DIV.006.BOX

Chapter 2

ETIOLOGY OF TRAUMA TO THE CERVICAL SPINE

JAMES McELHANEY, VERNE ROBERTS, JACQUELINE PAVER, AND MARET MAXWELL

INTRODUCTION

FROM a mechanical and structural point of view the neck is a very complex mechanism. While anatomists have studied the neck for many years, it is only recently that serious attempts have been made to quantitate its geometric and structural properties. The advent of high speed land and air transportation has made us increasingly aware of the serious consequences that can result from a structural failure of the neck. Also, as more people pursue leisure time activities, the potential for serious neck injuries increases. Football, diving, gymnastics, skiing, hang gliding, mountain climbing, and amusement rides are but a few activities that expose the neck to a risk of serious injury.

The human neck contains vital neurologic, vascular, and respiratory structures as well as the fragile cervical vertebrae and spinal cord. While injury statistics generally attribute only 2 to 4 percent of serious trauma to the neck, *any* neck injury can have debilitating if not life-threatening consequences.

Therefore, a variety of devices have evolved that offer a measured protection to the neck from mechanical trauma. Head restraints, motorcycle and football helmets, energy-absorbing pads and collars, and gymnastic mats are but a few examples of head and neck protective devices. Unfortunately, the design of many of these has proceeded with little biomechanical input.

This chapter summarizes research aimed at providing some biomechanical responses of the neck in a form that hopefully will be useful in design.

To that end, load-deformation characteristics of the neck and several helmet types are presented. Neck injuries are described and classified. Accidents that involve neck injuries are analyzed. Real-life neck injuries are investigated, and the mechanical aspects are simulated in the laboratory and mathematically.

It should be recognized at the outset that much of the work described here is ongoing, and the results are therefore preliminary and subject to modification as more data is accumulated.

LITERATURE REVIEW

Trauma to the cervical spine is, because of the extreme disability it can produce with the limited improvement now available for restoration of lost function, one of the most important areas of concern in the fields of biomechanics and injury prevention. The literature regarding the origins and treatment of cervical trauma is rich with

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references, yet our study has shown that there is still a limited amount of information available regarding the forces required to produce the injuries described and the precise mechanisms by which the fractures and dislocations of the spine occur. This chapter is limited to those injuries resulting from some compression with either flexion or extension of the spine and does not attempt to describe the extreme hyperextension injuries and their associated literature.

Because the study of cervical spine trauma has evolved from a wide variety of institutions and sources, several authors have attempted to classify the injuries so that there can be agreement as to what each investigator is referring to as he describes a particular injury. Babcock, in 1976 in his article in the *Archives of Surgery*, developed a classification which is complete and should be more widely used.[4] Other attempts to classify the injuries have been developed by Melvin et al.,[70] Moffatt et al.,[76] and Portnoy et al.[82]

While injuries to the cervical spine can result from almost any activity, the literature suggests that automobile and aircraft accidents, football, and diving are the circumstances most represented in the literature. Automobile and aircraft accidents undoubtedly produce extensive injuries because of the speeds involved and the associated energy that must be dissipated in a crash. According to Huelke et al.,[46] who cite an earlier study, 56 percent of all spinal cord injuries are the result of a highway accident, with 67 percent of those involved in highway accidents being vehicle occupants. Pedestrians and motorcyclists were also significantly involved in the injury statistics. Other studies of automobile and highway-related cervical spine injuries include the papers of Alker et al.;[3] Bowman and Robbins;[12] Kiesel et al.;[54] Kihlberg;[55] Langwieder;[57,58] Melvin et al.;[69] Mertz;[71] Schutt and Donan;[95] Sims et al.;[98] Voight and Wilfert;[108] Thorson;[104] Tonga et al.;[105] and Yule.[109]

Sports activities, especially football, also produce injuries to the cervical spine. The literature is divided into three primary areas of concern. The incidence of these injuries in the game of football is described by Albright et al.[2] Schneider, in his book on football injuries as well as in his numerous papers, related trauma to the impingement of the rear of the helmet shell on the neck structures.[91] Subsequent authors have described the role of the helmet in producing neck injuries and have seriously debated the mechanism proposed by Schneider. These authors, such as Hodgson and Thomas,[43] Mertz et al.,[72] and Virgin,[107] have attempted to verify Schneider's experiments with varying degrees of success. Gurdjian et al. in 1962 published the Wayne State tolerance curve and recommended its use for the design of helmets to reduce brain injury.[38]

Swimming and diving have also been identified as causing significant numbers of fractures and dislocations of the cervical spine. Kewalramani and Taylor[52] found that 18 percent of all spinal cord injuries in their series were related to diving accidents. Albrand and Walter in 1975 published curves that related depth in feet and the velocity of the head to the height from which a diver dove into the water.[1] McElhaney et al., in their paper published by the SAE in 1979, also provided experimental data relative to body velocity and the depth of the water.[64] Their series of accidents included not only springboard diving but water slides as well. They suggest that a head velocity of 10.2 feet per second with a following body is sufficient to cause compression fractures of the cervical spine, most frequently at the level of C5.

Automobile restraint systems have also been associated with spinal cord injuries. These injuries are frequently described as shearing injuries produced in high speed crashes with the seat belt providing a fulcrum for forced spinal rotation. Authors who discuss the relationship between restraint systems and spinal injury include Burke;[15] Epstein et al.;[21] Gogler and Athanasiadis;[36] Horsch et al.;[45] Marsh et al.;[62] Nyquist et al.;[78] Schmidt et al.;[89] and Taylor et al.[103]

Injuries to the spinal column are frequently diagnosed and described on the basis of radiological examination of the cervical spine. The articles describing the radiological findings are important in developing the mechanism of the injuries, since they frequently illustrate actual fractures as opposed to artists' renditions and on occasion yield data that can be translated into a biomechanical reconstruction of the injuries. Typical articles that we have found to be useful include those of Delahaye et al.;[19] Gehwerter et al.;[34] Harkonen et al.;[41] Scher;[88] and Taylor and Blackwood.[101]

Another source of information regarding the etiology of spinal injuries may be found in the clinical literature associated with the management of these unfortunate individuals. Particularly useful are those articles which describe in detail associated injuries beyond the spinal lesion itself. The associated injuries frequently yield valuable input regarding the location of the force and sometimes its direction. A limited selection of these clinical papers include those of Bailey;[5] Beatson;[9] Forsyth et al.;[26] Guttman;[39] Kessler;[51] La Rocca;[59] Makoyo;[61] Miller and Schultz;[75] Panjabi et al.;[81] Rogers;[87] Selecki;[96] Seljeskog and Chou;[97] and States.[99]

With the number of injuries reported and the seriousness of the resulting disability, numerous investigators have tried to experimentally create the injuries in the laboratory, develop mathematical models of the spine, and develop more reasonable mechanical simulations of the neck in order to better evaluate the design of restraint systems, helmets, and automotive interior configurations.

Experimental techniques frequently involve the use of cadaveric material either as isolated vertebrae, sections of the spinal column, or intact human cadavers. Roaf, in his classical study in 1960, utilized isolated disc-vertebrae combinations to document the mechanisms of cervical spine fractures.[84] Of particular significance was his documentation of the splitting action produced by the disc material being driven into the vertebral bodies. Bauze and Ardran, in their papers in 1975 and 1978, report on their ability to experimentally produce dislocations of the cervical spine and illustrate, using cineradiological techniques, the dislocations taking place.[7,8] Culver et al.,[17] in their study, utilized whole body impacts to experimentally produce fractures of the cervical spine. Other investigations have included Gosch et al.;[37] Hodgson et al.;[42] and Lange.[56]

The largest and most complete series using volunteers has been performed by Ewing and his coworkers. Their results, as typified by Ewing et al.,[22-25] have combined with the work of McElhaney et al.[63] and Mertz and Patrick[74] to describe the kinematics of the head-neck system.

A wide variety of mathematical models have been developed and reported in an attempt to model the experimental data. These include the work of Becker;[10] Bowman and Robbins;[12] Fox and Williams;[28] Huston and Advani;[49] Orne and Liu;[79] Robbins et al.;[86] Schultz and Galante;[94] Tarriere and Sapin;[100] Toth;[106] and Panjabi.[80]

Additionally, there have been numerous attempts to develop mechanical devices that produce motions consistent with experimental data. These devices are described by Culver et al.;[18] Haffner and Cohen;[40] Melvin et al.;[68] and Mertz et al.[73]

The result of the studies reported in the literature has been to develop our growing understanding of the etiology of the fractures, dislocations, and soft tissue injuries of the cervical spine. Portnoy et al., in their paper in 1971, were among the first to describe mechanisms by which impact locations and the direction of the blow contributes to the production of fractures and dislocations.[82] More recently, Bauze and Ardran,[8] by photographing an actual dislocation, have provided significant input regarding the injury type. Certainly, the early work of Schneider and his coworkers[92,93] yielded valuable information regarding the "hangman's" and "teardrop" fracture modalities. Blockey and Purser,[11] as well as Portnoy et al.,[82] described the process of injury to the odontoid, while Friede,[29] Garber,[33] and Jefferson,[50] discuss the atlas. Other investigations have included the works of Key[53] and Norton.[77]

MECHANICAL PROPERTIES

In the context of neck injury, the properties of greatest interest are the so-called structural properties. In addition, certain anatomical, physiological, and material properties play an important role in understanding the response of the human neck to potentially traumatic environments.

