CERVICAL SPINE PROTECTION REPORT
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INTRODUCTION

This report documents the current understanding of the injury tolerance of the human cervical spine. It also documents the state-of-the-art by which surrogate devices and models may be used to mimic the mechanical behavior of the human neck. Analysis and synthesis of these published observations are then provided to define what we believe is our current ability to evaluate the efficacy of existing and prototype devices which are thought to mitigate neck injury potential in sports.

Characterization of the mechanical properties and injury criterion of the cervical spine has been a challenge to our community. Numerous classifications of injury and mechanisms of injury have been developed, including Moffatt et al., 1971, Portnoy et al., 1971, Roaf, 1972, Babcock, 1976, Dolen, 1977, Allen et al., 1982, Harris et al., 1986, White and Panjabi, 1990. The rationale for these classifications has included retrospective reviews of patient data, whole and segmental cadaveric experimentation, and retrospective evaluations of various known injury environments (McElhaney et al., 1979, Allen et al., 1982, Torg, 1982, and Yoganandan et al., 1989). Methodologies for evaluation of the spine are numerous and frequently fail to adequately document the boundary and initial conditions of the experiments. Unfortunately, the responses of the spine are highly sensitive to these conditions (McElhaney, et al., 1983). Indeed, changes in the initial position of the point of impact, and the end condition have been shown to change the injury produced (Nusholtz et al., 1983, McElhaney, 1983, Myers, 1991). Characterization of injury is further complicated by the recognition that the observed motions and forces on the head do not reflect the motions or true injury mechanism of the cervical spine (Nightingale et al., 1995).

In this context, we define injury tolerance as the maximum internal load at a particular level of the spine, or maximum motion of a particular motion segment that the cervical spine can tolerate without incurring any injury. The load may be applied externally, or may be generated by inertia of the accelerating head, neck and torso. These external forces and body forces result in internal loads that cause injury at a particular level in the cervical spine. The internal load has been termed the Major Injury Vector (MIV), (White & Panjabi, 1990). In its original concept, MIV was a load vector, but presently we use MIV as a representation of both a load and motion injury tolerance, at a spinal level at which injury occurs. We define major injury load vector as a six component load vector consisting of three force components and three moment components in a local vertebral coordinate system. Similarly, we define major injury motion vector as a six component vector consisting of three translations and three angulations of one vertebra with respect to its lower neighboring vertebrae in the local coordinate system. It is also necessary, to complete the above definition of the injury threshold, to define the word "injury". We suggest that injury is a permanent change in the mechanical properties of the spine or its components: ligaments, disc, facet capsules, muscles, nerves, and arteries, as well as vertebrae.

Seen from the mechanical viewpoint, there are three types of experimental studies. Kinematic studies are those in which only the motions are documented. These studies provide motion injury tolerance. Kinetic studies document both the motions and the loads simultaneously, in the form of load-deformation curves. In contrast to the kinematic and kinetic studies, which are under physiological loads causing no injury, the trauma studies result in injury and provide the load injury tolerance.

Accordingly, the following sections review the current understanding of kinematic and kinetic motion and its role as a motion based tolerance of the cervical spine. The trauma studies including studies on human volunteers, whole cadavers, and the isolated cadaveric ligamentous cervical spines are also reviewed. The use of physical devices and mathematical models as surrogates for the human subject in the evaluation of injury potential is then discussed. These observations are then synthesized and our current understanding of our ability to describe and evaluate neck protection devices is discussed.

KINEMATICS & KINETICS STUDIES

The kinematic or motion studies have been conducted on human subjects (*in vivo* studies) or have utilized human cadaveric spinal segments (*in vitro* studies). The *in vivo* studies have been mostly the active kind, i.e. the subject is asked to bend and twist the spine into an end position. In the *in vivo* passive studies, the examiner pushes the spine into an end position, beyond the normal active motion. In both situations, x-rays are taken to document the position. The motion produced passively is greater than the corresponding active motion. Therefore, wherever possible, we should choose the passive ranges of motion to define the injury tolerance. The *in vitro* studies have used either fresh cadaveric functional spinal units (two vertebrae spinal segment) or whole cervical spine specimens. In general, the *in vitro* studies are better controlled and have higher accuracy, and therefore result in more reliable data. However, they lack the effects of muscle.

The vertebra has six degrees of freedom, consisting of three translations and three rotations respectively along and about the axes of a coordinate system. The motion presented is the relative motion, i.e. of one vertebra moving with respect to the other vertebra below it. Anatomically, the occipital-atlanto-axial joint complex is one integrated unit, e.g. the alar ligaments connect the dens of the C2 vertebra to the occiput. However, for kinematic purposes, we will present the information at each level separately.

Motion studies of the cervical spine have been carried out for a long time, (Fick 1911, Poirier and Charpy 1926, and Werne 1957). Because of the limited accuracy of these earlier studies, they are neither described nor their results included for consideration in this document. Below we provide a description of some modern *in vivo* and *in vitro* studies of the cervical spine. It has been difficult to measure motions at all cervical spinal level in a subject or a spinal specimen. In the case of the living person it is difficult to clearly identify the contours of the occiput and the T1 vertebra. For the spine specimen the methodology has not been available until recently to apply loads that do not constrain the spinal motions and measure the resulting large three dimensional motions. To our knowledge, not a single study, *in vivo* or *in vitro*, exists in which the motions of the entire cervical spine have been measured in all three planes, i.e. sagittal, frontal and horizontal. In some studies, motions were measured only in the upper cervical spine (Occiput-C1-C2), while in others in the middle and lower cervical spine (C2-C7). Therefore, it became necessary to construct a composite table in which main rotatory motions at all levels and in the three planes are assembled.

In Vivo Studies

Penning (1978) examined a group of 20 healthy young adults by taking lateral x-rays of their cervical spines in flexion and extension postures. He developed a technique of super-imposing the two films and computing the ranges of motion from occiput-C1 to C6-C7. Although he also graphed the centers of rotations, he provided no numerical values. His results are presented in Table 1.

Dvorak and associates (1987) studied 9 healthy adults (age: 17 - 49 years) and 43 patients with cervical spine injury using functional CT scans. The ranges of axial rotation at the levels of occiput-C1, C1-C2, and C2-T1 were measured. The results are given in Table 1.

Penning and Wilmink (1987) measured axial rotation at each vertebral level in 26 healthy volunteers (age: 20 -26 years). CT scans were taken throughout the length of the cervical spine with the head turned maximally to the right, and then to the left. To our knowledge, this is the first and only *in vivo* study in which axial rotations of the cervical spine have been measured. The results are presented in Table 1.

Dvorak and colleagues (1988) introduced the concept of *in vivo* passive motion measurements in contrast to the normally used *in vivo* active motion measurements. The cervical spines of 59 adults were examined by means of functional radiographs. They studied 28 healthy adults and 31 patients who had sustained soft tissue injury to the cervical spine. By laying the flexion x-ray on top of the extension x-ray and using simple graphical construction, they measured the rotational ranges of motion in flexion-extension. On average, the passive motion was about 2-3 ° larger than the active motion. Results of their study are given in Table 1. The paper also provides us with the results of older *in* vivo studies by Bakke (1931), Buetti-Bauml (1954), Deseze (1951), and Penning (1960).

Dvorak and associates (1991) measured motions of the cervical spine in 44 subjects (22 women and 22 men) in younger age group (23 - 49 years). These were normal subjects who underwent passive flexion-extension motion and were examined by lateral x-rays at the ends of the range of motion. Using digitization and a computer program, ranges of rotatory and translatory motions as well as centers of rotations were determined. The rotational ranges of motion data is provided in Table 1.

In vitro Studies

Lysell (1969) was probably the first modern investigator who conducted a thorough *in vitro* study of the cervical spine motions and motion patterns. He provided a comprehensive review of the older literature and credits Weber (1827) as performing the first real measurements of spinal motions. Lysell used whole fresh cadaveric cervical spine specimens (C2-T1). Using four 0.8 mm steel balls inserted into each vertebra and quantitative stereo radiography, he measured the three dimensional relative motions between vertebrae. He studied a total of 28 specimens (age: 11 - 67 years) and found no effect of degeneration on the motions. The results of this classical study are presented in Table 1.

Panjabi and associates (1986) measured the load-displacement curves of the cervical spine specimens (4 spines, age: 42 - 70 years). The loads applied were six forces: anterior-posterior shear, medial-lateral shear, and compression-tension. The maximum force used was 50 N. Using a novel technique they measured three-dimensional motions of each motion segment of the cervical spine specimens. The resulting rotatory motions are given in Table 1. Additionally, they also measured the translations as well as neutral zones for both the rotations and translations. Flexibility coefficients were also computed. This data could be used for the development of physical surrogates and mathematical models of the cervical spine.

Panjabi and co-workers (1988) , in another *in vitro* study, measured motions of the upper cervical spine in three dimensions. Ten fresh cadaveric spine specimens (age: 29 - 59 years) were used and subjected to a maximum moment of 1.5 Nm at the occiput, while the three-dimensional motions of occiput-C1 and C1-C2 were measured, using stereophotogrammetry. In addition to the ranges of motion, which are provided in Table 1, the authors also measured the elastic and neutral zones. These later parameters may be helpful in validating the cervical spine surrogates, especially the representation of the nonlinear behavior of the spine.

