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Biomechanics of the Neck

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1. Introduction

From a mechanical and structural point of view, the human neck is a very complex mechanism, containing vital neurologic, vascular, and respiratory structures as well as the cervical vertebrae and spinal cord. The incidence of neck injuries in traffic accidents appears to be relatively low compared to for instance head injuries, except for specific accident configurations like a rear end collision where more than 50% of the injuries appear to be in the neck area. However, the vehicle injury priority rating data indicated that neck injuries became the fifth most important injury category (after head, face, chest and abdomen). Knowledge on the mechanism causing neck injuries is still rather limited. Therefore, various biomechanical models (i.e. animal/human models and numerical models), injury mechanisms and injury tolerance of the neck will be presented in this chapter.

2. Neck biomechanics models

2.1 Animal models

2.1.1 Goat in vivo model

An in vivo goat model (Cavanaugh et al., 2006; Chen et al., 2005; Lu et al., 2005a, 2005b) was developed to investigate the injury threshold of cervical facet joint capsules (FJC) in vivo. The method incorporated a custom-fabricated testing frame for facet joint loading, a stereoimaging system, and a template-matching technique to obtain single afferent response. The C5 articular process was then pulled via a computer-controlled actuator at a rate of 0.5 mm/sec to simultaneously stretch the C5-6 capsule, record sensory nerve activity due to stretch and record strain by tracking targets on the capsule. In these studies Lu et al (2005b) demonstrated a quantitative relationship between capsule sensory discharges and applied capsule stretch from cervical facet joints. Neural responses of all mechanosensitive units showed statistically significant correlations (all $P<<0.05$) with both capsular load ($r^2 = 0.744 \pm 0.109$) and local strain ($r^2 = 0.868 \pm 0.088$). Most of the capsular neural receptors responded in the physiologic range of capsule stretch and fired at strains of $(10.2 \pm 4.6)\%$

that typically do not signal pain. However, higher capsular strains of (47.2 ± 9.6)% triggered discharges from higher threshold receptors which were most likely from nociceptors. Nociceptors transmit signals to the central nervous system to signal pain. Afterdischarges were reported in these goat studies after capsular strains of (45.0 ± 15.1)% and may be related to tissue injury and release of inflammatory mediators into the surrounding tissue. These changes may lead to central sensitization of pain pathways in the spinal cord which may lead to persistent or chronic whiplash pain.

2.1.2 Rat in vivo model

A rat model was used by Quinn et al (2007) who conducted a study to quantify the structural mechanics of the cervical facet capsule and define the threshold for altered structural responses in this ligament during distraction. Tensile failure tests were performed using isolated C6/C7 rat facet capsular ligaments (n = 8); gross ligament failure, the occurrence of minor ruptures and ligament yield were measured. Gross failure occurred at (2.45 ± 0.60)N and (0.92 ± 0.17)mm. However, the yield point occurred at (1.68 ± 0.56)N and (0.57 ± 0.08)mm, which was significantly less than gross failure ($P < 0.001$ for both measurements). Maximum principal strain in the capsule at yield was (80 ± 24)%. Energy to yield was (14.3 ± 3.4)% of the total energy for a complete tear of the ligament. Ligament yield point occurred at a distraction magnitude in which pain symptoms begin to appear in vivo in the rat.

Findings presented here suggest a relationship between structural damage of the facet capsular ligament and potential mechanisms of pain for subfailure distraction. Quinn et al (2007)'s data show ligament yield at a significantly lower distraction than gross failure. While these subfailure distractions may not produce visible ligament tears, detection of the ligament's altered structural response may provide an indication of an injury sufficient to elicit sustained nociceptor firing, pain symptoms, and persistent activity in the nervous system. Given the evidence that painful joint distractions begin near ligament yield, this study may suggest that the physiologic range of the facet joint is actually limited to prior to yield. This mechanical study provides a framework for future in vivo studies in determining a mechanical threshold for persistent pain, and also provides data for quantitative scaling to other animal models and to the human. These findings provide mechanical definition of altered ligament behavior corresponding with a loading condition known to produce pain, linking mechanical damage and persistent pain for the first time.