In the engineering disciplines, material properties such as ultimate strength and stress-strain relations are of great importance. A designer starts with a basic building material and shapes it into a structure with specified load and deformation responses. Since the human body exists, its load and deformation responses cannot be changed, and knowledge of the properties of the materials of which the body is composed is only useful insofar as it leads to a better understanding of structural responses. By definition, structural responses are those load and deformation characteristics which are relatable to the size, shape, configuration, and material of which a structure is composed. Material properties in contrast are generally represented as being independent of the structure or shape of the material under consideration.

The most important structural properties in this context are —

1. load to failure
2. stiffness
3. energy to failure
4. damping or energy absorbed

These properties are most frequently displayed in a load-deformation curve where the stiffness is the slope and the energy is the area under the curve.

We have been performing compression tests on neck components in order to better understand neck trauma and neck protection. The preparations have been entire cervical spines with the atlanto-occipital junction intact, a disc with a vertebra on each side, and vertebral bodies. The cervical spine specimens include the ligamentous structures but have the muscle tissue removed. A section of the base of the skull is included. This is molded into a flat section of dental acrylic parallel to the C7-T1 disc. Loads are applied through parallel steel plates at these sections. Seven cervical spines have been tested so far. The tests are performed on a Minneapolis Testing Systems Company closed-loop hydraulic testing machine. Load and deflection data are recorded in chart form and also tape recorded for later computer analysis. The neck tissues are obtained at autopsy and tested immediately or frozen and tested later. Tests are performed with the tissues at body temperature and 100% relative humidity or at room temperature and humidity. Long-term creep and relaxation tests show some temperature and humidity effects, but the results of the dynamic tests do not appear to be affected by small variations in temperature, humidity, and time after death if kept frozen.

Figure 2-1 shows the load-deflection for a human cervical spine. This specimen was obtained from a 44-year-old male and tested shortly after removal, i.e. 36 hours after death. A programmed 0.5 inch deformation with a constant velocity was applied in 0.1 second. Radiographic analysis and dissection showed no significant tissue damage. This curve shows nonlinear stiffness with a large hysteresis or damping factor. The energy applied was approximately 21 foot-pounds, and the hysteresis loop was approximately 15 foot-pounds. The loading stiffness varied from approximately 1200 to 3600 pounds per inch.

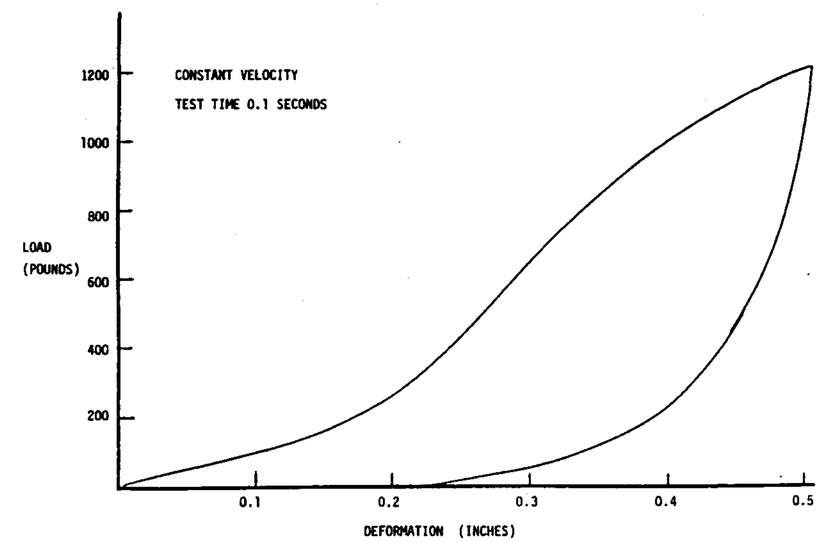

Figure 2-1. Typical compression test of the human cervical spine.

Low amplitude compression tests were performed to investigate the rate sensitivity of the intact cervical spine. A prestrain of 0.015 inch was applied, and a sinusoidal deformation of 0.060 inch peak-to-peak was applied. The load and time histories were measured at frequencies of 0.1, 1.0, and 10 Hz. Figure 2-2 shows the envelope for a typical test. No significant rate effects were observed, probably due to the low amplitude used. As the number of deformation cycles increased, a significant reduction in stiffness occurred. This was probably due to fluid transport through the annulus. Figure 2-3 shows typical test results from a standard load-relaxation test performed on a complete cervical spine specimen. The relaxation behavior is quite similar to that of the lumbar and thoracic spine. Figure 2-4 shows typical relaxation test data.

Results of load-relaxation tests indicate that the mechanical response of the intervertebral disc cannot be treated simply as a three-parameter solid, either linear or nonlinear. The presence of rapid initial load decay for fixed deformations renders a model with a single dominant long-term time constant a poor predictor of disc behavior. In order to incorporate short-term behavior into a behavior predictor, an

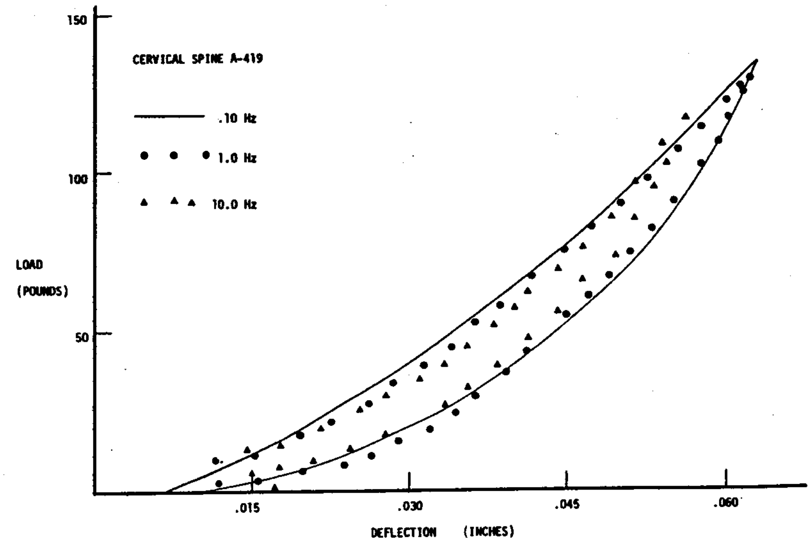

Figure 2-2. Variable rate compression of the human neck.

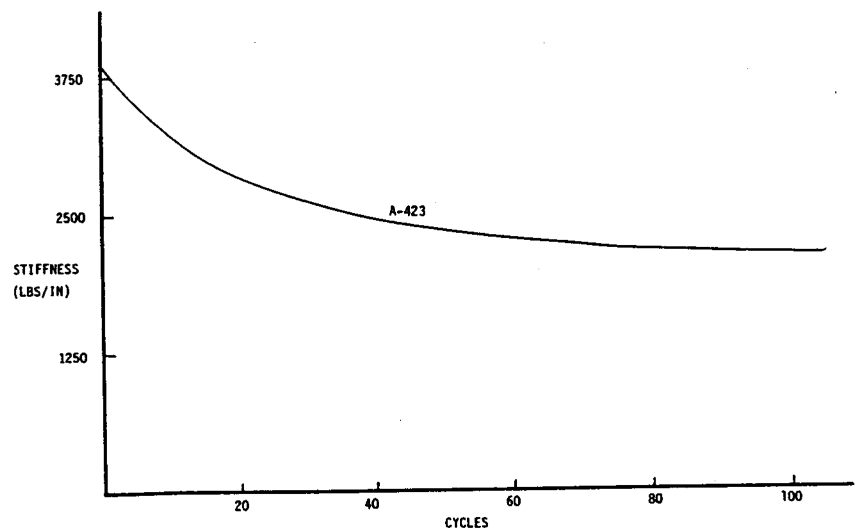

Figure 2-3. Stiffness tests, human necks.

ensemble of decay mechanisms and associated decay time constants must be considered. In a mechanical sense, a continuous spectrum of relaxation mechanisms may be viewed as arising from a generalized Maxwell-Weichert model. Such a model incorporates a continuous parallel linkage of Maxwell elements with degenerate elements introduced, when necessary, to mimic material behavior. To this end, a reduced relaxation function, $Y_r(t)$, may be written in the manner suggested by Fung:[30]

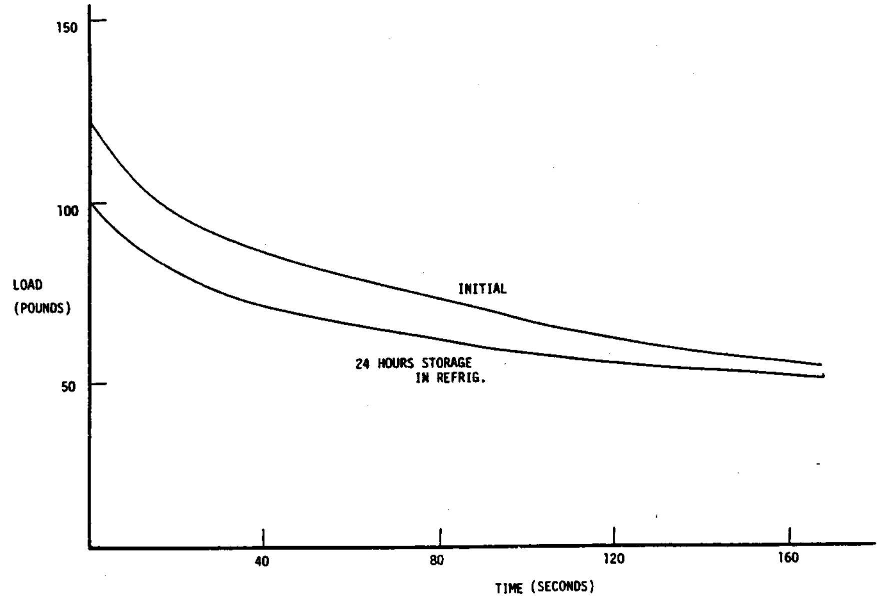

Figure 2-4. Relaxation tests, human neck.