Moroney and associates (1988) obtained 35 fresh adult cervical spine motion segments. As the age of the specimens was not known, they categorized the specimens by degeneration grading scheme on a scale of 1 - 4. Their specimens had an average degeneration grade of 2.5 with standard deviation of 1.0 They obtained load-displacement curves by applying forces of compression and shear and moments of flexion, extension, lateral bending, and axial torsion. The maximum shear forces applied were about 20 N, while the compression force was about 75 N. All moments applied were 1.8 Nm. The results are given in Table 1. In addition to the intact motion segment behavior, they also measured the kinematic behavior of the disc alone, after cutting the posterior elements. The authors also present the stiffness values of the spinal segments as well as the failure loads. As mentioned above, this data could be used for the development of surrogates and mathematical models of the cervical spine.

Goel and colleagues (1990) studied the C0-C1-C2 complex under the application of axial torque until failure. The experimental design allowed for unconstrained motion of the occiput. They studied a total of 12 fresh cadaveric specimens (age: 50 - 91 years) and documented the moment-rotation relationship. The average motion measured at 2.0 Nm, was 29.9 degrees with a standard deviation of 11.3 degrees. The failure load was 13.6 Nm with a standard deviation of 4.5 Nm. They measured the motion of the occiput-C2, and not the individual motions of the occiput-C1 and C1-C2. This is justified, as their focus was to determine the failure characteristics of the upper cervical spine complex.

White and Panjabi (1990) reviewed the literature on kinematics of the spine. They realized inadequacy of the available data concerning the cervical spine and produced a set of values ranges of motion that they thought best represented the motions of the spine. This compilation is also given in the Table 1, as a reference.

Recommended Ranges of Motion

As the results provided in Table 1 clearly show, there is not a single study that gives us complete ranges of motion for the entire cervical spine. We have made a composite table that synthesizes the data of Table 1 into recommended ranges of motion (Table 2). The C0-C2 and C1-C2 motion for all three motion types, is taken from Panjabi et al (1988). The C2-C3 to C6-C7 flexion-extension ranges of motion are taken from Dvorak et al (1988). The motion of the last segment of the cervical spine (C7-T1) in flexion-extension is taken from Lysell (1969). The lateral bending ranges of motion for C2-C3 to C7-T1 are also taken from Lysell (1969). Finally, the axial rotation ranges of motion for levels C2-C3 to C7-T1 is taken from Penning (1987).

TRAUMA STUDIES

As discussed, the responses of the cervical spine are highly sensitive to the initial orientation of the spine, the direction, and location of the applied forces. In this context, understanding injury requires an unambiguous nomenclature, and careful analysis of experimental design. In the following discussion, the nomenclature used will refer to the principle applied loading of the motion segments, i.e. the Major Injuring Vector (MIV), and not the observed motions of the head, or the loads on the head required to produce the resultant head motions. A detailed analysis of several important publications will be presented in an effort to outline the strengths and weaknesses of each study. These will include studies of human volunteers, whole cadaver cervical spines, and the isolated head and ligamentous cervical spines.

Volunteer Studies

Much of tolerance to flexion-extension loading stems from the dissertation thesis of Dr. Mertz at Wayne State University, published in Mertz and Patrick, 1967, and Mertz and Patrick, 1971. These papers summarize data from anterior-posterior and posterior-anterior restrained sled decelerations of Dr. Larry Patrick and a number of cadavers in which the head was weighted by additional mass at, above, or below the head center of gravity. The authors determined the flexion-extension bending moments at the craniocervical junction by solving the inverse dynamic problem.

For flexion, Dr. Patrick noted the onset of pain during a deceleration which produced 59.4 Nm. This was defined as a pain threshold. The largest deceleration pulse that Dr. Patrick was willing to be exposed to produced 88.1 Nm, and resulted from a 9.6 G deceleration with additional mass added above the head center of gravity. He noted the immediate onset of pain, and was sore for a number of days following but suffered no permanent disability. This was therefore defined as an injury threshold. The largest deceleration applied to a cadaver resulted in 190 Nm of moment without evidence of structural injury by conventional x-rays. This was suggested at the upper bound of tolerance to the ligamentous spine despite a lack of injury. It was noted that the chin frequently was on the chest during the deceleration, however, the method by which this is accounted for is not discussed.

For extension, the authors suggest a dynamic tolerance of 47.5 Nm based on twice the voluntary static load a subject could apply to themselves using a pulley system. A threshold for ligamentous injury of 56.9 Nm was suggested based on a scaled test from a smaller than 50th percentile cadaver in which posterior ligamentous injury occurred in the mid-cervical spine during a deceleration with a peak moment of 33.4 Nm. Notably the largest deceleration which Dr. Patrick was willing to be exposed to produced an extension moment of approximately 13.6 Nm.

The author's suggest tolerance to tensile and compressive forces of 1.13 kN and 1.11 kN, respectively, based on the maximum force a subject was willing to apply statically using a weight pulley system. As these are below the forces required to produce injury, they clearly represent a lower bound of tolerance. The author's selected a tolerance of 2.00 kN of anterior-posterior shear based on the absence of x-ray evidence of injury to 4 cadavers exposed to peak shear forces during flexion deceleration of 1.42, 1.59, 1.74, and 2.10 kN.

Instrumented studies of chiropractic cervical manipulations reported by Triano and Shultz, 1990, show upper cervical moments of 22 to 24 Nm, and resultant forces of 120-130 N without injury. Notably, these manipulations are performed with great constraint of the result motions, and loads are frequently applied directly to the cervical spine. Thus, their role in determining tolerance is uncertain.

Dr. Gabrielsen, 1990 studied 486 diving injuries. Sixty-seven of the diving accidents have been reconstructed by McElhaney et al., 1979, by observing anthropometrically similar volunteers performing the same dives in deeper water. Dive kinematics were quantified using 200 frame/s film camera. Analysis of dives from the pool edge, spring boards and water slides all suggested minimum head impact velocities of 3.1 to 3.8 m/s to produce compression and flexion injuries. This represents an equivalent drop height of approximately 0.5 m.

Whole Cadaver Studies

Whole cadaver studies have been performed by a number of authors. These studies provide useful kinematic data, particularly considering that the effects of head geometry, and head and torso inertia are accounted for. They also provide data on head force, and potentially, the injury rates associated with falls with a particular drop height or impactor momentum. Additionally, they frequently produce clinically observed fracture patterns. In contrast, these studies provide limited information regarding neck tolerance owing to the lack of spinal instrumentation. Further, it is difficult to quantify the boundary conditions and initial position of the spine at impact.

Lang, 1971, performed 15 simulations of frontal collisions with cadavers placed on automobile bucket seats and 3-point seat belts. Head accelerations were measured with an instrumented helmet. Sled accelerations varied from 20 to 27 G while torso acceleration varied from 22 to 51 G. In 10 of these tests a steering wheel was in front of the cadavers and the head hit the wheel. A variety of neck injuries were observed including fracture of the dens, disc and ligament ruptures, and vertebral body fractures. Another five head-on collisions were simulated without a steering wheel. Chest accelerations ranged from 28 to 47 G. More extensive tearing of cervical ligaments was observed with head rotations as large as 123°. Eight cadavers were tested in rear end simulations with sled accelerations ranging from 19 to 29 G and helmet accelerations between 24 and 47 G. No head rest was used and extreme hyperextension was observed. Cervical tension extension injuries were quite varied ranging from minor to severe.

Hodgson and Thomas, 1980, performed a series of impacts and quasi-static tests to helmeted whole cadavers in which a portion of the cervical spine was exposed and recorded with high speed film. His results are based on measurement of vertebral body strain at three vertebral levels, and while these cannot be used for tolerance, the conclusions are significant. Specifically, they note that a range of padding of the crown of the head did not mitigate neck injury, that increasing the degree of constraint imposed by the contact surface increased the likelihood of injury, that the guillotine theory in which lower cervical posterior element fractures result from impingement of the helmet on the back of the neck was unfounded, and that head load did not correlate with vertebral body strain.

Cheng et al., 1982, tested six unembalmed cadavers by applying a distributed frontal load to the chest through a predeployed driver airbag system. A sled was used to apply decelerations of 32 to 39 G. Severe neck injuries were found in 4 of the 6 cadavers. Observed cervical injuries included C1-C2 fracture separations, ligament tears and ring fractures with spinal cord transections.

Nusholtz et al., 1983, performed impacts to the vertex of 12 unembalmed cadavers. A 56 kg guided impactor was used with impact velocities of 4.6-5.6 m/s. Compression injuries occurred with the head motion of extension or flexion occurring later in the impact. Head impact loads ranged from 1.8 to 11.1 kN. Nusholtz et al., also performed free fall drop tests on 8 unembalmed cadavers with the initial impact to the crown of the head. Drop heights that varied from 0.8 to 1.8 m produced extensive cervical and thoracic fractures and dislocations. Head impact forces varied from 3.2 to 10.8 kN. Nusholtz noted that small changes in initial orientation influenced neck kinematics. He also suggested that energy absorbing materials reduce peak head impact force but do not necessarily reduce the amount of energy transferred to the neck. He further suggested that the complex nature of cervical spine kinematics and damage patterns in vertex impact may preclude the determination of a single tolerance criterion such as maximum force and the force measured at the head was a poor determinate of neck injury risk.

