rigid and padded conditions were employed. The cadaver thorax instrumentation included triaxial accelerometers and a deflection rod installed transversely through the rib cage and viewed photographically. The predominant traumfa was rib fractures with no visceral lesions being found. Mineralization tests were conducted on rib samples, post-test, to determine the suitability of the specimens employed. Force, acceleration and deflection were all considered as possible injury criteria, but deflection was found to provide the best correlation to trauma severity. A thorax relative deflection of 30 percent was found to equate to an injury level of AIS $\leq$ 3. This value compares reasonably well with Stalnaker's recommendation.

An important consideration in conducting side impact vehicle tests is the placement of the upper arm of the surrogate. The upper arm can be placed alongside the chest or raised, thereby exposing the thorax to direct impact. To resolve this question, it would be necessary to know the circumstances of arm placement in field accidents as well as the biomechanical effects of arm positioning on subsequent injury patterns to the thorax and arm.

Cesari, et al. (Ref. 63), studied the latter issue by conducting lateral impact tests to eight unembalmed cadavers. In most of these tests, the arm was alongside the thorax so that the thorax was impacted through the arm. These results were compared to those of a similar series, conducted previously, in which the arm was raised and the thorax was struck directly. Their impactor was a spherical sector with a 23.6 in (60 cm) spherical radius and a 6.9 in (175 mm) sector radius and weighed 51 lb (227 N). Impact velocities ranged from 6.2–16.8 mph (10–27.1 km/h). The arm was found to provide some protective value to the thorax when the arm received the blow. This protective effect was generally equivalent to a 10 percent change in impactor velocity. The nature of the thoracic injuries (as distinct from their severity) was not appreciably changed by the presence of the arm. Rib fractures were the most prevalent trauma but intrathoracic injuries also occurred. There was only one arm fracture, on a relatively severe test.

5.8 Abdomen—The abdomen is the least understood region of the body from the load tolerance viewpoint. It contains a variety of organs which can be exposed to impact forces. The organs most frequently injured as a result of blunt abdominal trauma are the liver, kidneys, spleen, intestines, pancreas and the urinary bladder. Only the liver and the spleen are partially protected by the lower aspects of the rib cage. 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.

5.8.1 TOLERANCE OF ABDOMINAL ORGANS (FRONTAL IMPACTS)—A large body of clinical literature has evolved over the years to document the various forms of injuries produced by blunt abdominal trauma. In contrast, there are only a few studies available on the loading condition, force levels, and impact velocities that characterize typical accident situations. One of the earliest of these is a 1953 report by Windquist, Stumm, and Hansen (Ref. 64). They employed upright seated, forward facing hogs to examine the effect of abdominal impacts against restraining belts (improperly worn lap belts) as well as objects that might be struck in an aircraft cockpit. These objects were a control wheel, a stick-like protrusion[6] and a large, flat surface similar to a radio box. The animals received impacts in both their midriff and lower abdominal regions at velocities of 20 and 40 ft/s (6.1 and 12.2m/s). All of the high velocity exposures (with belt loop forces ranging from 2360–6660 lb (10.5–29.6 kN)) were fatal. A force of 1080 lb (4.80 kN) against the 10 in (254 mm) square radio surface, a force of 893 lb (3.97 kN) against the projecting peg, and a 750 lb (3.34 kN) loop force through the abdominal belt were all considered survivable. Complete results are available in Table 11 which was given in a paper by Mertz and Kroell (Ref. 65).

A later study by Stalnaker, et al, probably provides the most extensive test results yet published on abdominal impact (Ref. 66). A series of 96 abdominal tests were carried out on four animal species-Rhesus monkey, Squirrel monkey, baboon, and pig. Various sized impactors were employed to simulate common automotive injuries. The abdomen was divided into three zones (upper, middle, and lower) which were analyzed separately. The voluminous data generated in this project was submitted to computer assisted statistical analysis to obtain correlations between the various impact parameters and the estimated injury severity ratings which were obtained on autopsy. Their overall results are summarized in Figure 3.