$$Y_r(t) = \left[K_\infty + \int_0^\infty k(\tau) e^{-t/\tau} d\tau \right] \Bigg/ \left[K_\infty + \int_0^\infty k(\tau) d\tau \right]$$

Defining $H(\tau)$, the relaxation time distribution function as:

$$H(\tau) = \tau k(\tau)$$

and substituting into this equation we find, after some rearrangement:

$$Y_r(t) = \left[1 + \frac{1}{K_\infty} + \int_0^\infty H(\tau) e^{-t/\tau} d(\ln \tau) \right] \Bigg/ \left[1 + \frac{1}{K_\infty} + \int_0^\infty H(\tau) d(\ln \tau) \right]$$

$H(\tau)$ may be approximated from experimental data as the negative slope of the logarithmic reduced load-relaxation curve.

The relaxation spectrum approximation may be incorporated in a hereditary integral representation in order to predict load-deformation behavior at various deformation rates. Employing $Y_r(t)$ in the quasi-linear viscoelastic representation, we find:

$$F(\delta, t) = \int_0^t Y_r(t - \tau) \frac{dF^e}{d\delta} [\delta(t)] \frac{d\delta(\tau)}{d\tau} d\tau$$

where $F(\delta,t)$ is the force as a function of deformation and time, $dF^{e}/d\delta$ is the slope of the "elastic" load-deformation curve, and $d\delta(\tau)/d\tau$ is the change in deformation with time.

In the integral representation, $Y(t-\tau)$ is obtained from relaxation tests on the disc under consideration and $d\delta(\tau)/d\tau$ is obtained from the deformation rate. For short stroke times (less than 1 second for full displacement), the load-deflection curves for any particular disc undergo rapid compaction. Instantaneous deformation of the disc is impossible to achieve in a physical sense, but load-deflection behavior appears to be relatively insensitive to deformation rate for time scales of less than 0.1 second. The elastic load-deflection curve, therefore, may be reasonably taken as the 0.1 second full stroke load-deflection curve.

For integral calculations, the pseudoelastic load-deflection curves were computer fitted to a power series in δ. Little additional information was gained by extending the power series beyond a 4th order expansion. For a more complete discussion of this model and comparisons with dynamic test data, refer to the study by Casper and McElhaney.[16]

Additional testing has been performed on vertebral bodies and intervertebral discs. Figure 2-5 shows typical compression responses at ram speeds of 10 inches/second and test times of approximately 10 milliseconds. It is interesting to note that the disc and body are approximately the same strength, but the body, because it has a softening load-deformation response, requires much more energy to cause failure.

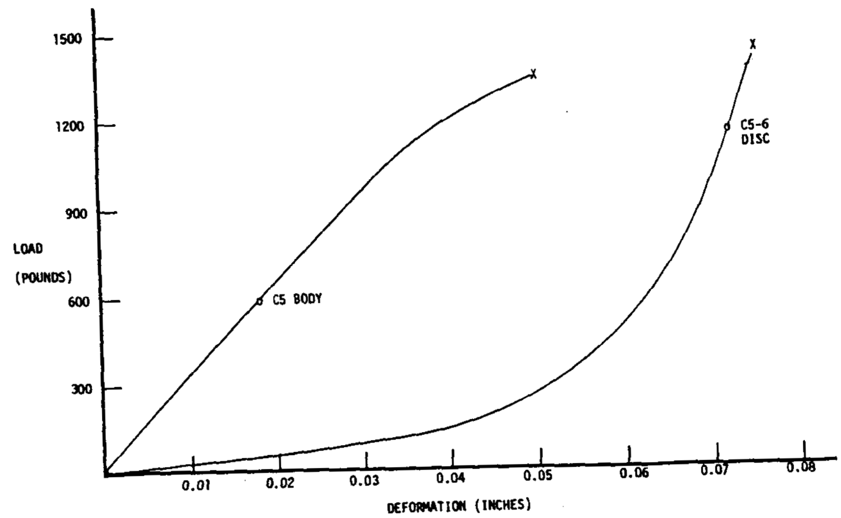

Figure 2-5. Load-deformation, human neck elements.

Cancellous bone specimens from the human vertebral body have been tested in compression. The specimens were unembalmed, and the tests were generally performed not less than five days postmortem. During this period, the specimens were kept refrigerated and wet with a calcium buffered solution of isotonic saline. To date, over 480 specimens from 81 donors have been tested. Ultimate strength data and age are presented in Figure 2-6. Regression analyses indicate a strong dependence of ultimate compressive strength and modulus of elasticity on age and density. The average values of the properties measured for a load direction parallel to the long axis of the spine were —

	Mean	*Standard Deviation*
Dry Density lb./in.^{3}	0.062	0.025
Modulus of Elasticity lb./in.^{2}	0.22×10^5	0.14×10^5
Ultimate Strength lb./in.^{2}	0.58×10^3	0.44×10^3
Poisson's Ratio	0.17	0.06

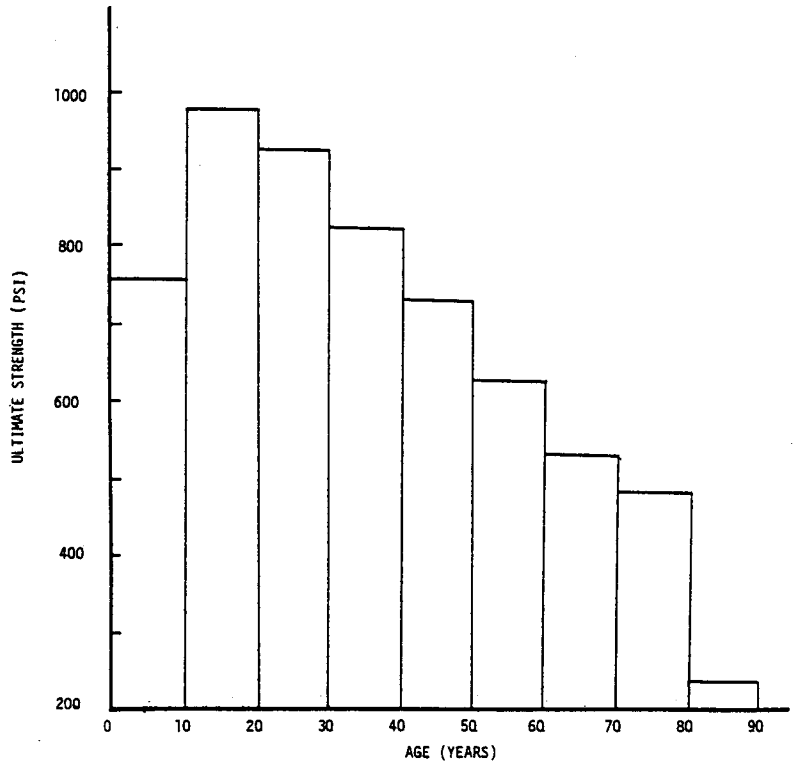

Figure 2-6. Ultimate strength versus age, vertebral bodies in compression.

There was no significant difference in the mechanical properties of the human vertebral cancellous bone when loaded in different directions. Histological studies indicated differences in trabecular patterns when sectioned in various directions, however, and work is continuing to document this observation. For many modeling purposes the cancellous bone of the vertebral body may be considered homogeneous and isotropic in the large.

The vertebral body properties also correlated well with the dry weight density. The following linear regression equations were obtained:

$$E = (3\gamma - 0.023) \times 10^6;\ C_c = 0.61$$
$$\sigma = (300\gamma - 5.9) \times 10^3;\ C_c = 0.67$$
$$\sigma = (3.2 \times 10^{-2})E \quad ;\ C_c = 0.71$$

where

E = modulus of elasticity loaded in the direction of the long axis of the spine (psi)
σ = ultimate compressive strength (psi)
γ = dry weight density (lb./in.^{3})
C_c = correlation coefficient

The vertebral body data also shows a strong linear correlation between modulus strength, indicating that a maximum strain theory of failure may be used for cancellous bone from the vertebrae with a maximum failure strain 0.03 to 0.05. These regression equations give approximately valid results for dry weight densities between 0.022 and 0.010 lb./in.^{3}

The mechanical properties of cancellous bone are strongly influenced by the structural arrangement of the trabeculae. Thus, in these tests, properties such as compressive strength and modulus are structural properties, and the large values of the standard deviations observed for these properties are primarily due to variations in the porosity and internal arrangement of the trabeculae. The similarities of the properties and histology of compact bone indicate that a single-material porous block model is justified as a first approximation in describing the relation between structure and mechanical response.[65]

The model described by McElhaney et al.[65] shows that the modulus of bone is approximately proportional to the third power of the density. Thus, small porosity changes in bone of low relative density result in only small changes in strength and modulus, while small porosity changes in bone of high relative density result in large changes in strength and modulus. The porosity distribution in a given sample of bone is much more significant in its effect on strength and modulus in bone of low relative density than in bone of high relative density.