2.2 Human models

2.2.1 Human cadaveric models

A bench-top trauma sled was used to apply four intensities of whiplash trauma to human cadaveric cervical spine specimens and to measure resultant intervertebral rotations using high-speed cinematography (Grauer & Panjabi, 1997). Objectives were to determine the cervical spine levels most prone to injury from whiplash trauma and to hypothesize a mechanism for such injury. Six occiput to T1 (or C7) fresh cadaveric human spines were studied. Physiologic flexion and extension motions were recorded with a motion analysis system by loading up to 1.0 N m. Specimens then were secured in a trauma sled, and a surrogate head was attached. Flags were fixed to the head, and individual vertebrae were monitored with high-speed cinematography (500 frames/sec). Data were collected for 12 traumas in four classes defined by the maximum sled acceleration. The trauma classes were 2.5 G, 4.5 G, 6.5 G, and 8.5 G (G: acceleration of gravity). In the whiplash traumas, the peak intervertebral rotations of C6-C7 and C7-T1 significantly exceeded the maximum physiologic extension of all trauma classes studied. The maximum extension of these lower levels occurred significantly before full neck extension. The upper cervical levels were consistently in flexion at the time of maximum lower level extension. It was concluded that in whiplash, the neck forms an S-shaped curvature, with lower level hyperextension and upper level flexion. A subsequent C-shaped curvature with extension of the entire cervical spine produced less lower level extension.

2.2.2 Volunteer models

The volunteers sat on a seat mounted on a sled that simulated actual car impact acceleration (Ono et al., 1997). Impact speeds (4, 6, and 8 km/h), seat stiffness, neck muscle tension, and alignment of the cervical spine were selected for a parametric study of the head-neck-torso kinematics and cervical spine responses. The effects of these parameters were studied without the headrest. Muscle activity was measured with surface electromyography. The cervical vertebrae motions were recorded by cineradiography (90 frames/sec radiograph) and analyzed to quantify the rotations and translations of cervical vertebrae during impact. Furthermore, the motion patterns of cervical vertebrae during impact were compared with the normal motions. A subject's muscles in the relaxed state did not affect the head-neck-torso kinematics on rear-end impact. The ramping-up motion of the subject's torso was observed, caused by the inclination of the seat back. An axial compression force occurred. The lower cervical vertebral segments extended and rotated before the motions of the upper segments. These motions were beyond the normal physiologic cervical motions, which could trigger the facet joint injury mechanism. In addition, the more rigid the seat cushion, the greater the axial compression force. In contrast, the torso rebounding caused by the softer seat cushion tended to intensify the shearing force applied to the upper vertebrae. It was also deduced that the initial posture of the cervical spine affected the impact responses of head and neck markedly. Based on the differences in the alignment of the cervical spine between male and female occupants, it was deduced that the female neck injury incidence may be higher than that of the male, because the female cervical spine takes the kyphosis position more often than does the male cervical spine.

2.3 Numerical models

2.3.1 Two-joint neck models

Two-joint neck models were proposed in the 1970's and 1980's (Bowman & Robbins, 1972; Seemann et al., 1984; Bosio and Bowman, 1986; Wismans et al., 1987). These are relatively simple models in which the head, neck and torso were linked by two pivots or ball-socket joints, depending on whether the model was 2-D or 3-D. The upper joint was normally located near the occipital condyles, and the lower joint was usually situated near the first thoracic (T1) vertebra. Although fairly simple, these models contributed greatly to the analysis of the global kinematics of the head-neck complex during frontal, lateral and oblique impacts, over the past 30 years. For example, Wismans et al. (1987) used this type of model to simulate the head-neck motion. In these studies, the neck was modeled as a rigid body, and the geometrical and mechanical properties of the joints were determined based on those test data. Performance requirements for future dummy neck designs were also defined by means of this model. Similar models have been proposed as well by Bosio & Bowman (1986), except an extensible neck was used instead of a rigid one. In these two studies, model validations were performed in the –Gx and +Gy directions. A nonlinear joint stiffness and a neck elongation stiffness were found to be necessary to obtain satisfactory simulation results. Very similar models were also used to perform a parametric study by Bowman & Robbins (1972) to investigate the injury mechanism of the human neck in frontal impact, as well as to test another theory by Seemann et al. (1984), who proposed that some volunteers "locked" their neck joints in anticipation of sled firing. Because two-joint models were originally developed for describing the global motion of the head and neck relative to the torso, they were not adequate to describe vertebral kinematics and cervical soft tissue deformation. It is apparent that more detailed models would be needed.

2.3.2 Multi-body (MB) neck models

MB neck models were the second category of numerical models of the human cervical spine. In these models, the head and vertebrae were modeled as rigid bodies, while the soft tissues were modeled as either nonlinear viscoelastic intervertebral joints or by a detailed arrangement of spring-damper elements to represent the intervertebral discs, ligaments, facet joints, and muscles. For instance, De Jager et al. (1994) developed a MADYMO-based 3-D MB model of the head-neck based on the model of Deng & Goldsmith (1987). In this model, the rigid head and vertebrae were connected by linear viscoelastic intervertebral joints and nonlinear elastic muscle elements. A sensitivity analysis was also performed in this study, and it was found that head mass, joint stiffness and damping coefficients for rotational deformations had a major influence on model response. Subsequently, De Jager et al. (1996) developed a more detailed version of this model, in which linear viscoelastic discs, nonlinear viscoelastic ligaments, frictionless facet joints and contractile muscles were simulated. Model responses to 15-G frontal and 7-G lateral impacts were validated against volunteer data. This model was then modified by Van der Horst et al. (1997) who included more muscles, and divided the muscles into segments to simulate muscle curvature. Volunteer frontal impact data at reduced G-levels were used for model validation.