5.8.2 TOLERANCE OF ABDOMINAL ORGANS (LATERAL IMPACTS)—One of the earliest studies on lateral tolerance of the abdomen was reported by Stalnaker, et al. (Ref. 31), in 1973. These researchers impacted a variety of live, anesthetized primates in the right and left sides of their abdomens, employing a scaled armrest on a 22 lb (98 N) moving striker. The force levels required to produce injury varied significantly with the site of the impact. The upper portion of the abdomen was found to be more easily injured than the lower portion of the abdomen.

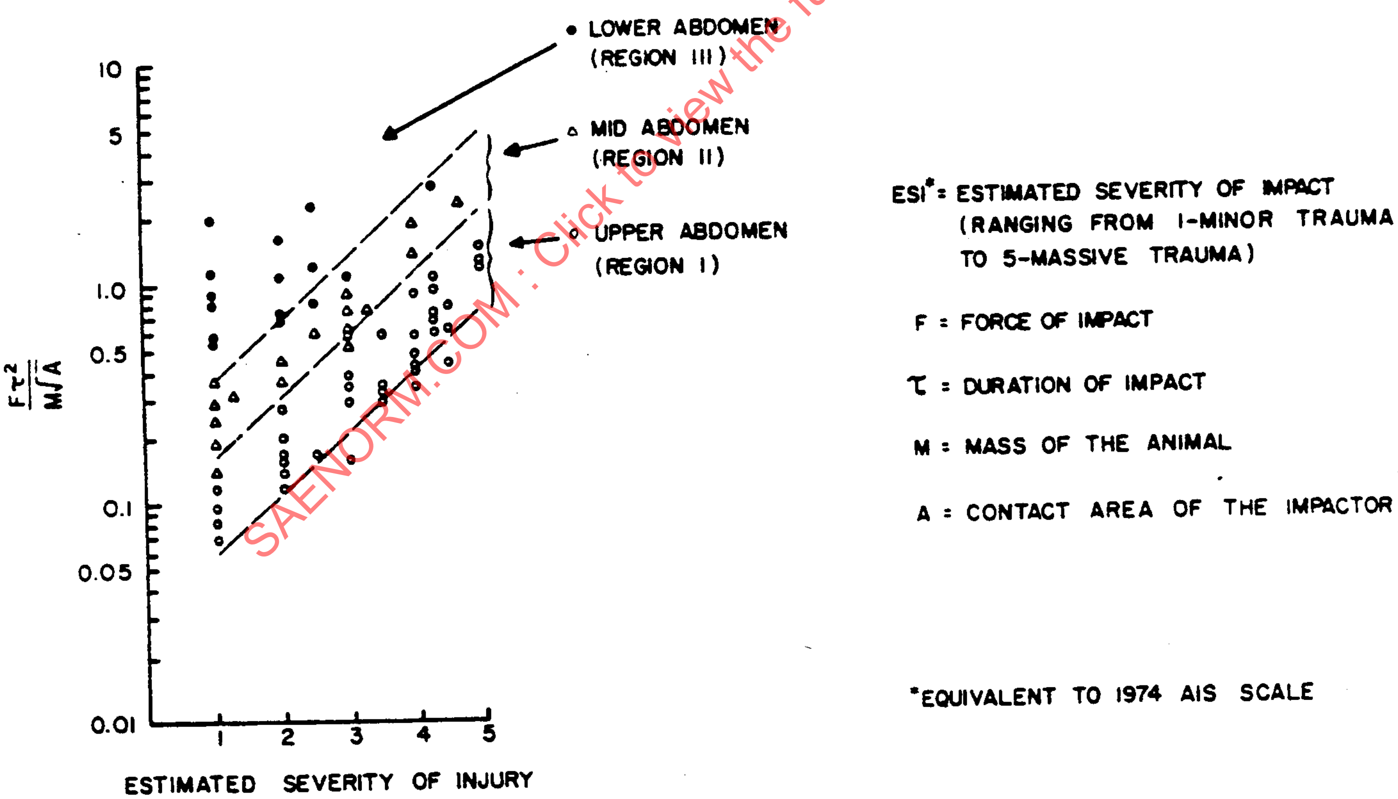


FIGURE 3—EXPERIMENTAL SCALING FACTOR FOR ABDOMINAL INJURY SENSITIVITY

6. struck end-on

TABLE 11—MIDRIFF LOADING (WINDQUIST, ET AL.)