ACCIDENT INVESTIGATIONS AND FIELD STUDIES

This section describes some of the observations made in the course of investigating serious neck injuries. These investigations generally include —

1. review of pertinent medical records and X-rays
2. patient and eyewitness interviews
3. study of the accident situation
4. analysis of the energies and velocities involved

In addition for these situations where it was felt sufficient information was available, a laboratory simulation was performed.

The types of accidents that have been studied to date are as follow:

automobile accidents	75
motorcycle accidents	12
swimming pool accidents	92

sports related	17
industrial accidents	5
miscellaneous	7

Many of these injuries result in death. Others cause permanent paralysis with attending personal and family suffering, and severe economic drain on the family and society. In order to reduce the incidence of these adverse sequelae, the initial extraction of the victim, subsequent transportation, and emergency therapy must be carefully executed. Since many of these victims have been rendered unconscious and cannot complain of neck pain, the ambulance attendants and emergency room physicians must be made aware of the injury by more subtle means — usually the recognition of associated craniofacial injuries. These craniofacial injuries are clues pointing to an underlying cervical injury and to the mechanism of injury. As stated by Forsyth et al.,[26] "The mechanism of injury is of the greatest importance, for only through a knowledge of the forces applied in each individual case is the proper management of the patient possible. A few minutes at the very onset devoted to learning the details of the accident are almost indispensible. Careful attention to the location of contusions and lacerations about the head may be the key to acquiring an understanding of the direction and intensity of the force applied through the head to the cervical spine."

Regardless of the circumstance of the accident, in most instances, the victim is propelled into head contact with some object. The position of the head and neck, the impact site, and the direction of cervical spine loading determines the resulting cervical fracture. The head and neck is either flexed, neutral, extended, laterally flexed, or rotated, and the cervical spine can be subjected to bending, compression, tension, shear, and/or torque. Impacts about the face and frontal regions tend to produce bending in extension, while flexion results from parietal or occipital contact. When the impact is "off-center," a lateral flexion and/or rotary component may also be imparted to the head and neck.[82]

Obviously, many combinations of head-neck position, impact site, and cervical spine loading can occur. From a practical standpoint, however, a review of these accidents suggests that in most instances one of the following conditions exists:

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

These basic groups are further modified by lateral bending and rotation. In a few instances, the head is not impacted and the cervical fracture is the result of direct trauma or impulsive motion of the torso.

Extension-Tension Fractures

Extension-tension fractures can occur in three ways — forceful hyperextension and fixation of the head with continued forward displacement of the body, forceful extension of the head and neck following a rear-end collision (whiplash), and extension of the head with the body submarining down and forward. Under these conditions, the head and neck are violently extended and tension is exerted on the rough anterior spinal ligament, pulling a chip of bone off the anterior, inferior margin of a verebral body[44,77] (Figures 2-7A,B).

The third type of injury is most interesting. Typically this occurs in automobile accidents when the occupant's body submarines following fixation of the face on the steering wheel or dash; he is literally hanged. This results in the classical "hangman's fracture" described by Schneider et al.[93] (Figure 2-7C). This fracture appears to occur almost exclusively in judicial hangings in which the knot of the noose is placed submentally and in auto accidents.

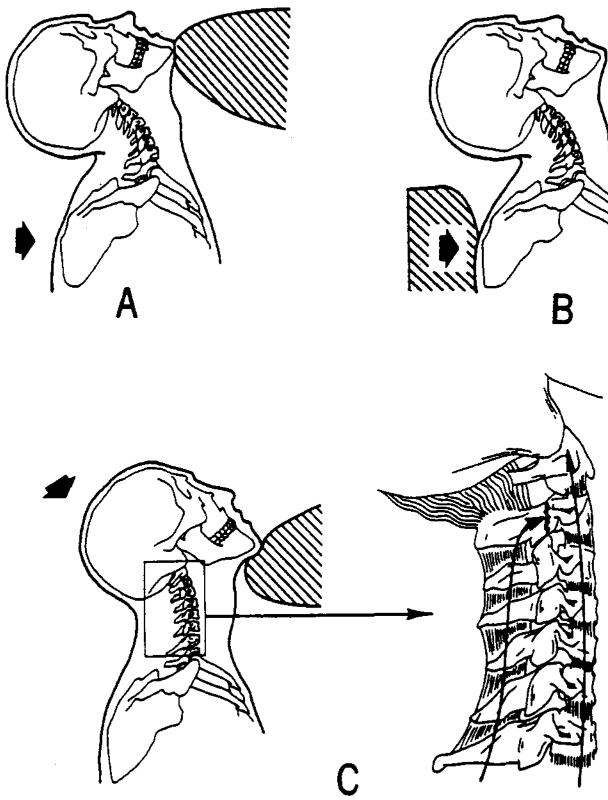

Figure 2-7. Extension-tension injuries.

Extension-Compression Fractures

Extension-compression fractures result in a spectrum of cervical fractures and dislocations (Fig.2-8), depending on the degree of head extension and rotation at impact. Midline contact tends to produce a symmetrical lesion involving the posterior elements of the vertebrae (pedicles, articular processes, laminae, and spinous processes), while a more lateral frontal contact imparts, in addition, a lateral flexion and rotational movement, which tends to load the contralateral posterior elements. This compression force is transmitted through the atlas to the cervical spine and thence along the posterior elements. These fractures are thus frequently multiple.[44,87] The degree of rotation imparted to the spine is also important, since the addition of a small degree of rotation decreases the threshold for fracture and ligament disruption.[87]

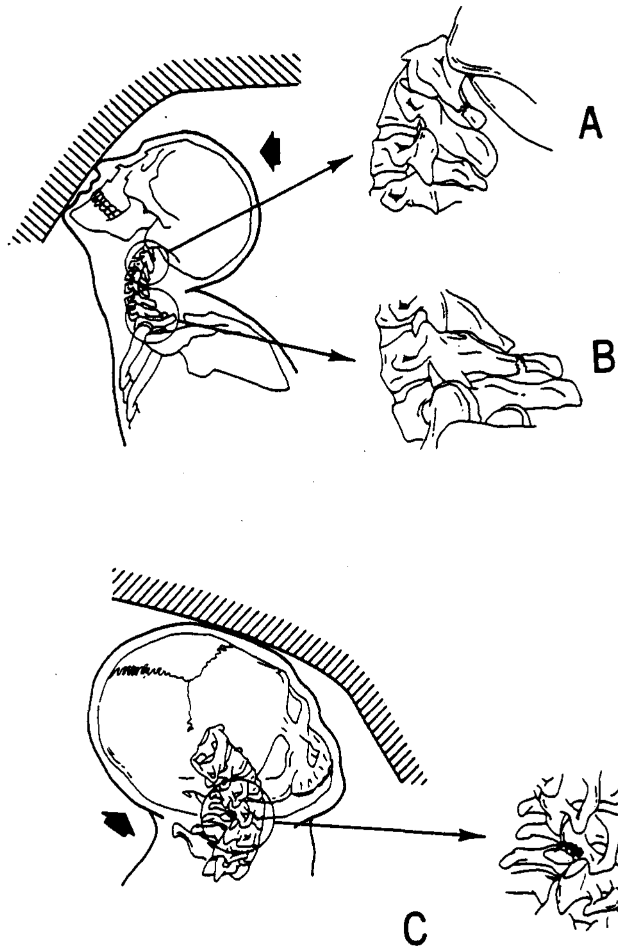

Figure 2-8. Extension-compression injuries.

In contradistinction to a flexion-compression fracture-dislocation, in which the superior articular facet of the lower vertebra is displaced forward and down, the extension-compression fracture results in a posterior and upward displacement of the inferior facet of the upper vertebra so that the facet appears more horizontally oriented on X-ray. In both types of fractures there is anterior dislocation of the upper vertebra.