Alem et al., 1984, conducted two series of impacts to the heads of 19 unembalmed cadavers in the superior to inferior direction. They used a 10-kg impactor which struck the vertex of the head approximately in line with the spinal axis at speeds between 7 and 11 m/s. In most of the tests, the impactor was padded with 5 cm thick ensolite foam to control the duration of the contact force and to minimize local skull fractures. Compression fractures and tearing of the anterior longitudinal ligaments occurred with impact forces of 3.0 to 6.0 kN. It has been suggested on the basis of this work that while impact force on the head was a poor predictor of neck injury that impulse was an improved predictor. It should be recognized however, that in the case of head impact, impulse describes the momentum of the impactor. Accordingly, larger impactor momentum is not surprisingly associated with increased risk of neck injury.

Yoganandan et al., 1986, conducted a study to evaluate the mechanism of spinal injuries with vertical impact. Sixteen fresh intact human male cadavers were suspended head down and dropped vertically from a height of 0.9 to 1.5 m. In eight out of the sixteen specimens, the head was restrained. Head impact forces ranged from 3.0 N to 7.1 kN in the unrestrained and from 9.8 kN to 14.7 kN in the restrained specimens. There were more upper thoracic and cervical fractures in the restrained compared to the unrestrained case. Cervical fractures were observed when the cadavers remained in contact with the impacting surface without substantial rotation or visible rebound. A mean neck force of 1.73 ± 0.57 kN was measured in six of the specimens in which a cervical spinal transducer was used.

Isolated Ligamentous Cervical Spine Studies

As discussed, mechanical analysis has suggested that neck injury potential exists for equivalent drop heights as low as 0.50 m. Unfortunately, most impact situations have considerably larger impact energies. In these situations, the head-neck complex must either move out of the path of, or be at risk for injury from, the energy of the impinging torso. The interaction of the head with the contact surface, the end condition, influences the ability of the head-and neck to avoid the impinging torso, the type of injury that results, and the structural stiffness and fracture force of the spine. Accordingly, any discussion of injury tolerance must carefully control and define this variable. In the following discussion, the importance of end condition in neck injury tolerance is discussed. It should be noted that while a compressive failure load is reported, the MIV of injury may be the resulting bending moment and shear forces produced by this load.

Compression Tolerance

Roaf, 1960, was unable to produce lower cervical pure ligamentous injury in unconstrained flexion. He also loaded single cervical spinal units in compression, extension, flexion, lateral flexion, horizontal shear and rotation. Selecki and Williams, 1970, produced several clinically observed injuries using a hydraulic jack. Unfortunately they report pressure without ram diameter. Clemens and Borow, 1972, in deceleration studies of isolated cadaver torsos with a synthetic instrumented cranium coupled to the base of the skull report that anterior-posterior accelerations of 15 G applied to the thorax (frontal collisions) produced cervical injury. Interestingly, the injury was a recoil extension of the head which occurred after the head had flexed due to the deceleration, and thus, was fully preventable with a head rest. Hodgson and Thomas, 1980, suggested that restriction of motion of the atlantoaxial joint greatly increased the risk of injury. Sances et al., 1982, tested isolated cadaver cervical spines in compression with fixed end conditions. Flexion-compression failures were reported at loads ranging from 1.78-4.45 kN.

McElhaney et al., 1983, performed studies of isolated cervical spines including the base of a the skull through T1. A flexion preload was applied to straighten the spine and mount it in the test frame with a fixed-fixed end condition, and an anterior-posterior translation stage was used to preposition the base of the skull relative to the torso. These experiments demonstrated the sensitivity of injury mechanism to the initial position in that the authors observed changes from extension injuries to flexion injuries with movements of the translation stage as a small as 1 cm. Using displacement rates ranging from 0.13 to 64 cm/s resulted in both upper and lower cervical injuries at a mean load of 3.86 ± 1.36 kN.

Maiman et al., 1983, performed compression tests to 13 cadavers including either the entire head, or a portion of the skull with a horizontal cast surface. Specimens were tested in either a pre-flexed, pre-extended or neutral position. Load rate used varied from 0.25 cm/s to 152 cm/s, and impactor position was varied from 2.5 cm anterior of vertex to 2 cm posterior to vertex. Specimens with intact skulls were tested with a free end condition, while the actuator was seated in the methacrylate casting of specimens with partial skulls. Impactor load to failure varied from 0.65 to 7.4 kN. Lower loads and larger piston displacements were observed in the specimens which were pre-flexed or pre-extended. Yoganandan et al., 1989, in a similarly designed experiment, compared the cadaver responses with those of the Hybrid III. A mean force to failure of 2.10 kN was reported for three specimens.

Pintar et al., 1990, measured the compression failure loads of six excised head-cervical spines. The head was intact and loads were applied through the vertex of the skull. The head was upright and stabilized through a weight and pulley system mounted anteriorly and a spring mounted posteriorly. High speed loading rates of 0.3 - 8.1 m were used. The resting lordosis of the spine was removed prior to impact. Mean compressive neck loads of mean = 3.5 ± 1.9 kN were measured at failure. Head impact force was 11.8 ± 5.7 kN. The high variability of the head load reflects the increasing acceleration imposed on the head by the differing impactor velocities. Variability in the neck loading rate may also influence the reported failure force as a result of viscoelastic effects.

Myers et al., 1991, performed a series of quasi-static experiments (rates 0.05 cm/s) in which the effect of end condition on injury type and force at failure were quantified. Three end conditions were evaluated on specimens comprising T1 to the base of the skull: unconstrained (fixed-free), fully constrained (fixed-fixed) and rotationally constrained (fixed-free translation, fixed rotation). In the unconstrained tests, no injury resulted despite an average imposed axial displacement of 8.6 cm. Full constraint resulted in compression injuries with a mean axial (vertical) load to failure of 4.8 ± 1.3 kN. Rotation constraint resulted in bilateral facet dislocations (BFD) at a mean vertical load of 1.7 ± 1.2 kN. These data illustrate the significance of the effect of end condition in quasi-static injury and injury tolerance. It also illustrates the importance of coupled motions, and the major injuring vector in producing neck injury. Specifically, the rotation constraint group developed large flexion bending moments and suffered a flexion bending moment MIV, a BFD, in the lower cervical spine as a result of a combined compression force and extension moment applied to the base of the skull. It should be noted that the axial force that produced the lower cervical flexion bending moment MIV should be viewed as an absolute lower bound on human tolerance; this, because the extension moments which are produced in the human by the passive responses of the lower cervical paraspinal muscles are absent in the cadaveric test. The muscle generated extension moment would oppose the flexion bending moment MIV caused by the compressive load, and therefore a larger compressive load would be required to produce the injury.

Panjabi et al., 1991, impacted 13 isolated upper cervical spine specimens including the base of the skull to C3. In these experiments a falling mass impacted a padded, instrumented impounder which was coupled to the base of the skull using a polyester casting. Testing was performed with the spine in the neutral position or with a 30 degree wedge interposed to place the spine in an extended position. A total of 10 specimens suffered Jefferson like fractures with a mean impactor load of 2.9 ± 0.6 kN.

Yoganandan et al., 1991, used a padded impactor to apply forces to intact head and neck specimens. A halo was mounted to the head to control the head prior during impact. A posterior-anterior directed spring in the occipital portion of the halo was pretensioned by a mass coupled to the anterior of the halo by an anteriorly directed cable using a pulley system. Impact velocities of 0.53 - 0.85 m/s resulted in peak axial neck forces of 4.47 ± 0.89 kN at displacements of 3.4 ± 0.6 cm.

Recent work by Nightingale et al., 1995, provides useful data on neck injury tolerance. Whole head and neck cadaveric specimens were mounted to a 16 kg effective dynamic torso mass with a multi-axis load cell. Impact surface stiffness was varied and orientations included 15 degrees rotation posterior to vertex (posterior superior portion of the head), vertex, and 15 degrees anterior rotation (anterior superior portion of the head). Specimens dropped from a height of 0.53 m were reliably injured. A broad spectrum of fractures were produced including multi-part atlas fractures, lower cervical compression fractures, lower cervical bilateral facet dislocations, and posterior element fractures. Despite the wide variation in the types of injuries produced, the axial force to fracture showed only a small variance, 1.7 ± 0.4 kN in seven specimens from donors with a mean age of 53 years. Additional data collected by these authors has resulted in a mean force of 2.0 ± 0.6 kN for a total of nine injuries. Injury occurred within 3 - 8 ms following head impact when the head impacted a rigid contact surface. In padded impacts, neck injury occurred after 14 - 19 ms following the head impact. They noted that the large head motions, the type which could be observed or recording using conventional techniques, did not occur until 20 - 100 ms following the injury. Not surprisingly, the resulting head motions did not correlate with neck injury MIV.