Type of Exposure	Hog Weight lb	Hog Weight N	Impacct Velocity ft/s	Impact Velocity (m/s)	Maximum Force Applied lb	Maximum Force Applied kN	MFRT[1] PD[2] ms/ms	Approx. Body Compression % of Normal Thickness	Descrption of Injuries
Control Wheel Impingement Block	95	422	20.7	(6.3)				40	**Survivable**-Small subpleural and subendocardial hemorrhages; incomplete defect in anterior wall of stomach.
	100	445	40.7	(12.4)				80	**Fatal**-Massive internal injuries including bilat. fract disloc's at costochondral junct., ruptured diaphragmatic hernia, large inguinal hernias, perforation of stomach and intestines and laceration of spleen.
	166.5	740	39.1	(11.9)					**Fatal**-Multiple comp. rib fract. bilaterally; destruction of liver; lacerations of pericardium, heart, right lung, diaphragm, colon, and peritoneum.
Ten Inch Square Impingement Block	104	463	20.3	(6.2)	1080	(4.8)	48/95	30	**Survivable**-Small subpleural and subendocarial hemorrhages; subserosal hemorrhage of the colon.
Ten Inch Square Impingement Block	95	422	39.5	(12.0)	2360	(10.5)	22/85		**Fatal**-Massive internal injuries including multiple rib fract, and disarticulartion, lacerations of pericardial sac, heart, pulmonary artery, stomach, and intestines.
	156.5	696	40.3	(12.3)	6660	(29.6)	38/72		**Fatal**-Massive internal injuries including multiple rib fract., hemorrhage and emphysema of lungs, multiple lacerations of liver and spleen, multiple ruptures of colon.
Projecting Peg Impingement Block	175	779	17	(5.2)	893	(4.0)	68/135	70	**Survivable**-Single rib fracture.
	149	663	39.7	(12.1)	3085	(13.7)	27/95	90	**Fatal**-Massive internal injuries including puncture into right pleural cavity, comp. rib fract. bilaterally, laceration of pericardium, heart, right lung, diaphragm, liver, intestines, stomach, and peritoneum.
Control Wheel Impingement Block	185	823	23.6	(7.2)	2365	(10.5)	68/100	80	**Fatal**-Petechiae over lungs, subepicardial ecchymosis over anterior portion of interventricular septum, hemorrhage in diaphragm, subcapsular hemorrhage of liver, multiple ruptures of wall and colon.
	187	832	39.6	(12.1)	5080	(22.6)	46/88	80	**Fatal**-Massive internal injuries including laceration of rectus abdominus muscle, ruptured diaphragmatic hernia, lacerations of liver and spleen, pericapsular hemorrhage about left kidney, ruptures of colon.
Three Inch Wide Abdominal Belt	62.8	279	19.3	(5.9)	750	(3.3) (loop load)	28/52		**Survivable**-Subendocardial hemorrhage over septal portion of left ventricle and multiple subserosal hemorrhages of mid portion of jejenum.
	69.8	310	44.2	(13.5)	4700	(20.9) (loop load)	38/60		**Fatal**-Massive internal injuries including ruptured diaphragmatic hernia, ruptures of stomach and colon, fragmentation of spleen, lacerations of kidneys and liver.