Flexion-Compression Fractures

Flexion-compression fractures are caused by impacting the head at or posterior to the vertex, causing various degrees of flexion and, if off-center, lateral bending and rotation. With the head and neck in flexion, pressure is exerted on the vertebral bodies

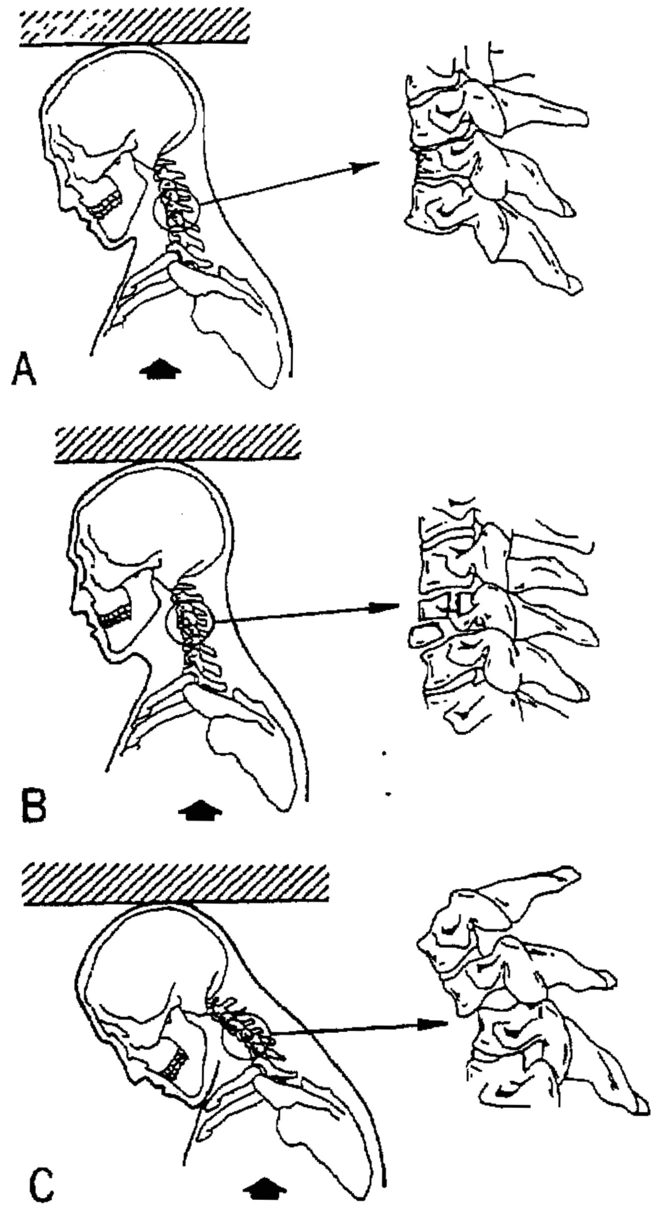

Figure 2-9. Flexion-compression injuries.

and intervertebral discs. These fractures usually occur in the lower three cervical segments, where the flexible cervical spine joins the less flexible thorax. Roaf[94] had demonstrated that the addition of a small rotational component significantly increases the incidence of fractures and ligament tears.

Initially, the compression results in a wedging of the vertebral body as it is squeezed between the segments above and below. Characteristically, the superior surface is crushed, most markedly anteriorly (Fig. 2-9A). With increasing compression, the vertebral body end-plates bulge and crack, disc material herniates into the vertebral body, and the body disintegrates, producing a bursting fracture[35,44] (Fig. 2-9B). When this occurs, an anterior fragment may displace forward, producing a "teardrop" fracture, while the posterior fragment displaces backward into the spinal cord, resulting in cord compression. Cord compression can also occur from posterior prolapse of the disc.

When the head is impacted more posteriorly in the posterior parietal or occipital regions, an additional shearing component is added to the injury. This produces an anterior dislocation with rupture of the posterior and articular ligaments, rupture of the intervertebral disc, and fracture of the superior articular facets (Fig. 2-9C).

Automobile and Motorcycle Accidents

Today, auto accidents are the most common cause of fractures and dislocations of the cervical spine. The impacted structures, force directions, and velocities are extremely varied and result in a wide range of cervical spine injuries. Because of the large numbers involved it is possible to draw good correlations between the head impact site and the type and level of cervical fracture. Frequently, there are facial or scalp lacerations that can be associated with permanent structural deformations or imprints in the vehicle. However, the vehicle motions and occupant kinematics are usually too complex to allow the detailed analysis required to estimate the impact forces, velocities, and accelerations. Thus, these accidents do not provide much neck tolerance data.

A review of eighty-seven serious neck injuries in automobile and motorcycle accidents shows the fracture pattern that could be associated with a head or facial impact site (Table 2-I).

TABLE 2-I

FRACTURE LEVEL – AUTO AND MOTORCYCLE ACCIDENTS

Level	*Low Facial Impact Extension-Tension*	*High Facial Impact Extension-Compression*	*Head Impact Flexion-Compression*
C1	2		
C1-2	9		
C2-3	4		
C2-3-4		2	
C3-4		2	2
C4	1	2	5
C4-5	1	1	6
C4-5-6		3	2
C5		6	12
C5-6		2	7
C5-6-7		1	1
C6		2	7
C6-7			1
C7			
T4-5-6-7			2
	17	21	45

In many of the motorcycle accidents, there was a clearly discernable crush area in the Styrofoam helmet liner. Occasionally, there was clearly defined imprinting of the helmet shell. Since the liner provides a permanent record of the impact pressure, we have been studying methods of interpreting it. Compression tests have been made in three cases on exemplar helmets in an attempt to duplicate the liner crush that occurred in the accident. The neck injuries were very nearly all compression with burst fractures of the body of C5. Loads of 1400 to 1800 pounds were required to approximate the crush patterns. This is a crude first attempt to establish neck tolerance in this way, but the method offers considerable promise. We are currently exploring finite element models of the helmet liner so as to better define the sources of error sensitivity and accuracy of this method.

Swimming Pool Neck Injuries

Ninety-two swimming pool accidents involving neck injury were investigated in detail. Twelve of the accidents were simulated using anthropometrically similar volunteers who performed the maneuvers that the accident investigation indicated precipitated the injury. To insure adequate safety, the simulations were done in much deeper water.

The investigation and study of swimming pool accidents can provide important information about neck injury tolerance and mechanisms. Frequently, the biomechanical factors associated with the injury can be determined with reasonable accuracy. Bounds can be established on the impact velocity, head and neck attitude, and impact position. The injury is usually well documented from medical records, while the activity that resulted in the injury is frequently observed and can be simulated and analyzed.

TABLE 2-II

FRACTURE LEVEL – SWIMMING POOL ACCIDENTS

Level	*Number*
C1	0
C1-2	4
C2-3	1
C2-3-4	1
C3-4	1
C4	4
C4-5	12
C4-5-6	1
C5	28
C5-6	18
C5-7	1
C6	10
C6-7	7
C7	1
T4-5-6-7	1
	90

Medical data including X-ray films, radiology reports, and operative summaries were studied for these injuries resulting from swimming pool accidents. The group consisted of 80 males and 12 females ranging between 9 and 54 years of age. Of these, 73 resulted in permanent quadriplegia, 2 skull fractures, 1 multiple middorsal vertebrae compression fracture without neurological involvement, 2 cervical spine fractures with some neurological deficiency, and 4 cervical spine fractures without permanent neurological damage. Table 2-II shows the distribution of the fractures with position along the spine.

Table 2-III identifies the various activities that were being performed by the victim when injured. Diving was the primary cause in 57 of the injured, while water slides accounted for 10 of the accidents.

TABLE 2-III

ACTIVITY ASSOCIATED WITH SWIMMING POOL ACCIDENTS

A. *Diving*	
1. Dive into shallow portion (4 feet or less of an in-ground pool containing variable depth)	32
2. Dive into an aboveground constant depth vinyl liner pool of 4 feet or less from attached platform, deck, coping of pool, or ladder	22
3. Dive into an aboveground vinyl pool with variable depth into shallow end from deck or attached platform	4
4. Dive from springboard 30 inches or less from water	6
5. Dive from 1 meter springboard	1
6. Dive from 3 meter springboard	5
7. Dive from pool's internal steps	1
8. A cannonball dive into shallow portion of pool	2
9. Dive from deck of pool into deep of hopper bottom pool	2
10. Dive from deck into water 5-1/2 feet deep	1
11. Struck upslope of bottom from dive off 36 inch high springboard	1
12. Dive from roof of house and balcony of apartment house	2
13. Running dive into shallow water at beach	3
B. *Water Slide*	
1. Headfirst entry into shallow water (3-1/2 feet or less)	10
2. Dive from top of slide into shallow water	1

Of the 67 cervical spine fractures, 63 were classified as compression or flexion-compression injuries. These resulted when the head struck the pool bottom or side. Typically, these injuries are caused by impacting the head in an area slightly anterior or posterior to the vertex causing various degrees of flexion and, if off-center, lateral bending and rotation. With the head and neck in flexion, pressure is exerted on the vertebral bodies and intervertebral discs.

Two had a pure flexion injury with a dislocation at C5-6 but no obvious fracture. This resulted from the head pocketing in a soft bottom and the torso violently following it.

Accident Simulations

Many of these accidents were simulated to better understand the dynamic factors involved. Anthropometrically similar volunteers performed the diving and sliding maneuvers described by the accident victims and eyewitnesses. High speed cameras (200 frames per second) above and below the water recorded the test subject's motions. In addition to duplicating the established accident kinematics, the volunteers performed a wide range of motions from the slide or diving site, jumping as high or as far as possible with various body positions. This was done in an attempt to bracket the possible head impact velocities associated with the neck injury.

A frame-by-frame analysis using a computer-coupled film analyzer was performed. The head position measurements from the film were used to construct trajectory and velocity curves for the various jumping and diving configurations.

Figure 2-10 shows a frame from the underwater cameras for a dive from the board. The background grid is 1 foot by 1 foot and the camera-to-subject distance is 25 feet.

Figures 2-11 and 2-12 show the head trajectory and velocity versus horizontal travel for the extremes of the edge of pool dives.