2.3.3 FE neck models

In the past 20 years, several FE models of the human cervical spine were developed by Camacho et al (Kleinberger, 1993; Dauvilliers et al., 1994; Camacho et al., 1997; Kumaresan et al., 1997; Yang et al., 1998; Halldin et al., 2000; Chancey et al., 2003; Meyer et al., 2004). Because FE models allowed for a more detailed and realistic representation of the neck geometry and its material behavior, more detailed neck injury mechanisms could be studied using this kind of model.

The first FE neck model presented at The STAPP Car Crash Conference was made by Kleinberger (1993). In this model, the geometries of the vertebrae and the skull were highly simplified, and the atlanto-occipital articulation was represented by a pin joint. Only the model responses to axial stiffness and neck versus head angle curve during an 8-G frontal sled test were compared with experimental data. No detailed validations were conducted at that time.

Dauvilliers et al. (1994) proposed another FE neck model, which was also fairly simple by current standards. In this model, each vertebra consisted of only 12 solid elements, and spring-damper elements were used to simulate the major ligaments. The model responses were compared to volunteer kinematics during frontal and lateral impacts. Head accelerations responses of the model matched the experimental data fairly well; However, the head displacement and angle data were not close to the reference data. One limitation of this model was that muscles which could have a significant influence on model response during frontal and lateral impacts were not simulated.

Kumaresan et al. (1997) presented a set of detailed FE models of a one-year old, three-year old, and six year old pediatric human cervical spine from C4 to C6 vertebrae. Although their study only focused on the pediatric neck, the development and validation of the adult FE model were also provided. Compared to other neck models, this model had complicated geometries and consisted of several very detailed anatomical structures, such as the endplate, uncovertebral joint and synovial fluid. The adult model flexibilities under quasi-static loading conditions were validated against experimental data. However, this model only simulated 3 vertebrae; and thus head-neck response during impact could not be simulated. For modeling the pediatric neck, it has been found that the inclusion of the local geometry and material property changes produced higher changes in the flexibilities than pure structural scaling. This finding emphasized the need to consider the local geometry and material property changes to better predict the biomechanical responses of the pediatric human cervical spine.

Camacho et al. (1997) proposed an FE head-neck model to investigate the dynamic responses of the head and neck to near-vertex head impact. In this model, highly simplified rigid vertebrae were connected by nonlinear spring elements, the stiffness of which was measured from quasi-static flexion-extension tests. To simulate head impact, a deformable head was defined and linked to the C1 vertebra. Strictly speaking, their model should be classified as a MB neck model coupled with an FE head model. The results showed that this model could accurately predict resultant neck forces, resultant head forces and accelerations measured in cadaveric near-vertex head impact tests with impact surfaces at -15 deg, 0 deg and +15 deg with respect to the horizontal. Although the model produced higher neck forces than the validation corridors during the latter portions of the simulations, the authors explained that this was because at that time some of the cadaveric specimens had suffered cervical spine injuries, and thus possessed decreased load carrying capacity. This model was also used to estimate human neck tensile tolerance (Chancey et al., 2003), but no further validations were conducted.

Yang et al. (1998) developed a detailed FE neck model the geometry of which was taken from an MRI. It incorporated a previously developed head and brain model, so that head-neck response could be simulated during impact. All the bony structures, articular surfaces, relevant ligaments and intervertebral discs were described. It should be noted that it was the first model that included the ligaments in the upper vertebrae (Occiput-C2). Near-vertex head impact test data with impact surface at 0° deg to the horizontal and cadaveric rear impact sled test data were used to validate the model. Model responses matched the experimental data fairly well. However, for the rear impact, only head kinematics from the model and test were compared. Although model results of facet capsule stretch percentage were presented, such results have not been validated as yet due to the limited experimental data available. The validated model has also been integrated into a skeleton torso model to study the neck response during head-torso-airbag interaction.

Another FE head-neck complex model was developed by Halldin et al. (2000), who attempted to use the model to evaluate a new car roof design concept. This was also a fairly detailed model, slightly smaller in size than the model developed by Yang et al. (1998). Component compression tests and near-vertex head impact tests with impact surface at

-15 deg, 0 deg, and +15 deg to the horizontal were used to validate the model. Most of the model responses matched the experimental data fairly well.