Buckling of columns has long been recognized as a cause of structural failure in engineering structures. Engineering analysis demonstrates that buckling in and of itself is not a material failure however. Material failure, injury, is more likely a function of the post-buckling stability of the structure (Chen and Lui, 1987). Buckling has long been thought to play an in important role in cervical spinal injury (Torg, 1982). Experimentally, buckling has been demonstrated in the cervical spine in both static and impact experiments (Myers et al., 1991, Nightingale et al., 1995); however, while the buckling influenced the kinematics of the spine, it did not result in injury in either the static or dynamic tests. Accordingly, the role of buckling in the cervical spine remains unknown.

The role of head constraint plays an important role in neck injury and understanding neck injury tolerance. As noted, Roaf, 1960, was unable to produce pure ligamentous injury in unconstrained specimens. Hodgson and Thomas, 1980, suggested that restricting the motion of the atlantoaxial joint increased the likelihood of injury. Yoganandan et al., 1986, suggested that increasing constraint increased the measured axial load and the number of injuries observed in cadaver experiments. McElhaney et al., 1988, noted that increasing the constraint of the end condition increased the axial and flexural stiffness of the spine. Myers et al., 1991, demonstrated that constraint of the cranial end condition increased axial stiffness, and axial force to failure. They also suggested that constraint of the head increased the likelihood of injury and surfaces and systems should be designed to minimize the degree of pocketing imposed during head impact. Nightingale et al., 1995, though insufficient for statistical analysis, further supported the hypothesis that pocketing of the impact surface may decrease the ability of the head and neck to escape injury by moving out of the way of the oncoming torso.

Tolerance to Other Forms of Loading

Tensile loading of the cadaveric cervical spine has been performed by a few authors producing both lower and upper cervical injuries. Mean load to produce occipito-atlantal ligamentous injuries of 5 complete cervical spines reported in Sances et al., 1982, was 1.5 $\pm$ 0.5 kN. Shea et al., 1991 report a tensile load to failure of 0.50 $\pm$ 0.15 kN in quasi-static cadaver tests of spines with an initial pre-extension of 30 degrees and a mean extension moment of 3.9 $\pm$ 3.1 Nm in a combined loading test frame. Both these results must be considered as a lower bound on the tolerance however, given the absence of passive muscle tone. Mertz and Patrick, 1971, suggest a tensile tolerance of 1.13 kN in tension during extension loading (i.e. posterior-anterior acceleration of the torso). Clemens and Burow, 1972, suggest a value of 1.6-2.0 kN, based on the experiments described above. Using a tensile stiffness of 1.64 kN/cm reported by Bowman et al., 1984, from volunteer tests conducted at the Naval Biodynamics lab, NBDL, and a mean tensile displacement to failure reported in Shea et al., 1992, of 1.88 cm, Myers et al., 1994, estimated that the human tolerance to tension may be as high as 3.1 kN.

Estimates of the lower bound of torsional tolerance are available from Myers et al., 1991. Axial torque of 17.2 ± 5.1 Nm produced upper cervical injury in cadaver whole cervical spines (occiput to T1). Goel et al., 1990, report a slightly lower value of 13.6 ± 4.5 Nm in isolated upper cervical spine studies; however, the rates of loading in these experiments were considerably lower than in the previous study (4 degrees/s versus 500 degrees/s). Extrapolating piecewise linear torsional stiffness data from volunteer decelerations reported by Wismans and Spenny, 1983, to the 114 ± 6.3 degrees of rotation required to produce injury in the cadaver, McElhaney and Myers, 1993, suggested an estimate of human torsional tolerance of approximately 28 Nm. They also suggested that the 114 degrees of axial rotation from the neutral position required to produce upper cervical injury in whole cervical spines mitigated the role of small axial rotations as a mediator of lower cervical injuries.

Tolerance data for horizontal shear is limited and stems primarily from cadaveric data collected on occipito-atlanto-axial injuries. In particular, loading required to produce transverse ligament failure and odontoid fracture have been reported by a number of authors. Fielding et al., 1974, observed transverse ligament rupture at 0.82 kN, when the atlas is driven anterior relative to the axis in the isolated cadaver upper cervical spine. Doherty et al, 1993, produced odontoid fractures at 1.51 ± 0.42 kN by applying posterior-lateral directed loads directly to the odontoid process. Understanding the relationship between these results, and the neck loads which develop dynamically in impact has not been determined. As discussed, Mertz and Patrick, 1971 suggest a lower bound of tolerance of 2.0 kN. Cheng et al., 1982, in a cadaveric study of neck injury from decelerations produced by a distributed load to the chest, report cervical injuries in four specimens with estimated shear loads of 2.8 ± 1.76 kN in the presence of total resultant loads of 5.5 ± 2.5 kN.

SURROGATES AND THEIR ROLE IN UNDERSTANDING INJURY

The complexity of the spine and the injury environment has created a need for repeatable, robust testing methods for the development of safety equipment. This need has been an impetus for the design and development of a group of mechanical, and computational models of both cervical and total body dynamics.

Physical Surrogates

The initial mechanical surrogates were based on the 50th percentile male, consisted of solid masses of butyl rubber, and had little biofidelity. Significant design improvements followed (Culver et al., 1972, Melvin et al., 1972) and in 1977 Foster et al., introduced the Hybrid III Anthropometric Test Device (ATD). These devices were designed and validated based on the kinematics of non-injurious, non-impact decelerations of belted occupants. The devices have then been used to determine the kinematics of larger, potentially injurious, decelerations. Neglecting the effects of this extrapolation, it is apparent that these devices are kinematic simulators. Their utility is limited to determining velocity and accelerations of the head and neck in non-impact decelerations. While this information provides the basis for determining the risk of non-contact head and neck injury (e.g. “the head injury criterion”), it does not provide insight into the risk of neck injury, or head injury during impact except from the standpoint of a prediction of impact velocity. Indeed, a number of investigations have shown the Hybrid III neckform to be 1 to 2 orders of magnitude stiffer than the human volunteer and cadaver cervical spine in compression (Yoganandan et al., 1989, Myers et al., 1991, Wismans et al., 1987, Svensson and Lovsund, 1992).

Realizing that load follows stiffness, data on neckform loading from impact of these devices cannot be directly related to human neck loads in real world accidents. Recognizing this limitation, Mertz et al., 1978, proposed criterion on which Hybrid III neckform data may be interpreted based on impact testing of the Hybrid III dummy into a football tackling device. Using this criterion, measured neck load is weighted by the duration of the neck pulse. Peak instantaneous neckform loads of 6.67 kN are linearly decreased to a minimum load of 1.11 kN for pulse durations of 35 ms for the young. For the adult population, peak instantaneous neckform loads of 3.56 kN are reduced to a minimum load of 1.11 kN for pulse durations of 30 ms. These pulse durations are large with respect to the durations of force necessary to produce cervical spinal injury (Nightingale et al., 1995), however, because the device is undamaged during loading, the use of these pulse durations may be appropriate. It should be noted that to date, no investigation has demonstrated that neck impulse correlates with neck injury risk. In either case, the validity of this neckform criterion in predicting human injury risk under similar conditions is the subject of further investigation.

Kabo and Goldsmith, 1983, developed a head neck model capable of three dimensional motions which they instrumented and validated for sagittal plane impact loading. The device, in addition to being accurate from an anthropometric standpoint was designed to provide data on intracranial fluid pressure, intervertebral disk pressure, and simulated muscle deformation. Liang and Winters, 1991, have recently produced a complete anthropometric head-neck replica from engineering materials including individual vertebral bodies, disks and passive and active muscle elements. The active muscle components are represented by a total of 10 pneumatic actuators. Preliminary data suggest that the device shows similar kinematic behavior to the cervical spine. The suitability of this device for impact studies has not been discussed however. Recent efforts have been conducted by a research group at Chalmers University of Technology, Sweden. They have developed a head and neck physical surrogate which has been validated for rear end collisions (Svensson and Lovsund, 1992). This device has not been designed for head impact. Efforts to design an improved dummy neck for the evaluation of neck injury potential due to impact are currently being conducted at NHTSA/VRTC. Unfortunately, each of these devices is a prototype device, not currently in production for use in evaluation of neck injury potential.

Mathematical Surrogates

Mathematical surrogates of the spine have evolved from three basic groups: continuum models, lumped parameter models, and finite element models. An extensive review on this subject is provided by Yoganandan et al., 1987. The continuum model of the spine was first developed in the aviation industry in the late 1950's to determine the relationship between emergency pilot ejection and the risk for spinal injury (Hess and Lombard, 1958). This model evolved over subsequent years to include viscoelastic material properties (Terry and Roberts, 1968), head mass (Liu and Murray, 1966), varying stiffness along the length of the column (Shirazi, 1971), an initial curvature of the beam (Moffatt et al., 1971, Li et al., 1971), the contributions of the musculature (Soechting and Paslay, 1973), the effects of transverse and rotary inertia (i.e. full 2-D analysis) (Cramer et al., 1976), and the effect of load orientation on measured stiffness (Liu and Dai, 1989). The goal of this approach is to couple the stress and acceleration determined from the model to the risk of spinal and head injury from estimates of human tolerance. The obligatory simplification of the spine to a continuous beam is frequently cited as a limitation in this approach. The absence of acceleration or stress based criteria limits the interpretation of these models. Thus, while these are able to give insight into expected forces and accelerations from impact, as well as provide insight into experimental data, an accurate determination of the risk of cervical injury under the conditions modeled cannot be made.