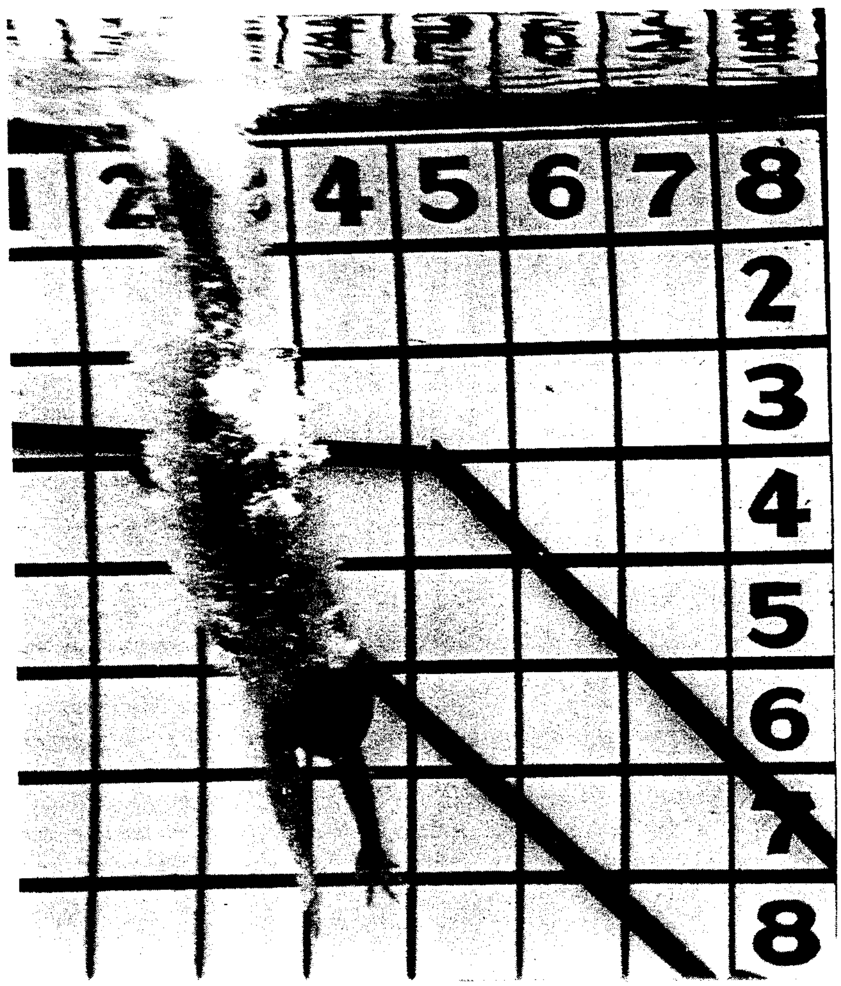

Figure 2-10. Underwater study of diving.

Figure 2-11 is the result of the test subject's effort to dive as deep as possible from a standing position at the water's edge with a clean entry. Figure 2-12 is the result of the test subject's effort to dive as far as possible without limiting his penetration.

Since 64 of the neck injuries were sustained through dives from the side of the pool into 4 feet or less of water, this situation was carefully studied. Dives of all types were analyzed, ranging from a simple falling in to full springs for maximum height and/or distance. Test subjects were chosen to match the weight, height, and sex of specific accident victims.

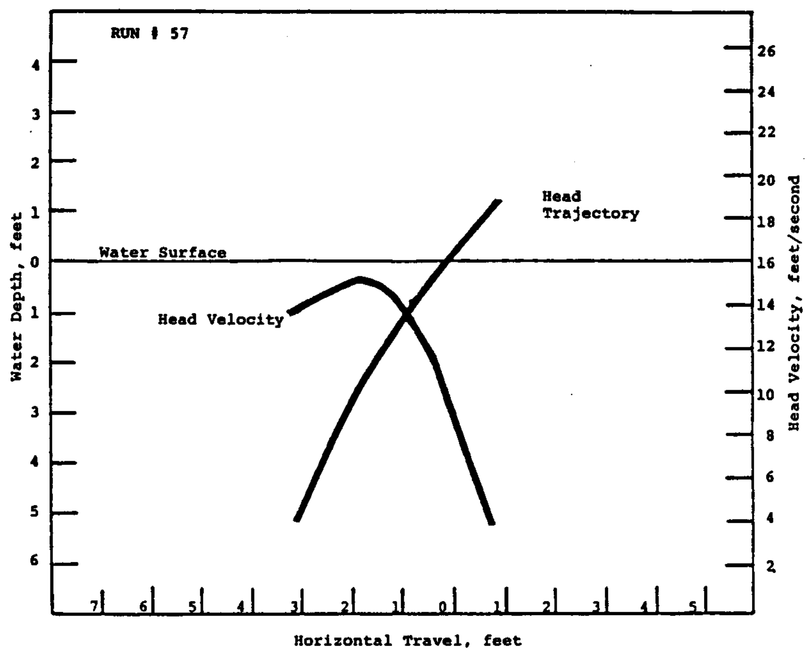

Figure 2-11. Head trajectory and velocity for a typical deep dive from the edge of a pool.

A similar study was performed on swimming pool slide activities. Commercial slides with heights ranging from 6 to 12 feet were studied. Additionally, an experimental slide with an adjustable height, length, and exit angle was used. A wide range of test subjects was used to simulate various attitudes and entrance configurations.

The mechanics of jumping is an important consideration in an analysis of this type. Gerrish[35] in an experimental study of 270 male Columbia University students, found that in a free-standing jump from a crouch, the distance that they could raise their center of mass was between 12 and 24 inches.

Batterman[6] indicates that an expert springboard diver can raise his center of mass 29 inches in a hurdle or running jump, and with the aid of a springboard, he can increase this to approximately 55 inches.

Thus, in analyzing the various accidents in this study, the velocity bounds were developed by assuming free-standing divers could raise their center of mass a maximum of 24 inches over its location when standing erect, running divers by 29 inches,

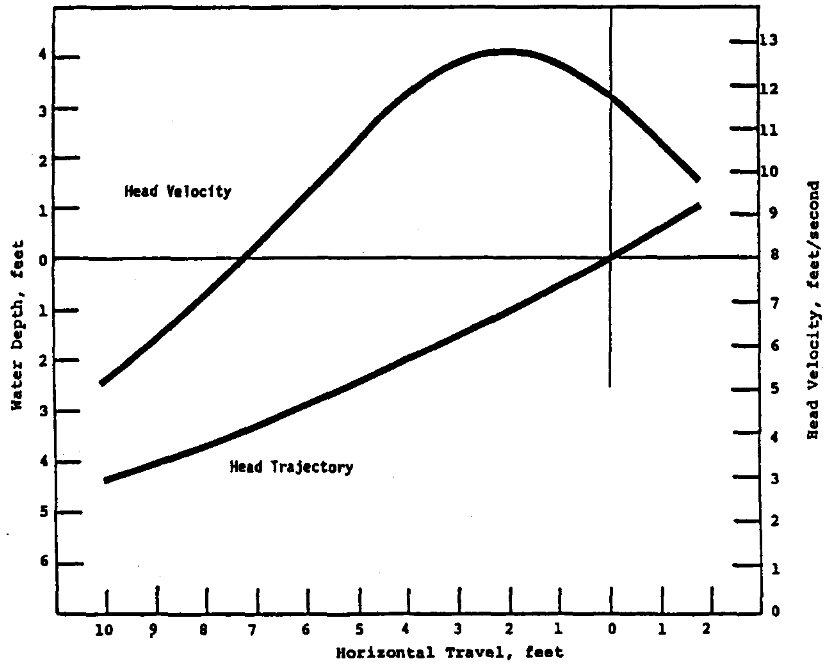

Figure 2-12. Head trajectory and velocity for a typical dive from the edge of a pool.

and springboard divers by 55 inches. These assumptions are consistent with the measurements made from the high speed camera films and are important considerations in any attempt to predict head impact velocities in swimming pool and gymnastic activities. A more detailed discussion of this work is given by Enis et al.[20] and Gabrielsen and McElhaney.[31]

Results

Fifty-eight of the cases studied were from the edge of the pool into shallow water (4 feet or less). The high speed film measurements and the analysis indicate that the fall height for the center of mass of these subjects ranged from a maximum of 7.2 feet to a minimum of 3.8 feet. Estimated head impact velocities for this group ranged from a maximum of 21.5 ft./sec. to 10.2 ft./sec. All but three of these injuries involved a compression fracture of a vertebral body, most often C5, and were classified as either a compression or flexion-compression injury. While the patients were generally aware that they struck their head on the pool bottom, there was frequently no medical record of a bruise or contusion of the head.

Sixteen of the cases were from springboards or platforms of various types, usually from a walking or running launch. The injuries all involved flexion-compression. The range of velocities estimated from the simulation and analysis were from 12.5 ft./sec. to 26.5 ft./sec. These accidents involved more unknown factors than the edge of the pool group and a much wider range of free-fall heights. The water depth was usually much deeper and the path through the water had many more possible configurations.

Ten of the accidents involved headfirst entry from a water slide into 3.5 feet or less of water. The normal body position at the water surface is with the head and hands up and the back arched. This causes the slider to skim across the surface with little penetration. If, however, the head and/or the hands are lowered, a snap roll or tumble occurs. The head and hands increase the drag, causing the body to flex at the hips and rotate toward the pool bottom. As it rotates, the eccentric drag forces increase, the hands tend to be forced down, and the head impacts the bottom, unless the water is deep enough to allow completion of the roll. The head impact velocities estimated from the photographic measurements and analysis indicate a range of 11.7 ft./sec. to 16.2 ft./sec. for the snap roll mode. Again, the neck injuries in this group were all classified as flexion-compression fracture dislocations.