1. Maximum force rise time.
2. Pulse duration.

Walfisch, et al, (Ref. 67), conducted lateral impacts to eleven unembalmed cadavers to determine lateral abdominal tolerance levels. Based on their previous accident study, they concluded that liver injuries were the most common of the serious abdominal traumas. Accordingly they impacted only the right side of their cadavers (which contains the bulk of the liver) in order to obtain results which they felt would be conservative. The cadavers were suspended horizontally, right side downward, and allowed to free fall from a height of 3.3 ft (1 meter) or 6.6 ft (2 meters). The abdomen struck a simulated armrest which was mounted to a load cell. The armrest was 2.8 in (7 cm) wide (corresponding to the s-i direction of the cadaver) and its depth and stiffness in the l-r direction were varied by employing various types of supporting material under a wooden form. Armrest depth ranged from 1.2 in (31 mm) – 2.2 in (55 mm). Abdominal penetration was obtained photographically, but an ambiguity developed in this measurement due to sagging of the abdomen and crushing of some of the armrests. In addition, the abnormal attitude of the cadaver relative to the seated position may affect the positioning of the abdominal organs and hence the damage results. Their force measurements and abdominal (liver) injury ratings are provided in Table 12.

TABLE 12—CADAVER DROP TESTS ON THE ABDOMEN (LATERAL)

Test No.	Height of fall (m)	Protrusion of simulated armrest (mm)	Supporting Material for armrest	F. Max da N	Abdominal AIS
205	1	31	rigid	160	0
206	1	51	rigid	535	4
209	1	51	polystyrene	380	4
210	1	51	polystyrene	415	3
211	1	53	phenespan	170	0
212	1	55	polystyrene	150	(1)
219	1	41	rigid	195	1
213	2	55	polystyrene	490	3
215	2	31	rigid	510	5
216	2	51	rigid	420	1
217	2	41	rigid	500	5

1. diseased livers - subjects eliminated from analysis

5.8.3 LOADING OF THE ABDOMEN BY A LAP BELT—Walfisch, et al. (Ref. 68), subjected fourteen unembalmed cadavers to a series of sled impacts in which the lap belt and cadaver were configured to promote over-riding of the pelvis by the lap belt. Belt loads and abdominal penetrations were measured during the tests and resultant injuries were determined post test. The damage found on these cadavers was considered to be similar to those of accident victims. The nature of the cadaver damage and their frequency of occurrence were:

Fracture of the lumbar spine	4
Tearing of the mesentery	2
Damage to the liver	2
Fracture of the pelvis	1
Perforation of the colon	1

This work led the authors to recommend that a conservative tolerance level for the abdomen (lap belt above pelvis) would be a lap belt load of 450 lb (2.0 kN) per side accompanied by an a-p lap belt intrusion of 1.4 in (35 mm) per side. In a later paper, Leung (Ref. 69), a member of the earlier research team, increased the recommended abdominal tolerance level to an average lap belt tension of 790 lb (3.5 kN), (average of inboard and outboard lap belt loads) accompanied by an average lap belt penetration of 1.5 in (39 mm) (average of the a-p belt penetrations on the right and left side of the abdomen). It should be noted that their load tolerance recommendation refers to the lap belt load and not to the force on the abdomen. This recommendation should be applied cautiously to routine vehicle testing since lap belt geometries in some vehicles and abdominal stiffnesses of most test devices can vary appreciably from the conditions obtained in the referenced study. In addition, their recommendation may not be applicable to the situation in which lap belt over-ride occurs on only one side of the pelvis since that condition was not investigated.

Research by Nusholtz, et al, (Ref. 70), indicated that extreme care is needed when determining abdominal tolerance levels through the use of post-mortem specimens. This group compared the injuries to live and post-mortem primate subjects produced under blunt impacts delivered laterally to the thoraco-abdominal region. The live primates were injured more seriously than their corresponding post-mortem subjects. They suggested that the disparity may have been due to the lack of pressurization in the post-mortem specimens in this series.

5.8.4 LOADING BEHAVIOR OF ABDOMINAL ORGANS—Melvin, et al., studied the loading behavior of livers and kidneys which were surgically mobilized[7] from anesthetized Rhesus monkeys and then placed on an uniaxial load cell while still being supplied with blood by the living animal (Ref. 71).