Meyer et al. (2004) presented another detailed FE head-neck complex model. Because the aim of their study was not to reproduce bony fracture, but to simulate more moderate lesions, the vertebrae were modeled by rigid shell elements. The major contribution of this model was the model validation procedure. The model responses were validated against volunteer and/or cadaveric impacts in frontal, lateral, oblique, and rear directions. Moreover, frequency domain information was also taken into account to improve model responses. To the best of our knowledge, this was the first time researchers validated the FE neck model in both the time and frequency domains.

3. Neck injury mechanisms

Because the neck is a slender column that can be subjected to a variety of bending loads in association with an axial load, the injury modes can be classified as compression, tension-extension, tension-flexion, compression-extension, compression-flexion, and lateral bending.

3.1 Compression injuries

These injuries result from crown impacts to the head which produce a high compressive load on the neck accompanied by bending loads that can depend on the initial orientation of the head, initial configuration of the neck, and surface friction. They are not common automotive injuries but can occur in ejections and rollovers. The compression comes from the mass of the body following the head, which is stopped by a resisting surface.

In order to understand the injury mechanisms of this kind compression loading, this chapter summarizes a numerical investigation of factors affecting cervical spine injuries during rollover crashes by Hu et al (2008). In fact, rollover crashes are the most hazardous vehicular crashes to the human cervical spine. The incidence rate of AIS 3+ (AIS: Abbreviated Injury Scale) cervical spine injuries in rollovers was nearly 4 times that occurring in frontal and side crashes (Yoganandan et al., 1989). The cervical spine was also the third most commonly injured body region during rollovers after the head and thorax (Hu et al, 2005), and injuries to the cervical spine may often lead to permanent disabilities.

Previous experimental (Bahling et al., 1990) and numerical (Hu, 2007) studies have shown that cervical spine injuries in rollovers were mainly caused by the inertia of the occupant's torso compressing the head into the roof/ground, which is often referred to as "the diving mechanism." In the past 3 decades, several cadaveric experiments were conducted to investigate cervical spine injuries under diving type of impacts. The injury mechanism under sagittal plane loading (combined compression and flexion/extension) is reasonably well understood, and many important and widely acknowledged conclusions have been drawn, including: (1) neck injury mechanisms are dependent on the neck orientation and impact direction, (2) padding will add constraints to the head which, in turn, will increase the risk of neck injury, and (3) if the neck can escape from the direction of impact momentum, the risk of neck injury will be greatly reduced.

However, during rollovers, the centrifugal force tends to maintain a belted occupant erect with his/her head upward and outboard (Bahling et al., 1990), thus it is common that an impact to the occupant's head would come from the upper and lateral sides of the vehicle. Therefore, a combined lateral bending, compression, and flexion/extension could be a very common neck loading mode during rollovers. Because of the complex nature of rollover crashes, there is no typical rollover scenario that can be specified easily (Hu et al, 2007). To investigate the neck injury mechanism under this complex combined loading condition, multiple experiments under different loading directions have to be conducted to mimic different rollover scenarios. Because of the difficulty in securing test samples and the high cost associated with performing cadaveric experiments, this research strategy needs to be modified. With recent advancement in computing technology and software, numerical modeling could provide a cost-effective way to perform this type of research. Recently, several published finite element (FE) human head-neck models (Camacho et al., 1997; Halldin et al., 2000; Zhang et al., 2005) have been validated under diving type of impact conditions, but until now none has simulated the combined lateral bending and loading in the sagittal plane–a loading condition similar to real rollover scenarios. Furthermore, although many risk factors have been identified in previous studies to be associated with cervical spine injuries, no study has systematically analyzed those factors. Therefore, Hu et al. (Hu et al., 2008) carried out a study to investigate neck responses under various complex loading conditions similar to real-world rollover scenarios using a detailed FE head-neck model. The effects of changing the coefficient of friction (COF), impact velocity, padding material thickness and stiffness, and muscle force on the risk of neck injuries were analyzed in 16 different impact orientations based on a Taguchi array of design of experiments.

In summary, the following primary simulation results were found: (1) Impact velocity is the most important factor in determining the risk of cervical spine fracture (P = 0.000). (2) Decreases in the COF between the head and impact surface can effectively reduce the risk of cervical spine fracture (P = 0.038). (3) If the COF is not 0, an impact with lateral force component could sometimes increase the risk of cervical spine fracture; and the larger the angle of the impact surface, the more important it becomes to reduce the COF to protect the neck. (4) Soft (P = 0.033) and thick (P = 0.137) padding can actually decrease the neck fracture risk, which is in contrast to previous experimental data. Furthermore, Hu et al. (2008) concluded that a careful selection of proper padding stiffness and thickness, along with a minimized COF between the head and impact surface or between the padding and its supporting structure, may simultaneously decrease the risk of head and neck injuries during rollover crashes.