It is clear from the number and severity of the accidents presented here that diving or headfirst sliding into shallow water is potentially very dangerous and should be actively discouraged. The snap roll motion probably occurred in many of these accidents. Keeping the head and hands up and the back arched is critical in shallow water diving.

The neck injuries observed were amazingly similar. There were only four head injuries and one facial trauma reported. In the head-down impacts that probably occurred in the majority of these accidents, the forces were less than that required to cause head trauma, but because of the body driving into the neck, catastrophic neck injuries occurred. In the head-up configuration, the face will impact the pool bottom with a glancing blow. Of course, this happens in the swimming pool environment. But the lack of serious reported head and neck injuries connected with facial impacts indicate that the velocity range associated with diving is less than critical for this mode of injury.

Sports-related Neck Injuries

Seventeen serious sports-related neck injuries have been investigated. Of these, seven were football related, two occurred in lacrosse games, three were on trampolines, and five were on gymnastic mats or air mats. In the football-related incidents, four were observable on game films, while the others usually had a wealth of eyewitness data. Unlike the swimming pool accidents, there were no clear injury patterns. Estimates of impact velocities ranged from 10 ft./sec., but the usefulness of these figures is obscured by the football helmets and the complicated nature of the impacting surface.

The five injuries on mats or air mats are quite interesting. Two were extension injuries with little obvious compression and neurological lesions at the C2 level. The impact site was the face and the victims in attempting a flip landed on a gym mat face first. The velocity of the face was estimated at 15 ft./sec. to 20 ft./sec. The other three injuries occurred on very soft air mats approximately 30 inches thick. These were pure flexion injuries wiht dislocaitons at the C4 and C5 level but no obvious bony fractures. It is theorized that these injuries occurred when the head pocketed in the soft air mat and the torso flexed the neck. Head impact velocities were estimated to be 15 ft./sec. to 22 ft./sec., based on eyewitness descriptions of the activity.

Experimental Simulations

In order to develop analytical methods of estimating the velocities, accelerations, and forces involved in the accidents described above, a variety of helmets, foams, and mats have been tested. Figure 2-13 shows the load-deformation responses for a variety of helmet types for vertex loading. Test time was 0.2 second with constant velocity. These tests were performed using the American National Standards Institute Z89 headform and provide a basis for estimating their stiffness and damping characteristics.

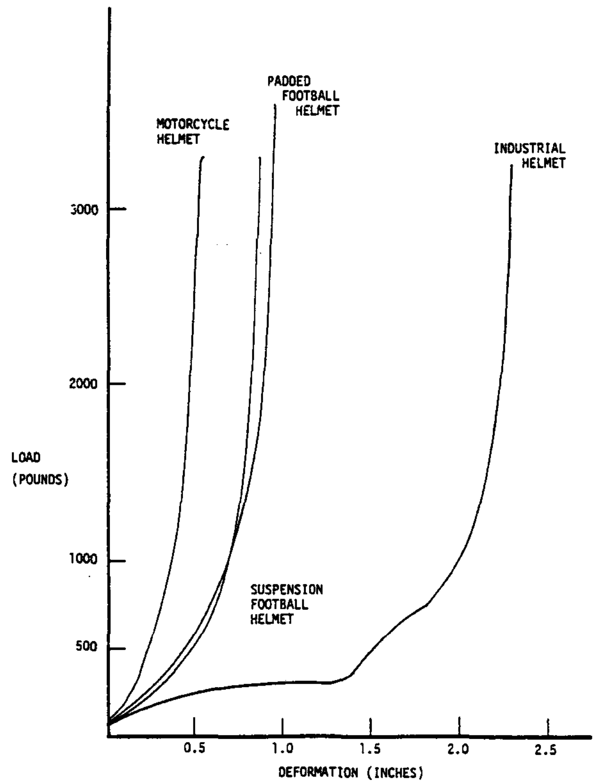

Figure 2-13. Helmet load-deformation characteristics in compression.

Additional data is being developed through laboratory simulation of accidents using instrumented anthropomorphic dummies. We have simuated some of the diving, gymnastic, and football accidents using a modified automotive crash test dummy. A more realistic neck and head were used.[66] Compliance in the superior-inferior direction was added to better match the human neck (see Fig. 2-1). The neck was instrumented with compression and flexion transducers, and the head incorporated a triaxial accelerometer. High speed photography allowed detailed kinematic analysis. Figure 2-14 shows typical data from tests where a 5% Alderson dummy impacted a rigid steel plate at 8 feet per second with the head slightly flexed. The head and neck protection offered by a helmet is clearly demonstrated. Not the flexion-extension rebound. We are continuing to refine our accident simulation methods, but at this time we do not feel confident enough to offer represenive neck tolerance estimates from this source.

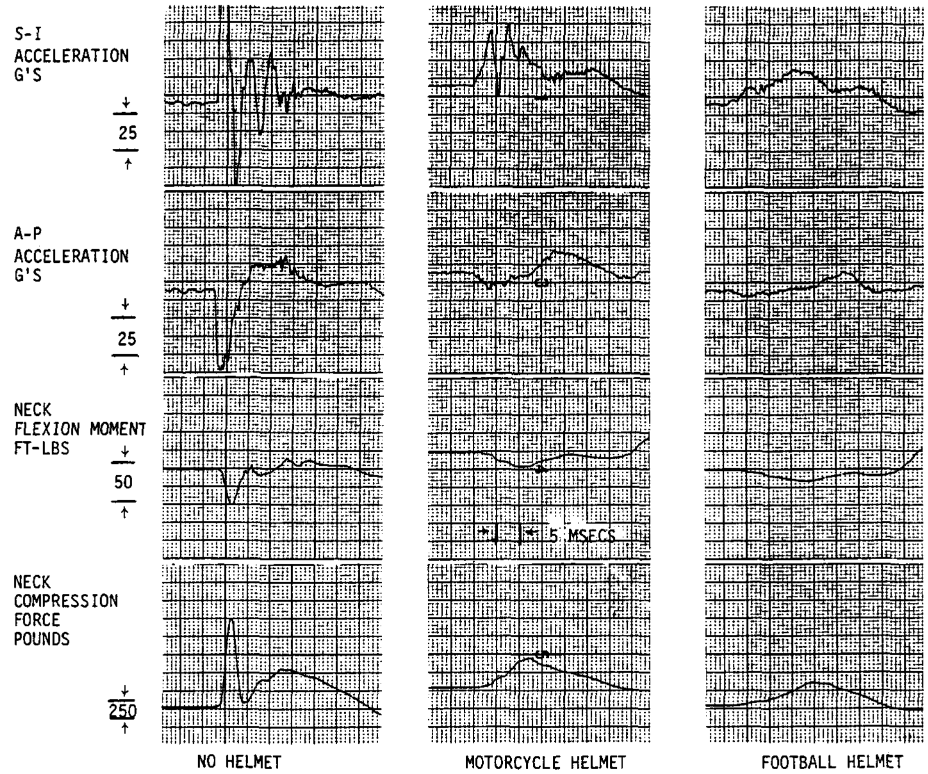

Figure 2-14. Instrumented dummy responses (V = 8 ft./sec.).

Mathematical Modeling

A lumped-parameter mathematical model aimed at representing uniaxial compressive loading of the helmet, head, neck and torso has been developed. This model consists of a helmet mass coupled by a Kelvin element to the head, which is coupled by a Kelvin element to the torso. The model can be exercised both with head represented by a pure mass or the MSC Head Injury Model developed by McElhaney et al. Head injury potential can be estimated through SI, HIC, or MSC calculations.[67] Impact entails contact with a relatively immovable object of varying stiffness. Head accelerations and neck loads as affected by different helmet model parameters, contact surface stiffness, impact velocities, and effective dynamic torso masses can then be estimated (Fig. 2-15).

The model assumes that the cervical spine is in an anatomically straightened position, that the related muscular and ligamentous structure offer negligible load-bearing support in compression, and that the helmet, head, neck, and effective dynamic torso mass motion is confined to one dimension and coincident with the neck axis. Under these conditions, the torso and upper and lower limbs are lumped into a single reduced mass termed the effective dynamic torso mass. Table 2-IV lists the model constants.

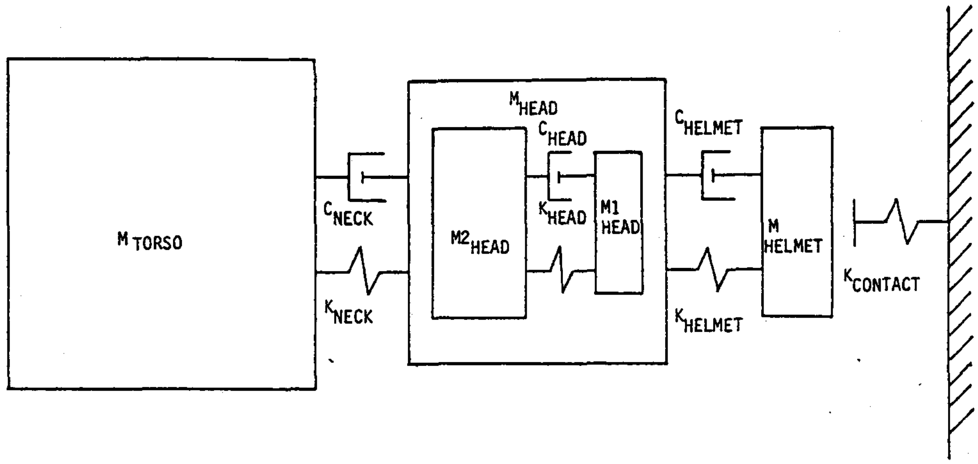

Figure 2-15. Head/neck injury model.