Tests were performed at ram speeds of 120, 6000, and 12,000 in/min (0.05, 2.5, and 5.1 m/s) and average stress/strain[8] curves were obtained. In addition, the resulting injury severity was estimated immediately after impact using an ESI[9] injury scale of 1 (minor) to 5 (massive).

The authors concluded that both organs were sensitive to loading rate effects with the liver being more affected. A liver trauma rated as an ESI of 3+ was produced at a dynamic average stress level of approximately 45 psi (310 kPa). The kidney, with its thick, tough capsule, displayed wide variations in injuries at a given dynamic stress level. Its injury severity appeared to be more properly related to its strain level. The organ injuries in this study were noted to be similar to those seen clinically.

5.9 Lower Extremities—The structural elements of the lower extremities consist of the pelvis, the femurs, the tibias, the smaller fibulas, and the ankle and foot bones. In addition, there are the patellas (bony kneecaps) which cover the knee joints in front and serve as termini for ligaments and tendons. Also noteworthy is the pronounced offset of the head of the femur where it articulates in a ball and socket type joint at the pelvis.

5.9.1 TEST TECHNIQUES—Two types of test procedures have been employed to study the strengths of the patella/femur/pelvis bone complex. One technique employs an instrumented moving impactor to load the upper leg of a stationary, seated cadaver. The other procedure employs a moving test sled to propel the entire cadaver against instrumented surfaces arranged to simulate a vehicle interior.

5.9.1.1 Impactor Test Data—Powell, et al. (Ref. 72) tested the legs of nine cadavers and obtained fractures at loads ranging from 1600–2970 lb (avg. 2360 lb) (7.12–13.20 kN) (avg. 10.50 kN). Eighty percent of their legs suffered patellar fracture, 33 percent sustained condylar fractures (the portion of the femur adjacent to the patella), and only 6.7 percent were fractures to the shaft of the femur. They attributed the fracture patterns to the rigid impactor which they used.

7. Surgically separated from its surrounding tissues but leaving connecting blood vessels intact.

8. $$\text{average stress} = \frac{\text{impactor force}}{\text{iimpactor area}}$$

$$\text{average strain} = \frac{\text{deflection}}{\text{initial thickness of specimen}}$$

9. This scale is equivalent to the 1974 AIS scale.

Melvin, et al. (Ref. 73), employed an impactor with 1 in (25 mm) of Ensolite padding to test the femurs of fourteen stationary, seated cadavers. No fractures were obtained below 3000 lb (13.30 kN) and it was noted that a threshold impactor momentum of 40–50 lb·s (180–220 N-s) appeared to be necessary to cause fracture. Their relatively high fracture load levels (as compared to previous studies) were attributed to their exclusive use of unembalmed cadavers. All of these fractures were in the patella or in the distal third and supracondylar region of the femur.

Viano et al. (Ref. 74), conducted a series of axial knee impact tests on a total of six seated cadavers. A 22.3 lb (10.1 kg) impactor was used in conjunction with varying degrees of padding. The skin was removed from the impacted areas, but the structural integrity of the knee joint, including the ligaments, was left intact. This procedure was used in an effort to measure the time of fracture initiation, based on analysis of high-speed movies of the patella/femur during impact.

All six of the tests with rigid impactors produced both patella and femoral shaft fractures, and also produced either a condylar or neck fracture. The peak force ranged from 3010 to 6410 lb (13.4 to 28.5 kN), with an average of 4110 lb (18.3 kN). Many of the specimens, especially those impacted without padding, had multiple fractures. The authors contend that most of these fractures were initiated after the load peaks. The average load that the authors associated with shaft fractures was 2300 lb (10.2 kN) compared to the average peak load of 3350 lb (15.0 kN). The two tests conducted with lightly padded impactors both produced bilateral condylar fractures. The peak loads were 3600 and 3460 lb (16.0 and 15.4 kN). Only two of the five tests with thickly-padded impactors produced fractures (one condylar and one femoral shaft) and both of these involved cadavers which were considered by the authors to have bones in "abnormal" condition. The three "normal" specimens produced peak loads of 1190, 3100 and 3150 lb (5.3, 13.8 and 14.0 kN), and presented no fractures.