3.2 Tension-extension injuries

Tension-extension loading is common and is responsible for a group of injuries including whiplash, hangman's fractures, and structural injury to the anterior column of the spine. Tension-extension loading can occur in three primary ways:

- Fixation of the head with continued forward displacement of the body. This occurs commonly in unbelted occupants hitting the windshield and as a result of falls and dives.
- Inertial loading of the neck following an abrupt forward acceleration of the torso, as would occur in a rear-end collision (whiplash mechanism (Chen et al., 2009)).
- Forceful loading below the chin directed postersuperiorly (as in a judicial hanging).

Herein, this chapter summarizes research aimed at providing some kinematics responses and injury mechanisms of the neck in the rear-end collision.

3.2.1 Neck kinematics responses in the rear-end collision

Yang and King (2003) carried out a study to document the kinematics of the neck during low-speed rear-end impacts. In a series of experiments reported by Deng et al. (2000), a pneumatically driven mini-sled was used to study cervical spine motion using six cadavers instrumented with metallic markers at each cervical level, a 9-accelerometer mount on the head, and a tri-axial accelerometers on the thorax. A 250-Hz x-ray system was used to record marker motion while acceleration data were digitized at 10,000 Hz. Results show that, in the global coordinate system, the head and all cervical vertebrae were primarily in extension during the entire period of x-ray data collection. In local coordinate systems, upper cervical segments were initially in relative flexion while lower segments were in extension. Facet joint capsular stretch ranged from 17% to 97%. In the vertical direction, the head and T1 accelerated upward almost instantaneously after impact initiation while there was delay for the head in the horizontal direction. This combination was the result of a force vector which was pointed in the forward and upward direction to generate an extension moment. Upward ramping of the torso was larger in tests with a 20-deg seatback angle. The study concluded that the kinematics of the neck is rather complicated and greatly influenced by the large rotations of the thoracic spine. Significant posterior shear deformation was found, as evidenced by the large facet capsular stretch. Although the neck forms a "mild" S-shaped curve during whiplash, using its shape as an injury mechanism can be misleading because the source of pain is likely to be located in the facet capsules.

3.2.2 Axial compression hypothesis

In volunteer tests, McConnell et al (1993) found that a vertical acceleration can be measured during a low speed rear-end impact. This ramping up phenomenon was due to the straightening of the spine or the mechanical interaction between the seatback and the torso. This same phenomenon was also reported in a high-speed X-rays study of the neck for volunteers subjected to rear impact forces (Matsushita et al, 1994) and in Hybrid II dummy tests by Viano (1992). However, the measured vertical acceleration and movement were rather small and McConnell et al later reported it to be insignificant compared to that measured horizontally (McConnell et al, 1995).

Although the vertical acceleration may seem small, it plays a significant role in the cervical spine biomechanics. The head generally possesses about 4.5 kg of inertial mass. Even a small acceleration could generate a significant compressive force at the neck. In a rear-end impact, the car seat pushes (shears) the torso forward while the neck is subjected to this axial compression. Based on this observation, Yang and Begeman (1996) proposed a new hypothesis to explain the rear-end neck injury mechanism stating that axial compression can cause loosening of ligaments and make it easier for the cervical facet joint capsule and other tissues to be injured. Because these injuries occur in soft tissues, this new theory explains why there is generally no objective evidence.

The facet joint geometry of the cervical spine also plays an important role. In frontal impacts, the upper vertebra will shear anteriorly, relative to the lower vertebra. By observing the anatomy of the facet joints, it is evident that contact of the facet joints can protect against excessive frontal shear. However, in a rear-end impact, the lower vertebra shears anteriorly, the facet joint offers no protection to such a motion. This can be the reason that the rate of neck injury is much lower in frontal impacts of the same or even higher severity.

To test this hypothesis, an in vitro experiment was designed to investigate the theory that axial compression reduces the shear stiffness when the cervical spine is moved due to a rear-end impact (Yang et al., 1997).