The first lumped parameter model was developed in the late 1950's by Latham. Subsequent work increased the number of degrees of freedom and allowed for two dimensional analysis (Orne and Liu, 1971, Vulcan and King, 1971, Kaleps et al., 1971, McElhaney et al., 1983, and Zhu et al., 1990). These models, which include the individual structures of the spinal motion segment, have been developed to describe aircraft ejections, whiplash injuries, and the more general problem of the three-dimensional dynamics of the spine to an arbitrary loading history (Panjabi, 1973, Prasad and King, 1974, Belytschko et al., 1978, and Reber and Goldsmith, 1979). More recently, Burstein and Otis, 1994, developed a linear, one dimensional neck model which suggested that the design requirements of devices which would shunt load directly from the head to the torso following head impact were beyond currently available technologies. Similarly, using a nonlinear one dimensional model, Bishop and Wells, 1989, suggested padding deformations of 6 to 17.5 cm without bottoming out would be required to mitigate neck injury in impacts with velocities between 1.8 and 3.0 m/s. Historically, limitations of existing computation facilities required restrictions in the number of parameters, choice of material properties, and geometric complexity of these models. Recent efforts however, have become increasingly complex, and include elements which proscribe facet joint kinematics, and muscle tractions (Seireg and Arbikar, 1975, Williams and Belytschenko, 1983, Liu et al., 1986, Gudavalli and Triano, 1990). Total body models based on numerical solutions to multibody equations of motion, either Lagrangian or Eulerian, have also been developed. These include the Crash Victim Simulator, the Articulated Total Body Model, PAM-Crash, Madymo 3D and others (McHenry and Naab, 1967, Bowman, 1971, Robbins et al., 1974, Belytschko and Privitzer, 1978, Wismans et al., 1982, De Jager et al., 1994). These models incorporate cervical spines of varying complexity.

The recent availability of relatively affordable, high speed computation has resulted in an increased popularity of finite element modeling. Finite element models of the spine have been used by a number of authors to assess a variety of problems. These include, the performance of spinal instrumentation, the influence of surgical intervention on spinal stability, the role of torsional loading on injury and the influence of disc degeneration on spinal kinetics (Balasubramanian et al., 1979, Shirazi-Adl, et al., 1986, Ueno and Liu, 1987, and Kim et al., 1991). The bulk of this effort has been limited to the lumbar and thoracic spine however (Yoganandan, 1987).

More recent efforts have focused on cervical spine modeling for the assessment of injury potential (Williams and Belytschko, 1983, Kleinberger, 1993). Finite element modeling promises to play an increasingly important role in the study of the behavior of the spine.

Clearly, considerable effort has resulted in a wide variety of models. While a number of these have been validated against volunteer deceleration data, and therefore represent useful kinematic simulators, we are unaware of a single neck model which is validated to accurately predict the risk for neck injury for head impact. This is in large part the result of an absence of experimental data suitable for such model validation. As with all modeling efforts, experimental validation provides the only way to access their usefulness. Accordingly, we are not recommending computational methods to analyze the problem of neck injury in sports at this time.

SYNTHESIS OF CERVICAL TOLERANCE

Kinematic Data

Using Table 2 data, we have computed the upper limits (mean value plus 2 standard deviations) and lower limits (mean value minus 2 standard deviations) for all spinal levels and motion types, Table 3. From the viewpoint of the injury causing motions, we interpret the lower and upper limits in the following manner. The lower limit of a range of motion represents the motion at a specific intervertebral level that would not cause injury in most people. The upper limit is the threshold motion if the spine is bent or twisted beyond this threshold, there will be injury in most persons.

Trauma Data

The cervical spine is a multisegmented column with nonlinear structural properties. Its geometry is complex, and its constituent elements have nonlinear material properties. Cervical injury mechanisms have been shown to be sensitive to the initial position of the neck, the direction of loading, the degree of constraint imposed by the contact surface, and possibly to the rate of loading. In addition, a variety of host related factors contribute to injury biomechanics. These include, the bone mineral content of the vertebra, the presence of degeneration, the degree of muscular stimulation at the time of impact, and the variance associated with the geometric differences within the population. Cadaveric experimentation together with the analysis of falls and dives demonstrates that neck injury can occur as a result of drop heights as low as 0.5 m, or a velocity of 3.1 m/s (McElhaney et al., 1979, Nusholtz et al., 1983, Yoganandan et al., 1986, Nightingale et al., 1995).

In sports related injury, perhaps more so than the automotive environment, axial force is likely the most important determinant of neck injury risk (Torg et al., 1982). Initial analysis of data in the reported literature suggests dramatic variation in reported axial forces. Including the whole cadaver tests results in reported loads from 0.95 kN to 14.6 kN. Careful analysis of these results considerably reduces the scatter and provides a preliminary basis for an understanding of cervical spinal tolerance.

First and foremost is the effect of inertial loading from either the head or the impactor. Impactor or head force data are considerably larger than the applied neck force, because the neck force is reduced from the head/impactor force by the product of head / impactor mass and head acceleration. Accordingly, data must then be interpreted with the realization that neck loading will be greatly reduced and will depend heavily on the inertial characteristics and accelerations of the head and impactor. This effect was observed by Yoganandan et al., 1986, and Nightingale, et al., 1995. The later observed peak head loads of 5.9 $\pm$ 3.0 kN while measured neck loading at failure was 1.73 $\pm$ 0.565 kN. Nightingale et al., 1995, also noted that peak head loading occurred prior to the development of significant neck load. Accordingly, data on head or impactor loading should not be used as a measure of cervical spinal tolerance.

Using only measured neck load results in reported tolerance of the cervical spine of ranging from 1.73 kN to 4.47 kN (McElhaney et al., 1983, Yoganandan et al., 1986, Pintar et al., 1990, Myers et al., 1991, Panjabi et al., 1991, Yoganandan et al., 1991, and Nightingale et al. 1995). Consideration of the degree of constraint, and the rate of loading imposed by the testing frame profoundly reduces the scatter of these reported values. Indeed, the neck force at fracture of 4.47 kN is reported by Yoganandan et al., 1991, in which high rate impact tests are performed on cadavers with constrained whole heads. McElhaney et al., 1983 report force at failure of 3.86 kN in fully constrained straightened spines. In contrast, Nightingale et al., 1995 in impacts of unconstrained heads and necks into padded and unpadded surfaces report force to failure of 1.95 kN. Similarly, Yoganandan et al., 1986, report compressive force of 1.73 kN in head and neck impacts without head constraint. Panjabi et al., 199l, in a test with limited constraint report tolerances of 2.86 kN. These data suggest that human cervical spinal tolerance to impact from falls is within the range of 2 kN to 3 kN and more likely in the lower end of this range (2 to 2.5 kN).

The effects of age should also be considered. Typical cadaveric experimentation uses specimens with a mean donor age in the fifth decade. Data from McElhaney et al., 1970, suggest that the mechanical properties of cancellous bone is 45% stronger when comparing specimens from 20 year old donors to specimens from 50 year old donors. Keaveny and Hayes, 1993, report that cortical bone is approximately 10% stronger when comparing the same age groups. While the relative contributions of cortical and cancellous bone to the structural properties of a vertebral body is the subject of debate, these data suggest that scaling the cadaver responses by a factor of 20 to 30% is appropriate. Accordingly, we suggest that cervical spine tolerance of 2.5 kN to 3.125 kN be used for the tolerance of the young human cervical spine.

Thus, with consideration of the rate of loading, the degree of constraint imposed by the contact surface, and the injury mechanism, considerable insight into cervical compressive tolerance can be gained. These results should be tempered realizing that this compressive force tolerance does not provide insight into the site of injury, the type of injury, or its clinical sequel. The basis for this limitation stems from the recognition that the compressive force has a point of application which varies along a curved spine, and therefore gives rise to bending moments and shear forces. Unfortunately, the injury results from one of these MIVs or as a result of the combination of these MIVs. Future work, in which forces, eccentricity of forces, and MIV are reported at the site of injury will be required if we are to predict neck injury in greater detail. Also of importance will be the effect of the passive properties of the spinal musculature on injury tolerance.

RECOMMENDATIONS FOR NECK INJURY PROTECTION

A lack of availability of cadaveric tissue for the study of injury mechanisms and tolerance has been the primary limitation on the size of most studies, and perhaps the greatest limitation to our understanding of neck injury. Biomechanical evaluations of neck injury are also hampered by limited financial resources. As a result, reported studies frequently fail to provide adequate data on neck tolerance. For example, in the kinematic data, we notice that at each level and in all three motion directions, there is significant variation as shown by the relatively high values of the standard deviations compared to the mean values (Tables 1 and 2). This is especially true in motions of lateral bending and axial rotation, where the standard deviation is often about 50% of the mean value. In this report we have made an effort to synthesize these studies and derive both motion based MIV tolerance, and load based MIV tolerance. Clearly, new studies should be designed and carried out that are especially tailored to the development of both kinematic and load based tolerances of the cervical spine.