A shortcoming of some of these studies with moving impactors has been the poor correspondence in the location of the femur fractures as compared to those found in the field. Melvin, et al. (Ref. 73), described the distribution of lower limb fractures that occur in real world collisions. They found 142 unbelted vehicle occupants who suffered lower limb fractures in frontal collisions. Of these, thirty-nine (27 percent) had patellar or distal femur (adjacent to kneecap) fractures. The impactor studies have produced proportionately more such fractures.

This disparity is felt to be due to the non-representative ridigity and orientation of the laboratory impactors. The rigid or near-rigid impactors produced pulse durations of only 3 to 10 ms, as compared to the 30 to 50 ms durations which characterize actual instrument panel impacts. In addition, the impactors were aligned with the femur, maximizing its compressive load carrying capacity. In accidents, it is more likely that the knee impact force vector and the femoral axis will not be aligned, thereby resulting in greater bending stresses and uneven force distribution between the patella and condyles. These factors reduce the load carrying capacity of the legs. For these reasons, impact tests of 30–50 ms duration, with non-aligned femurs, are probably more relevant to the automotive collision environment than are the results from the various impactor studies.

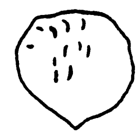
TABLE 13A—DYNAMIC FRACTURE FORCES FOR FEMUR, PATELLA, AND PELVIS (REF. 74A)
(RIGHT THIGH/LEFT THIGH)

Right Thigh Cadaver No.	Right Thigh Max Force Applied lb	Right Thigh Max Force Applied kN	Result	Left Thigh Max Force Applied lb	Left Thigh Max Force Applied kN	Result
1	950	4.23	Supracondylar fracture (bone defect suspected)	1400	6.23	intertrochanteric fracture (fractured through bone screw)
2	1500	6.68	Mid-shaft fracture	1600	7.12	No fractures
3	1500	6.68	No fractures	1600	7.12	No fractures
4	1650	7.34	Supracondylar fracture	1650	7.34	Supracondylar fracture
5	2150	9.57	No fractures	2100	9.35	No fractures
6	2250	10.00	No fractures	2250	10.00	Superacondylar fracture
7	1900	8.46	No fractures	1850	8.23	Superacondylar fracture
8	2800	12.46	No fractures	1750	7.79	No fractures
9	3850	17.13	No fractures	2650	11.79	Dislocated intertochanteric fracture
10	2400	10.68	Comminuted fracture of distal third of shaft and intercondylar notch	1800	8.01	No fractures

FEMUR

TABLE 13B—DYNAMIC FRACTURE FORCES FOR FEMUR, PATELLA, AND PELVIS (REF. 74A) (RIGHT KNEE/LEFT KNEE)

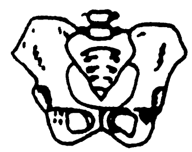
Cadaver No.	Max Force Applied lb	Max Force Applied kN'	Result	Max Force Applied lb	Max force Applied kN	Result
1	950	4.23	No fractures	1400	6.23	No fractures
2	1500	6.68	No fractures	1600	7.12	No fractures
3	1200	5.34	No fractures	1600	7.12	Abnormality-but not adequately identified as fracture
4	1650	7.34	No fractures	1650	7.34	No fractures
5	2050	9.12	No fractures	1550	6.90	No fractures (padded)
	1550 1800 1950 2150	6.90 8.01 8.68 9.57	Heavy abrasion and fracture of patella (unpadded)	1500 1800 2000 2100	6.68 8.01 8.90 9.35	Complete fracture of patella, damage to articular cartliage (unpadded)
6	1700 2050 2250	7.57 9.12 10.00	No fractures Comminuted fracture of patella	1950 2000	8.60 8.90	No fractures ominuted fracture) of patella
7	1900	8.46	No fractures	1850	8.23	No fractures
8	2550	11.35	Comminuted fracture of patella	1750	7.79	No fractures
9	3850	17.13	Linear fracture of patella	2650	11.79	No fractures
10	2400	10.68	No fractures	1800	8.01	No fractures