Cervical spine specimens from C1-T1 were dissected from the entire spine. The C1 vertebra was fixed to an aluminum plate with screws. The other end (T1) was potted in epoxy and attached to a six-axis load cell. Two LED markers were attached to each vertebral body from C2-C7. One additional LED maker was attached to the frame of the Instron as reference. During the test, the actuator moves upward to simulate the seat back pushing from behind.
Five tests were done for each specimen. In the first test, the T1 was moved anteriorly to stimulate a rear-end impact for 20 mm displacement at a quasi-static speed of 0.04 mm/s. In the next four tests, an axial compression of 44.5 N, 89.0 N, 133.5 N and 178 N of dead weight were applied through a cable-pulley system. The same procedure as in the first test was then reported. Shear stiffness values were calculated from the load cell and motion data.
Result showed that shear stiffness values were reduced significantly with increased axial compression. Based on the total shear force vs the shear deflection data for a typical C5-C6 motion segment, it can be clearly seen that the shear stiffness decreased as the applied axial compression increased. The shear force vs deflection curves were nonlinear due to coupling rotations of vertebrae. The shear stiffness, defined as the final linear portion of the force-deflection curve, was reduced significantly with increased axial compression. For example, for the C2-C3 portion of the specimen No. 715, shear stiffness was only 50% of that without axial preload (Table 1).

	Stiffness (N/mm)			
	C2-C3	C3-C4	C4-C5	C5-C6
No preload	14.9	9.0	10.9	18.6
178 N preload	7.5	4.5	6.3	5.0

Table 1. Shear stiffness values calculated at each vertebra level (Specimen No. 715)

It should be noted that in previous typical static tests, the shear stiffness is expected to increase as the axial compression increase. Yang (1997)'s experimental data show the opposite trend. This explain why the neck injury rate is higher in a rear-end impact than that of a frontal impact. The axial compression presented in rear-end impact reduce the shear stiffness of the cervical spine and make it easier to be injured. Dynamic tests can give researchers more insight into the neck injury mechanism. Those data can be useful in the design of new equipment such as head restraints to protect the neck from rear-end impact injury.

3.2.3 Axial pretorque hypothesis

Whiplash victims who had their head turned at impact have more severe and persistent symptoms than patients who were facing forward (Sturzenegger et al., 1994; Sturzenegger et al., 1995). These findings have prompted biomechanical studies using human cadaveric necks to investigate why a head-turned posture increases injury potential. Dynamic rear-impact tests of prerotated ligamentous spines (occiput-T1) produce increased neck flexibility (interpreted as injury) in extension, lateral bending and axial rotation (Panjabi et al., 2006). Though concentrated in the lower cervical spine, these "injuries" were not isolated to particular spinal ligaments. Detailed measurements of the strain field in the facet capsule have also shown that a head-turned posture generates higher capsular strains than a neutral head posture, but the quasi-static loads applied during those tests were limited to pure flexion/extension moments and did not include the axial compression or posterior shear present during whiplash loading (Winkelstein et al, 2000). Thus the question of how a head-turned posture combined with multiaxial whiplash loads affects facet capsular ligament strain has yet to be answered.

For this reason, Siegmund et al. (2008) used human cadaveric motion segments to: (1) quantify the intervertebral kinematics and facet capsule strains under whiplash-like loads in the presence of an initial axial rotation, and (2) compare the capsule strains generated by these combined loads to the previously-published strains needed to injure these ligaments in isolated shear failure (Siegmund et al, 2001). Their overall hypothesis was that capsular strains during this simulated whiplash exposure are similar to those needed to injure the capsular ligament (Siegmund et al, 2008).

According to Siegmund et al. (2008), thirteen motion segments were used from 7 women donors (50 ± 10 years). Axial pretorques (±1.5 N m), axial compressive preloads (45, 197, and 325 N), and quasi-static shear loads (posteriorly-directed horizontal forces from 0 to 135 N) were applied to the superior vertebral body to simulate whiplash kinematics with the head turned. Three-dimensional displacements of markers placed on the right facet capsular ligament were used to estimate the strain field in the ligament during loading. The effects of pretorque direction, compression, and posterior shear on motion segment motion and maximum principal strain in the capsule were examined using repeated-measures analyses of variance.

Results showed that axial pretorque affected peak capsule strains more than axial compression or posterior shear. Peak strains reached 34%±18% and were higher for pretorques toward rather than away from the facet capsule (i.e., head rotation to the right caused higher strain in the right facet capsule).

Similarly, based on a validated intact head to first thoracic vertebra (T1) computational model, parametric analysis was used to assess effects of increasing axial head rotation between 0° and 60° and increasing impact severity between 8 and 24 km/h on facet joint capsule strains (Storvik et al., 2011). Rear impacts were simulated by horizontally accelerating the T1 vertebra. Characteristics of the acceleration pulse were based on the horizontal T1 acceleration pulse from a series of simulated rear impact experiments using full-body post mortem human subjects. Joint capsule strain magnitudes were greatest in ipsilateral facet joints for all simulations incorporating axial head rotation (i.e., head rotation to the left caused higher ligament strain at the left facet joint capsule). Strain magnitudes increased by 47%–196% in simulations with 60° head rotation compared to forward facing simulations. These findings indicate that axial head rotation prior to rear impact increases the risk of facet joint injury.