Table 2-IV

MATHEMATICAL MODEL PARAMETERS

HEAD/NECK INJURY MODEL PARAMETERS

MHelmet (lb.)	*MHead* (lb.)	*MTorso* (lb.)	*KContact* (lb./in.)	*KHelmet* (lb./in.)	*KNeck* (lb./in.)	*CHelmet* (lb.-sec./in.)	*CNeck* (lb.-sec./in.)
2.0	10.0	50-100	500-5000	500-5000	2500	1.0	1.0

MSC HEAD INJURY MODEL PARAMETERS

Species	*M1Head* (lb.)	*M2Head* (lbs)	*KHead* (lb./in.)	*CHead* (lb.-sec./in.)
Human Lateral	0.4	9.0	26000	2.4
Human Longitudinal	0.6	10.0	50000	2.0

EMPIRICAL HELMET PARAMETERS

	STIFFNESS		
Helmet	*Initial* (lb./in.)	*Overall* (lb./in.)	*Equivalent Viscous Damping* (lb.-sec./in.)
Motorcycle	2000	1000 - 15000	0.17 - 0.30
Padded Football	1000	650 - 1350	0.19 - 0.27
Suspension Football	750	500 - 6000	0.07 - 0.25
Industrial	500	150 - 1000	0.08 - 0.21

The helmet and neck stiffnesses and damping constants have been calculated from the cervical spine load-deformation measurements previously described and tabulated in Table 2-IV. Linear spring characteristics were computed from the loading curves. Equivalent viscous damping constants were computed from the hysteresis loops. Neck model constants were computed in the same manner from load-deformation measurements. Contact surface stiffness (K_o = 500 lb./in., K_o = 2500 lb./in., and K_o = 5000 lb./in.) were chosen to simulate collisions with soft surfaces as observed in gymnastics, with other players and helmets as in football, and with very rigid surfaces as in swimming pool and automotive accidents. Effective dynamic torso masses (W = 50 pounds and W = 100 pounds) were selected to simulate situations where the motion of the torso and extremities is not coincident with the neck axis. Selected impact velocities (VO = 120 in./sec., VO = 240 in./sec., and VO = 360 in./sec.) range from the critical velocity required to break the neck in swimming pool accidents to the higher speeds attainable by well-conditioned athletes.

Figures 2-16 and 2-17 show the model predictions for head acceleration and neck compression as a function of helmet stiffness. The last points are the values obtained without a helmet. The initial conditions for the model exercises were impact velocities of VO = 120 in./sec., VO = 240 in./sec., and VO = 360 in./sec. The contact surface stiffness for Figure 2-16 was K_o = 2500 lb./in. The contact surface for Figure 2-17 was K_o = 5000 lb./in. The effective dynamic torso mass was W = 50 pounds and W = 100 pounds as noted.

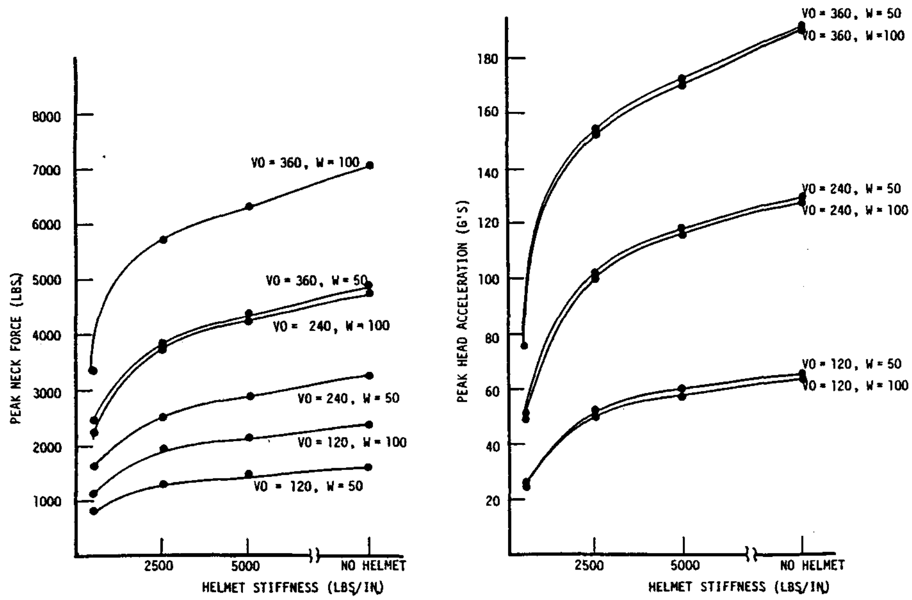

Figure 2-16. Head/neck injury model results for contact surface stiffness K_o = 2500.

This model indicates that torso mass only slightly influences head deceleration but strongly influences neck compression. It also indicates that helmet stiffness and effective crush distance strongly influence both head deceleration and neck compression.

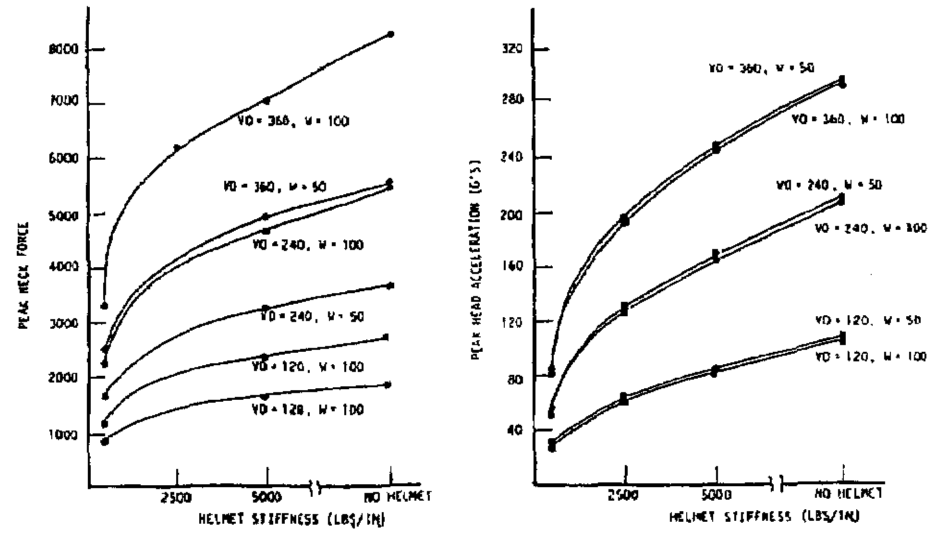

Figure 2-17. Head/neck injury model results for contact surface stiffness K_c = 5000.

The helmets with stiffnesses of 500 lb./in. required effective crush distances that were impracticably large. However, the required crush distances for the helmets with stiffnesses of 2500 lb./in. and 5000 lb./in. were all between 1 and 2.5 inches.

DISCUSSION

There are many real-life situations that can lead to serious neck injuries. Some of these, when carefully investigated, analyzed, and simulated, can provide valuable information about neck injury mechanisms and tolerance. The location of the head or facial impact site can frequently be used to predict the type and level of neck injury. Thus the location of contusions and lacerations about the head may be the key to acquiring an understanding of the direction and intensity of the force applied through the head to the cervical spine. This knowledge can then be used to establish proper X-ray techniques and reduction.

Sports and swimming pool accidents are frequently well documented and sufficiently uncomplicated that accurate simulation is possible. Our research in swimming pool accidents has shown that head crown impacts of 10 ft./sec. into rigid objects can cause serious neck injuries when the torso is free and following. We have also concluded that suitable energy-absorbing materials in the form of mats or helmets can offer some protection of the head and neck. Instrumented dummy tests provide a method for the evaluation of the protective potential of candidate systems. The conclusion that energy-absorbing materials offer neck protection has recently been challenged by Hodgson and Thomas.[43] Our results are in agreement with Mertz et al.[72] In over seventy-five instrumented dummy tests we have observed a high degree of variability in the neck compression and flexion loads. Careful alignment of the torso, neck, and head with the impacted surface removes much of this and provides repeatable results. Perhaps this is the explanation of the differing opinions of various researchers.

The head accelerations and neck compression loads show a potential phasing problem in helmet design. Under some circumstances, depending on the characteristics of the helmet and/or the struck surface, the head can rebound in the helmet before the torso has stopped. This rebound phenomenon can enhance the neck loading. Incorporating adequate damping properties in the helmet can significantly reduce the rebound.

The mathematical model described here, although quite simple, provides a method to predict the effect of varying parameters. The predicted waveforms of head and neck loading compare reasonably well with the experiments results when the neck flexion moments are low. When neck flexion predominates, this model becomes inappropriate.

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