PATELLA

TABLE 13C—DYNAMIC FRACTURE FORCES FOR FEMUR, PATELLA, AND PELVIS (REF. 74A) (RIGHT HIP/LEFT HIP)

Right Hip Cadaver No.	Right Hip Max Force Applied lb	Right Hip Max Force Applied kN	Result	Left Hip Max Force Applied lb	Left Hip Max Force Applied kN	Result
1	950	4.23	No fractures	1400	6.23	No fractures
2	1500	6.68	No fractures	1600	7.12	No fractures
3	1200	5.34	No fractures	1600	7.12	Fractures of superior and inferior rami of pubis
4	1650	7.34	No fractures	1650	7.34	No fractures
5	2150	9.57	No fractures	2100	9.35	No fractures
6	2250	10.00	No fractures	2250	10.00	No fractures
7	1900	8.46	Severe multiple fractures	1850	8.23	No fractures
8	1400	6.23	Possible mid fracture of ischium	1750	7.79	No fractures
	2250	6.90	Severe multiple fractures			
9	2750	12.24	No fractures	1950	8.68	Possible mild fracture of transverse ramus
	3850	17.13	Severe multiple fractures	26.50	11.79	Severe multiple fractures
10	2400	10.68	No fractures	1800	8.01	No fractures



PELVIS

5.9.1.2 Sled Test Data

5.9.1.2.1 Loading Through the Knee Joint—The earliest lower limb studies of automotive interest were conducted by Patrick, Kroell, and Mertz (Ref. 74a). Their objective was to determine the strength of the patella/femur/pelvis complex in impacts simulating knees striking instrument panels. Ten embalmed cadavers were tested in full-scale impact sled experiments. The seated cadavers translated forward during sled deceleration to impact against four padded load cells. The head, chest, and each knee struck a separate load cell. These cells were geometrically arranged to simulate the forward surfaces of an automobile passenger compartment. These researchers concluded that the femur was slightly more vulnerable to fracture than the patella or the pelvis, but that distinction was too small to allow confident prediction as to which bone structure would fail first. Their complete results are available in Table 13. A later study by the same investigators obtained loads of 1470, 1710, 1950, and 1970 lb (6.54, 7.61, 8.68 and 8.76 kN) on two cadavers without fractures (Ref. 75).

Viano and Culver (Ref. 76) conducted sled tests with cadavers restrained by a shoulder-belt-plus-knee-bolster configuration (i.e., no lap belt). For six of these subjects the bolster was positioned so that the knees impacted it squarely. No injuries were produced for bolster loads which ranged from 1190 to 1800 lb (5.3 to 8.0 kN) per leg with an average of 1420 lb (6.3 kN).

5.9.1.2.2 Loading Below and Across the Knee Joint—In the preceding studies, the loading of the femur was primarily through the patella and/or femoral condyles and resulted in patella, femur and/or pelvis fractures. If the loading is applied below or across the knee joint, damage to the knee ligaments and/or fractures of the tibia and fibula may result. Viano, et al. (Ref. 77), impacted seated cadavers on the anterior portion of the tibia, just below the knee joint. They found that impactor forces ranging from 740 to 1550 lb (3.28 to 6.89 kN) with an average of 1140 lb (5.09 kN) produced knee ligament tearing and/or tibia-fibula fractures. In two tests, no damage was observed for peak loads of 1090 lb (4.87 kN) and 1290 lb (5.74 kN). For eight impacts which spanned the knee joint (involving both the patella and tibia), knee joint damage was produced for impactor forces ranging from 1330 to 1880 lb (5.91 to 8.36 kN) with an average of 1580 lb (7.02 kN). The predominant injury mode was avulsion of the posterior cruciate ligament from the tibial plateau.

Sled tests were conducted by Viano and Culver (Ref. 76) in which shoulder belted cadavers (without lap belts) impacted knee bolsters. The two below-the-knee leg impacts produced significant ligament tears at peak bolster loads of 790 lb (3.5 kN) and 940 lb (4.2 kN) per leg.