3.2.4 Facet joint impingement hypothesis

Ono et al. (1997) and Yoganandan et al. (1998) both proposed a facet joint impingement hypothesis. Specifically, Ono et al (1997) theorized that the facet synovium or a portion of the facet capsule could be trapped between the facet joint surfaces and pinched, causing pain. However, there is no biomechanical evidence that the capsule is loose enough to be trapped between the facet joint and even if it was trapped, evidence is lacking to show that nociceptors are present in the synovium or the trapped portion of the capsule that is indeed set off by the pressure. Kaneoka et al (1999) hypothesized that the center of rotation moved superiorly during a whiplash and caused the tip of the inferior facet (of the upper vertebra) to impact the superior facet surface (of the lower vertebra). This proposition that compression of the facet surfaces can produce pain is probably untenable since cartilage is devoid of nociceptors and there is no neurophysiological evidence that the nociceptors in the subchondral bone can be made to fire by this presumed compression.

3.2.5 Transient pressure gradient hypothesis

Aldman (1986) proposed a neck injury hypothesis for rear impact which states that injuries to the nerve root region in the cervical spine are a result of transient pressure gradients in the spinal canal during rapid neck extension (Svensson et al., 2000). In experimental neck trauma research on animals, pressure gradients were observed and indications of nerve cell membrane dysfunction were found in the dorsal root ganglia (DRG) of the cervical spine ganglia. The experiments covered neck extension, flexion and lateral bending. A theoretical model in which fluid flow was predicted to cause the transient pressure gradients was developed and a neck injury criterion based on Navier-Stokes Equations was applied on the flow model. The theory behind the Neck Injury Criterion indicates that the neck injury occurs early on in the rearward motion of the head relative to the torso in a rear-end collision. Thus the relative horizontal acceleration and velocity between the head and the torso should be restricted during the early head-neck motion to avoid neck injury. The flaw in this theory is that the observed transient pressure should affect all DRG's in the neck but whiplash pain is generally limited to the lower cervical spine. Furthermore, injury to nerve roots leads to radiculopathy (pain in the extremities) and not pain in the neck.

3.3 Tension-flexion injuries

These are relatively uncommon because complaints of chronic or persistent neck pain by belted occupants involved in frontal crashes are rare. In very severe frontal crashes, atlanto-occipital and C2/C3 separation can occur. Thomas & Jessop (1983) produced these injuries in subhuman primates that were fully restrained and were subjected to a frontal deceleration of 120 G.

3.4 Compression-extension injuries

These injuries can occur to unrestrained front seat occupants involved in a frontal crash. When the head impacts the windshield, the neck is placed into extension and compression simultaneously. Such occupants are likely to sustain fracture of one or more spinous processes as well as symmetrical lesions of the pedicles, facets, and laminae. If there is a fracture-dislocation, the inferior facet is displaced posteriorly and upward and appears to be more horizontal than normal on X-ray (Viano & King, 1997).

3.5 Compression-flexion injuries

Wedge compression fractures of the vertebral bodies are the results of a combination of a flexion bending moment and compressive loading of the vertebral motion segment resulting in greater compressive stresses and failures in the anterior of the vertebral body. This type of injury is classified as compression-flexion and may be produced by compressive loading of the head, with or without actual head rotation.

3.6 Lateral bending injuries

Lateral bending occurs when there is a side or oblique impact. This is usually accompanied by shear and axial loading. Injuries characteristic of this type of loading are lateral wedge fractures of the vertebral body and fractures to the posterior elements on one side of the vertebral column. Avulsion of the brachial plexus can also occur. When the neck is subjected to twisting, unilateral facet dislocations or unilateral locked facets are seen (Moffat et al., 1978). However, pure torsional loads on the neck are rarely encountered in automotive crashes.

4. Neck injury tolerance

There is no widely accepted tolerance for the various loading modes on the neck. The reasons for this inability to set tolerance levels are many. The spine is a multisegmented column with nonlinear structural properties. Its geometry is complex, it produces large strains at physiologic loading, 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 the rate of loading. These factors are in addition to the normal biological variation in the strength of human tissue and the level of injury we are willing to tolerate. For example, serious neck problems are not encountered in frontal crashes unless the G-level of impact is very high. However, in rear-end collisions, long-term neck pain can result, even though the impact is of low level.