Because of the difficulty and limitations in performing cadaveric investigations, physical and computational models of the cervical spine are becoming important elements in safety engineering. It is evident that much progress has been made with both physical and computational modeling; however, it is also evident that much remains to be done. It may be reasonably argued that existing anthropometric test devices, particularly the Hybrid III, may be used to predict neck injury from head impact using compression criterion developed for those devices (Mertz et al., 1978, Bishop and Wells, 1990; Bahling et al., 1990). Further, it may be argued that the effects of padding the head may be evaluated using this test fixture. However, it should also be recognized that considerable uncertainty currently exists, particularly with regard to the use of moment based criterion, and that continued efforts to define tolerance criterion for the human neck for the ATD neckforms are required.

Of particular importance to the design of sporting safety equipment, is the relationship of the head, neck and torso. It has been suggested that devices or systems which transfer load directly from the head to the torso may mitigate neck injury. In contrast, a considerable portion of the literature suggests that increasing the constraints placed on the neck decreases the ability of the head and the neck to escape the moving torso and thereby increases injury risk. Clearly, these are opposing design criterion. In order to be useful for the evaluation of neck injury potential and the potential mitigating influence of equipment on neck injury risk, the ATD must adequately reproduce the *in vivo* responses of the head, neck, torso and shoulder complex. This response, must consider the ability of the head and neck to escape injury. It must also consider the mechanical properties of the torso-shoulder complex as the ability to transfer load from the head directly to the torso is a function of head, neck, torso and safety device stiffness. Despite the importance of the torso and shoulder stiffness to injury prevention strategies, we are unaware of any biomechanical data describing these relationships. Further, we are unaware of any study which evaluates the biofidelity of the torso and shoulder complex of any ATD to vertical loading. Accordingly, the biofidelity of existing devices in this form of loading is unknown at this time.

Given the fundamental mechanical axiom that load follows stiffness, we cannot advocate a standard test device or model which adequately characterizes these effects. However, legislation of a standard device for evaluation of safety equipment will no doubt influence the design of the equipment. Given the influence of a standard on design, the standard must be based on a device which adequately mimics the true biomechanical behavior of the spine. We do not believe that a single device or model adequately meets all of biomechanical criterion necessary to allow the evaluation of neck injury risk or the evaluation of the efficacy of an arbitrary safety device in preventing neck injury. Accordingly, we do not recommend the development of a safety standard at this time.

However, we do recommend a course of action that, if followed, will lead to the development of standards for the testing of neck protection devices. Studies need to be designed specifically for the purpose of developing neck injury protection standards. The new kinematic and kinetic studies would provide the true characteristics of the cervical spine, head and shoulder, that must then be incorporated into the design of the physical and mathematical surrogates. These studies would include the inherent nonlinearity of the cervical spine mechanical characteristics. The new trauma studies need to be conducted with well defined end-conditions, simulating the specific sport activity for which the standard is being developed. This is necessary as the trauma tolerances differ with respect to the direction and kind of loading. The new trauma studies should provide the MIV values at the site of injury, and not just the external load causing the injury.

REFERENCES

Alem NM, Nusholtz GS, and Melvin JW. Head and Neck Response to Axial Impacts, The 28th Stapp Car Crash Conference, 275-287, 1984.

Allen BL, Ferguson RL, Lehmann TR, O'Brien RP. A mechanistic classification of closed indirect fractures and dislocations of the lower cervical spine. Spine 7(1):1-27, 1982.

Babcock JL. Cervical Spine Injuries. Arch. Surg. 111:646--651,1976.

Belytschko TB, Schwer L, and Privitzer E. Theory and application of a three-dimensional model of human spine. Aviat. Space and Environ. Med. (1),158, 1978.

Belytschko TB, Privitzer E. Refinement and Validation of a 3-D Head Spine Model, Contract AF33615-76-C506, AMRL, WPAFB, Dayton Ohio, 1978.

Bahling GS, Bundorf RT, Kaspzyk GS, Moffatt EA, Orlowski KF, Stocke JE. Rollover and drop tests - The influence of roof strength on injury mechanics using belted dummies. The 34th Meeting of the Stapp Car Crash Conference. 1990 101-112.

Balasubramanian K, Ranu HS, King AI. Vertebral response to laminectomy. J Biomech 12, 813,1979.

Bishop PJ, Wells RP. A computer simulation model for studying cervical spine injury prevention. AGARD Conference Proceedings No. 471, Neck Injury in Advanced Military Aircraft Environments. 18-1, 18-5, 1989.

Bishop PJ, Wells RP. The inappropriateness of helmet drop tests in assessing neck protection in head-first impacts. Am J Sports Med. 18(2): 201-205, 1990.

Bowman BM. An analytic model of a vehicle occupant for use in crash simulation. Ph.D. Dissertation, Dept. of Eng. Mech., University of Michigan. 1971.

Bowman BM, Schneider LW, Lustak LS, Anderson WR, and Thomas DJ. Simulation Analysis of Head and Neck Dynamic Response. The 28th Stapp Car Crash Conference, 173, 1984.

Burstein AH, Otis JC; The response of the cervical spine to axial loading: feasibility for intervention. American Society for Testing and Materials, Symposium Head and Neck Injuries in Sports, ASTM STP 1229, 1994.

Clemens HJ, Burow K. Experimental Investigation on Injury Mechancisms of Cervical Spine at Frontal and Rear -Front Vehicle Impacts. Society of Automotice Engineering Transactions Paper # 720960, 78--104, 1972.

Chen WF, Lui EM. Structural Stability, Theory and Implementation. Elsevier New York, 1987.

Cheng, R., Yang, K.H., Levine, R.S., King, A.I., Morgan, R. Injuries to the cervical spine caused by a distributed frontal load to the chest. Proceedings of the 26th Stapp Car Crash Conference, SAE Paper #821155, 1982.

Cramer HJ, Liu YK, von Rosenberg DU. A distributed parameter of the inertially loaded human spine. J Biomech 9:115, 1976.

Culver CC, Neathery RF, Mertz HJ. Mechanical necks with human-like responses. Proceedings of the 16th Stapp Car Crash Conference, SAE 720959, 61-75, 1972.

Doherty BJ, Esses SI, Heggeness MH. A biomechanical study of odontoid fractures and fracture fixation. Spine, 18(2):178-84, 1993.

Dolen KD. Cervical spine injuries below the axis. Rad. Clin. No. Am. XV(2):247-259, 1977.

Dvorak J, Fröhlich D, Penning L, Baumgartner H, Panjabi MM. Functional radiographic diagnosis of the cervical spine: flexion / extension. Spine, 13 (7): 748-755, 1988.

Dvorak J, Hayek J, and Zehnder R. CT-functinal radiographic diagnosis of the rotary instability of the upper cervical spine. Part 2. An evaluation on healthy adults and patients with suspected instability. Spine, 12 (8):726, 1987.

Dvorak J, Panjabi MM, Novotny JE, Antinnes JA. *In vivo* Flexion / Extension of the Normal Cervical Spine. JOR, 9:828 - 834 1991.

Fielding JW, Cochran GVB, Lawsing JF III, Hohl M. Tears of the transverse ligament of the atlas. Journal of Bone and Joint Surgery, 56-A(8):1683-1691, 1974.

Fick R. Handbuch der Anatomie und Mechanik der Gelenke unter Berücksichtigung der bewegenden Muskeln. T.3. Fischer, Jena 19ll.

Foster JK, Kortge JO, Wolanin MJ. Hybrid III - A biomechanically based crash test dummy. Proceedings of the 21st Stapp Car Crash Conference, SAE 770938, 975-1014, 1977.

Gabrielson MA. Diving injuries: The etiology of 486 case studies with recommendations for needed action. NOVA University Press, Ft. Lauderdale, 1990.

Goel VK, Winterbottom JM, Schulte KR et al. Ligamentous laxity across the C0-C1-C2 complex: Axial torque-rotation characteristics until failure. Spine, 15:990-996, 1990.

Gudavalli MR, Triano JJ. An analytical model for the estimation of loads and strains in the posterior ligaments of the lumbar spine. Advances in Bioengineering, 231-234, 1990.

Harris JH Jr., Edeiken-Monroe B, Kopaniky DR. A practical classification of acute cervical spine injuries. Orth. Clin. No. Am. 17(1):15-30, 1986.

Hess JL, and Lombard CV. Theoretical investigations of dynamic response of man to high vertical accelerations. J Aviation Med 29:66, 1958.

Hodgson VR, Thomas LM. Mechanisms of cervical spine injury during impact to the protected head, SAE Transactions, Paper #801300, 17-42, 1980.

de Jager M, Sauren A, Thunnissen J, Wismans J A. Three-dimensional head-neck model: validation for frontal and lateral impacts. The 38th Stapp Car Crash Conference, 93-109, 1994.

Kabo JM, Goldsmith W. Response of a human head-neck model to transient sagittal plane loading. J Biomech 16(5):313-325, 1983.

Kalips I, von Gierke HE, Weis EB. A five degree of freedom mathematical model of the body. Symposium on Biodynamic Models and their Applications. Aerospace Medical Research Laboratory (AMRL-TR-71-29), WPAFB, Ohio, 1971, p. 211.

Keaveny TM, Hayes, WC. Mechancial properties of cortical and trabecular bone. Bone 7:285-344, 1993.