5.9.1.3 Static Tests of Knee Joints—Viano, et al. (Ref. 77), also performed low-speed ligament tolerance tests on five isolated knee joints mounted in a universal testing machine. In these tests, the knee joint angle was maintained at ninety degrees while the tibia was displaced rearward relative to the femur until complete joint failure occurred. Loads corresponding to the initiation of joint failure ranged from 320 lb (1.43 kN) to 575 lb (2.56 kN) with an average of 455 lb (2.02 kN). The corresponding displacement of the tibia relative to the femur at the initiation of joint failure ranged from 0.37 in (9.5 mm) to 1.18 in (30.0 mm) with an average of 0.57 in (14.4 mm). The load corresponding to complete joint failure ranged from 375 lb (1.67 kN) to 675 lb (3.0 kN) with an average of 560 lb (2.48 kN).

5.9.2 THEORETICAL ANALYSES AND PROPOSED INJURY CRITERIA—Viano and Khalil (Ref. 78) have analyzed the stress distribution in the axially-loaded femur. They concluded that the location and magnitude of peak femur stresses can be significantly affected by small shifts in the location of the applied load, such as moving its point of application from one condyle to the other.

Time dependent, compressive force femur injury criteria have been proposed by King, et al. (Ref. 79), and Viano (Ref. 80).

The criterion proposed by King, et al., is:

$$F = A - B \log_{10} T \quad \text{(Eq. 3)}$$

where

F = permissible peak compressive femur force
A = 1370 lb (6.09 kN)
B = 215 lb (960 N)
T = pulse duration in seconds

Viano's proposed criterion is:

for T less than 20 ms

$$F = A - BT \quad \text{(Eq. 4)}$$

and for T greater than 20 ms

$$F = C \quad \text{(Eq. 5)}$$

where

F = permissible peak compressive femur force
A = 5200 lb (23.1 kN)
B = 160 lb (710 N)
C = 2000 lb (8.9 kN)
T = pulse duration in milliseconds

The femur limit currently specified in MVSS 208 is a compressive load of 2250 lb (10 kN) for each femur (Ref. 81). Previous MVSS 208 specifications were 1400 lb (6.23 kN) and 1700 lb (7.55 kN).

5.9.3 CONCENTRATED LOADING OF THE PATELLA—It was mentioned previously (see 4.3.1) that the unique construction of the patella makes it vulnerable to concentrated loadings. This phenomenon was studied by Melvin, et al. (Ref. 82), employing three different impactor sizes, all unpadded. Two of the impactors were flat surfaced, circular areas with diameters of 0.61 in (15.5 mm) and 0.43 in (10.9 mm), while the third impactor was ring shaped with an outer diameter of 0.50 in (12.7 mm) and an inner diameter of 0.25 in (6.4 mm). Ninety embalmed patellas were tested with the results shown in Table 14.

TABLE 14—PATELLA LOCALIZED LOADING (RESULTS AVERAGED OVER ALL TEST SPEEDS)

Impactor Size	Area in^2	Area cm^2	Average Failure Load lb	Average Failure Load kN	Minimum Failure Load lb	Minimum failure Load kN
0.43 in (10.9 mm) dia Circular Impactor	0.15	(0.97)	1030	(4.58)	560	(2.49)
0.61 in (15.5 mm) dia Circular Impactor	0.29	(1.87)	1260	(5.60)	700	(3.11)
0.50 in (12.7 mm) dia Ring Impactor	0.15	(0.97)	1320	(5.87)	650	(2.89)

These tests were carried out at three different test conditions to determine the effect of velocity: -static, 10 mph (4 1/2 m/s) and 20 mph (9 m/s). The patella damage pattern varied dramatically with speed. The impactors caused a clean punch-through of the patella during the static tests but multiple fractures or near total destruction of the patella during the 10 and 20 mph impacts (4 1/2 and 9 m/s). The change in fracture