Kim YE, Goel VK, Weinstein JN, Lim TH. Effect of disc degeneration at one level on the adjacent level in axial mode. Spine 16(3),331-335, 1991.

Kleinberger M. Application of finite element techniques to the study of cervical spine mechanics. The 37th Stapp Car Crash Conference 261-272, 1993.

Lang W. Mechanical and physiological response of the human cervical vertebral column to severe impacts applied to the torso. A Symposium on Biodyanimc Models and Their Applications. Wright--Patterson AFB, Ohio, Aerospace Medical Research Laboratory 1971.

Latham F. A study of body ballistics: seat ejections. Proc. R. Soc. Lonson Ser., B147:121, 1957.

Li TF, Advani SH, Lee YC. The effect of initial curvature on the dynamic response of spine to axial acceleration. Symposium on Biodynamic Models and Their Application. AMRL-TR71-29, WPAFB, Dayton Ohio, 553, 1971.

Liang D, Winters JM. Mechanical response of an anthropomorphic head-neck system to external loading and muscle contraction. First World Congress of Biomechanics, II, 316, San Diego Ca., August 30-September 4, 1990.

Liu YK, Murray A. Theoretical study of the effect of impulse on the human torso. In Fung, YC (ed) Biomechanics. New York, ASME, 167-186, 1966.

Liu YK, Dai QG. The second stiffest axis of a beam column: implications for cervical spine trauma. Journal of Biomechanical Engineering, 111:122-127, 1989.

Lysell E. Motion in the cervical spine. Acta Orthop Scand, 23 (suppl.), 1969.

Maiman DJ, Sances A, Myklebust JB, et al. Compression injuries of the cervical spine; a biomechanical analysis. Neurosurgery 13:254-260, 1983.

McElhaney JH, Alem N, Roberts, V. A porous block model for cancellous bones. ASME #70-WA/BHF-2, 1-9, 1970.

McElhaney JH, Snyder RG, States JD, Gabrielsen MA. Biomechanical analysis of swimming pool injuries. SAE # 790137, 47-53, 1979.

McElhaney JH, Paver JG, McCrackin HJ, Maxwell GM. Cervical spine compression responses. The 27th Stapp Car Crash Conference, SAE, 831615:163-177, 1983.

McElhaney JH, Doherty BJ, Paver JG, Myers BS, Gray L. Combined bending and axial loading responses of the human cervical spine. The 32nd Stapp Car Crash Conference, 21-28, 1988.

McElhaney JH, Myers BS. Biomechanical aspects of cervical trauma in accidental injury: biomechanics and prevention. Nahum A., and Melvin, JW, Eds., Springer-Verlag, New York, 311-361, 1993.

McHenry RR, Naab KN. Computer simulation of the crash victim: a validation study. Proceedings of the 10th Stapp Car Crash Conference, SAE Paper #660792, 1967.

Melvin JW, McElhaney JH, Roberts VL. Improved neck simulation for anthropometric dummies. The 16th Stapp Car Crash Conference, 45-60, SAE # 720958, 1972.

Mertz HJ, Patrick LM. Investigation of the kinematics and kinetics of Whiplash. SAE #670919, 1967.

Mertz HJ, Patrick LM. Strength and response of the human neck. The 15th Stapp Car Crash Conference SAE Transactions, Vol. 80, #710855, 1971.

Mertz HJ, Hodgson VR, Thomas LM, Nyquist GW. An assessment of compressive neck loads under injury producing conditions. Physician & Sports Medicine, 6:11, November 1978.

Moffatt CA, Advani SH, Lin CJ. Analytical end experimental investigations of human spine flexure. ASME #71-WA/BHF-7, 1971.

Moroney SP, Schultz AB, Miller JAA, and Andersson GBJ. Load-displacement properties of lower cervical spine motion segments. J Biomech, 21 (9):769, 1988.
Myers BS, McElhaney JH, Doherty BJ, Paver JG, Nightingale RW, Ladd TP, Gray L. Responses of the human cervical spine to torsion. The 33rd Stapp Car Crash Conference, 215-222, 1989.
Myers BS, McElhaney JH, Richardson WJ, Nightingale R, Doherty BJ. The Influence of End Condition on Human Cervical Spine Injury Mechanisms. The 35th Stapp Car Crash Conference, 215-222, 1991.
Myers BS, McElhaney JH, Doherty BJ, Paver JG, Gray L. The role of torsion in cervical spinal injury. Spine 16(8):870-874, 1991.
Myers BS, McElhaney JH, Richardson WJ, Nightingale R. Cervical spine injury mechanisms, 107-156. In Head and Neck Injury, Levine RS, Ed., SAE International, Warrendale, PA, 1994.
Nightingale RW, McElhaney JH, Richardson WJ, Myers BS. Dynamic responses of the head and cervical spine to axial impact loading. J Biomechanics, In press, 1995.
Nightingale RW, McElhaney JH, Richardson WJ, Best TM, Myers BS. Experimental cervical spine injury: relating head motion, injury classification, and injury mechanism. JBJS, In press 1995.
Nusholtz GS, Huelke DE, Lux P, Alem NM, Montalvo F. Cervical spine injury mechanisms. The 27th Stapp Car Crash Conference, 179-197, 1983.
Orne D, Liu YK. A mathematical model of spinal response to impact. J Biomechanics, 4(1):49-71, 1971.
Panjabi MM. Three-dimensional mathematical model for the study of the mechanics of the human vertebral column. J Biomechanics, 6(6):671-680, 1973.
Panjabi MM, Summers DJ, Pelker RR, Videman T, Friedlaender GE, and Southwick WO. Three-dimensional load displacement curves of the cervical spine. J Orthop Research, 4:152, 1986.
Panjabi MM, Dvorak J, Duranceau J, Yamamoto I, Gerber M, Rauschning W, and Bueff HU. Three-dimensional movements of the upper cervical spine. Spine, 13 (7):726, 1988.
Panjabi MM, Oda, T, Crisco JJ III, Oxland TR, Katz L, Nolte L-P. experimental study of atlas injuries 1: biomechanical analysis of their mechanisms and fracture patterns. Spine, 16(10) S460-S465, 1991.
Penning L. Normal movements of the cervical spine. Am J Roentenol, 130:317, 1979.
Penning L. and Wilmink JT. Rotation of the cervical spine. Spine, 12 (8):732, 1987.
Pintar FA, Myklebust JB, Sances A Jr., Yoganandan N. Biomechanical properties of the human intervertebral disk in tension. Advances in Bioengineering: 38-39, 1986.
Pintar F, Sances A Jr., Yoganandan N, Reinartz J, Maiman DJ, Suh JK, Unger G, Cusick JF, Larson SJ. Biodynamics of the total human cervical spine. The 34nd Stapp Car Crash Conference, 55-72, 1990.
Portnoy HD, McElhaney JH, Melvin JW, et al. Mechanisms of cervical spine injury in auto accidents. In Proceedings, 15th Conference, Amer. Assoc. for Automotive Medicine, Ann Arbor, Michigan 1972, pp. 58-83.
Prasad P, King AI, Ewing CL. The role of artricular factors during $+G_z$ acceleration. J Appl Mech, 41:321, 1974.

Prasad P, King AI. An experimentally validated dynamic model of the spine. J Appl Mech, 41:546, 1974.
Reber JG, Goldsmith W. Analysis of Large Head-Neck Motions. J Biomechanics, 12:211-222, 1979.
Roaf R. A study of the mechanics of spinal injury. JBJS 42B:810-823, 1960.
Roaf R. International classification of spinal injuries. Paraplegia, 10:78, 1972.
Robbins DH, Bowman BM, Bennett RO. MVMA two-dimensional crash victim simulation. The 18th Stapp Car Crash Conference, 657-677, 1991.
Sances A Jr., et al. Head and Spine Injuries. AGARD Conference on Injury Mechanism, Prevention and Cost. Koln, Germany, 13-1 to 13-33, 1982.
Selecki BR, Williams HBL. Experimental study of mechanisms of injury. injuries to the cervical spine and cord in man. Australian Medical Publishing Co. Ltd., Australia, 1970.
Seireg A, Arvikar RJ. A comprehensive musculoskeletal model for the human vertebral column. Proc. ASME Biomechnaics Symposium, Houston TX, 74, 1975.
Shea M, Edwards WT, White AA, and Hayes WC. Variations of stiffness and strength along the human cervical spine. J. Biomechanics 24(2):95-107, 1991.
Shea M, Wittenberg RH, Edwards WT, White AA III, Hayes WC. In vitro hyperextension injuries in the human cadaveric cervical spine. JOR, 10(6):911-6, 1992.
Shirazi M. Response of the Spine in Biodynamic Environments. Symposium on Biodynamic Models and Their Applications, Aerospace Medical Research Laboratory (AMRL-TR-71-29), WPAFB, Ohio, 843, 1971.
Shirazi-Adl A, Ahmed AM, Shrivastave SC. Mechanical response of a lumbar motion segment in axial torsion alone and combination with compression. Spine 11(9):914-927, 1986.
Soechting JF, Paslay PR. A model for the human spine during impact including musculature influence. J.Biomech, 6, 195, 1973.
Svensson MY, Lovsund P. A dummy for rear-end collisions - development and validation of a new dummy-neck. IRCOBI, 299-310, 1992.
Terry CT, Roberts VL. A viscoelastic model of the human spine subjected to G_z accelerations. J Biomechanics 1:161-168, 1968.
Torg JS, (ed). Atheletic injuries to the head neck and face. Lea & Febiger, Philadelphia, 1982.
Triano JJ, Schultz AB. Cervical spine manipulations: applied loads, motions and myoelectric responses. Advances in Bioengineering, 249-250, 1990.
Ueno, Liu YK. A three-dimensional nonlinear finite element model of lumbar intervertebral joint in torsion. J Biomech Eng., 109:200-209, 1987.
Vulcan AP, King AI. Forces and moments sustained by lower vertebral column of seated human during seat-to-head acceleration. In Dynamic Response of Biomechanical Systems, ASME, New York, 1970.
Weber EH. Anatomisch-physiologische Untersuchung über einige Einrichtungen im Mechanismus der menschlichen Wirbelsäule. Arch Anat Physiol, 240, 1827.
Werne S. Studies in spontaneous atlas dislocation. Thesis. Acta Orthop Scand, Suppl. 23, 1957.

White AA, Panjabi MM. Clinical Biomechanics of the Spine, 2nd Edition, Lippincott Publishers, Philadelphia, PA, 1990.
Williams JL, Belytschenko TB. A three-dimensional model of human cervical spine for input simulation. J Biomech Eng 105:321-32, 1983.
Wismans J, Maltha J, van Wijk JJ, Janssen EG. MADYMO - A crash victim simulation program for biomechanical research and optimization of designs for impact injury prevention. Proceedings of the AGARD meeting April, 1982, Cologne, Germany.
Wismans JS, Spenny C. Performance requirements of mechanical necks in laberal flexion. The 27th Stapp Car Crash Conference, 137, 1983.
Wismans J, Philippens M, van Oorschot E, Kallieris D, Mattern R. Comparison of human volunteer and cadaver head-neck response in frontal flexion. Proceedings of the 31st Stapp Car Crash Conference, 1-11, 1987. SAE #872194.
Yoganandan N, Sances A Jr., Maiman DJ, Myklebust JB, Pech P, and Larson SJ. Experimental spinal injuries with vertical impact. Spine, 11(9):855-860, 1986.
Yoganandan N, Myklebust JB, Ray G, Sances A Jr. Mathematical and finite element analysis of spine injuries. CRC Critical Reviews in Biomedical Engineering 15(1):29-93, 1987.
Yoganandan, Sances A Jr., Pintar F. Biomechanical evaluation of the axial compressive responses of the human cadaveric and manikin necks. Journal of Biomechanical Engineering, 111:250--255, 1989.
Yoganandan N, Pintar FA, Sances A, Jr, Reinartz J, Larson SJ. Strength and kinematic response of dynamic cervical spine injuries. Spine, 16(10) Supp 1991, S511-17.
Zhu Y, Yang-He X, Zhu D-M, Sun C-Z. A lumped-parameter dynamic model of parachuter's head-neck system. First World Congress of Biomechanics, I, 161, San Diego CA, August 30 - Spetember 4, 1990.

Table 1. CERVICAL SPINE RANGES OF MOTION (degree)

FLEXION		C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
#7	Mean			6.3	6.3	6.3	6.3	6.3	
Invitro	SD			1.2	1.2	1.2	1.2	1.2	
#8	Mean	3.5	11.5						
Invitro	SD	1.9	6.3						
#9	Mean			5.6	5.6	5.6	5.6	5.6	5.6
Invitro	SD			1.8	1.8	1.8	1.8	1.8	1.8

EXTEN.		C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
#7	Mean			3.6	3.6	3.6	3.6	3.6	
Invitro	SD			1.2	1.2	1.2	1.2	1.2	1.2
#8	Mean	21.0	10.9						
Invitro	SD	6.0	3.5						

#9	Mean			3.5	3.5	3.5	3.5	3.5	3.5
Invitro	SD			1.9	1.9	1.9	1.9	1.9	1.9

FLEXION + EXTENSION		C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
#1	Mean	30.0	30.0	12.0	18.0	20.0	20.0	15.0	
Invivo	SD	5.0	5.0	2.8	3.3	3.5	3.3	4.8	
#4a	Mean		12.0	10.0	15.0	19.0	20.0	19.0	
Invivo	SD		4.0	3.0	3.0	4.0	4.0	3.0	
#4b	Mean		15.0	12.0	17.0	21.0	23.0	21.0	
Invivo	SD		3.0	2.0	5.0	3.0	4.0	4.0	
#5a	Mean		14.3	12.3	17.5	21.4	22.7	21.4	
Invivo	SD		4.9	2.9	3.9	3.2	4.2	3.6	
#5b	Mean		13.0	12.3	18.1	22.0	23.3	22.0	
Invivo	SD		3.2	2.9	4.9	3.9	4.2	3.6	
#6	Mean			4.9	10.2	13.0	14.5	13.5	8.0
Invitro	SD			2.2	2.7	3.5	3.2	4.4	2.6
#7	Mean			9.9	9.9	9.9	9.9	9.9	
Invitro	SD			1.2	1.2	1.2	1.2	1.2	
#8	Mean	24.5	22.4						
Invitro	SD	4.0	4.7						
#9	Mean			9.1	9.1	9.1	9.1	9.1	9.1
Invitro	SD			1.8	1.8	1.8	1.8	1.8	1.8
#10	Mean	25.0	20.0	10.0	15.0	20.0	20.0	17.0	9.0
	SD			2.8	4.8	4.0	4.0	5.0	0.8

LAT. BEND. (ONE SIDE)		C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
#6	Mean			4.0	4.9	4.6	4.5	4.2	3.2
Invitro	SD			2.9	3.4	2.9	2.7	3.4	2.2
#8	Mean	5.5	6.7						
Invitro	SD	2.5	4.4						
#9	Mean			4.7	4.7	4.7	4.7	4.7	4.7
Invitro	SD			3.0	3.0	3.0	3.0	3.0	3.0
#10	Mean	5.0	5.0	10.0	11.0	11.0	8.0	7.0	4.0
	SD			2.3	1.5	4.0	4.0	4.3	4.3

AXIAL ROT. (ONE SIDE)		C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
#2	Mean	4.0	43.1	28.5 (motion between C2 and T1)					
Invivo	SD	1.6	5.5	7.9					
#3	Mean	1.0	40.5	3.0	6.5	6.8	6.9	5.4	2.1
Invivo	SD	1.8	4.3	2.5	1.8	2.8	2.5	2.0	2.3
#6	Mean			3.0	4.9	5.2	4.0	2.9	2.4
Invitro	SD			3.4	4.9	4.1	3.9	4.3	1.9
#8	Mean	7.3	38.9						
Invitro	SD	2.2	5.4						
#9	Mean			1.9	1.9	1.9	1.9	1.9	1.9
Invitro	SD			0.7	0.7	0.7	0.7	0.7	0.7
#10	Mean	5.0	40.0	3.0	7.0	7.0	7.0	6.0	2.0
	SD			2.5	1.8	2.8	2.5	2.0	1.8

REFERERNCES

#1 Penning, 1978 AJRoentlg 130:317
#2 Dvorak, 1987 SPINE 12/8:726
#3 Penning, 1987 SPINE12/8:732
#4 Dvorak, 1988 SPINE 13/7:748
#5 Dvorak, 1991 JOR 9:828-834
#6 Lysell, 1969 Acta Orth Scan Supp.123
#7 Panjabi, 1986 JOR 4:152-161
#8 Panjabil, 1988 SPINE 13/7:728
#9 Moroney,1988 J Biom 21(9):769
#10 White & Panjabi, 1990 Clinical Biomechanics of the Spine
#4a Active motion, Ref #4
#4b Passive motion, Ref #4
#5a Males, passive motions, Ref # 5
#5b Females, passive motion, Ref #5

TABLE 2. RECOMMENDED RANGES OF MOTIONS (DEG.)

		C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
Flex + Extn	Mean	24.5	22.4	12.0	17.0	21.0	23.0	21.0	8.0
	SD	4.0	4.7	2.0	5.0	3.0	4.0	4.0	2.6

Lat Bend	Mean	5.5	6.7	4.0	4.9	4.6	4.5	4.2	3.2
	SD	2.5	4.4	2.9	3.4	2.9	2.7	3.4	2.2
Axial Rot	Mean	7.3	38.9	3.0	6.5	6.8	6.9	5.4	2.1
	SD	2.2	5.4	2.5	1.8	2.8	2.5	2.0	2.3

TABLE 3. LOWER AND UPPER LIMITS OF MOTIONS

Flex + Extn	Lower	16.5	13.0	8.0	7.0	15.0	15.0	13.0	2.8
	Upper	32.5	31.8	16.0	27.0	27.0	31.0	29.0	13.2
Lat Bend	Lower	0.5	-2.1	-1.9	-1.9	-1.3	-0.9	-2.6	-1.3
	Upper	10.5	15.5	9.8	11.7	10.4	9.9	11.0	7.6
Axial Rot	Lower	2.9	28.1	-2.0	3.0	1.3	1.9	1.4	-2.4
	Upper	11.7	49.7	8.0	10.0	12.3	11.9	9.4	6.6