4.1 Tolerance of the neck in flexion-extension

There is ample evidence that the neck can take a fairly high frontal deceleration without injury. Ryan (Ryan, 1962), using himself as a test subject, withstood a 23-G impact without injury. He was wearing a single belt with tighteners. Ewing et al. (1969) conducted volunteer tests at 10 G and reported only belt contusions from the military harness used. Head accelerations at the mouth mount reached (38:6 ± 6:8) G. Similar tests with male volunteers performed by Cheng et al. (1979) reached a maximum sled deceleration of 10 G with no reported injuries. However, in the same series, only one of three female volunteers was willing to reach the 10-G level. The main reason given for discontinuing the tests was the intolerable whipping of the head due to weakness in the neck. Volunteer test results reported by Mertz & Patrick (1967, 1971) are the most frequently cited and widely used. The only volunteer was Prof. Patrick himself. He withstood a flexion moment of 59.4 N m (Newton-meters) with neck pain. This was defined as the pain threshold. At 87.8 N m, he had an immediate onset of pain and prolonged soreness. This was defined as a flexion injury threshold.

Many cadaveric studies on neck flexion have been reported. Lange (1971) produced a variety of neck injuries at high levels of sled deceleration in both frontal and rear-end impacts. The injuries were above human tolerance, and the only reason for citing this paper is to emphasize the fact that the observed “disc ruptures” were, in fact, disc separations from the endplates or transvers cleavage through the centre of the disc, accompanied by rupture of the longitudinal ligaments.

4.2 Tolerance of the neck in extension

A large number of studies has been conducted to study the problem of whiplash-associated disorders (WAD). These were mostly at low impact levels aimed at understanding the causes of neck pain resulting from minor rear-end crashes. On the other hand, there were the studies by Clemens & Burow (1972), who created disc injuries that were frequently associated with anterior longitudinal ligamentous rupture. There were also some joint capsular tears and bony fractures. Because of the overly severe input used (approximate sled acceleration of 25 G), it was not possible to establish a threshold for any of the injuries that were documented. Returning to the work of Mertz & Patrick (1967, 1971), we find that the static limit for Prof. Patrick was 23.7 N m and the average static limit from 10 volunteers was 21.2 N m. Their dynamic results show that the moment tolerated at the base of the skull was 16.7 N m for Prof. Patrick. The proposed noninjurious limit is 47.4 N m or twice the static limit of Prof. Patrick, and the proposed ligamentous injury limit is 57 N m. This limit is based on ligamentous damage to a small cadaver at 33.4 N m, which when scaled to the size of Prof. Patrick was 57 N m. The scaling method used was proposed by Mertz & Patrick (1967). These limits may be too high for neck pain associated with whiplash. Pain can occur without any visible damage to the soft tissue. Microscopic examination of the tissue may be necessary to establish a basis for whiplash-induced pain.

4.3 Tolerance of the neck in lateral bending

There do not appear to be much tolerance data of the neck in lateral bending. Analysis of volunteer data obtained by Ewing et al. (1977) by Wismans & Spenny (1983) show that there were no obvious injuries from runs made at 5 to 10 G. These tests resulted in a lateral bending moment of 20 to 60 N m and lateral rotations of 52 deg. Cadaveric studies have been conducted by Kallieris et al. (1987) simulating three-point belted near-side occupants. The 58 cadavers tested ranged in age from 19 to 65 years and the impact speeds were between 40 and 60 km/h. A variety of injuries, ranging from AIS 1 soft tissue damage to AIS 3 or higher bony fractures were found, frequently at the C6 level. The maximum head resultant acceleration for these tests was 163 G. Far-side lateral impacts were studied by Horsch et al. (1979) and by Kallieris & Schmidt (1990). When an in-board shoulder beltwas used, some AIS 1 cervical injuries were found in both studies. However, in the older cadavers used by Horsch et al. (1979), they found transverse clefts of cervical discs as described above. It is not clear whether the tolerance values from this study are valid or not. The delta V for the Horsch experiments was between 33 and 37 km/h. In the second study by Kallieris & Schmidt (1990), the cadavers used were younger and only AIS 1 injuries were found for a delta V of 50 km/h.

5. Conclusions

Neck injuries can range from mild to catastrophic. Generally, the injuries involving the spinal cord at the higher cervical levels are life threatening whereas those at the lower levels can result in paralysis. To injure the cord, it is necessary to disrupt the alignment or integrity of the cervical column. Burst fractures of cervical vertebral bodies can propel fragments into the cord and cause permanent cord damage. Subluxation of one vertebra over another decreases the size of the spinal canal, again causing cord damage. It is not necessary to sever the cord to produce quadriplegia. If the cord is impacted or crushed temporarily, sufficient damage can be done to paralyze the extremities. In the upper cervical area, separation of the atlas from the occiput is generally a fatal injury. Other life-threatening injuries to the upper column are multiple fractures of the arches of C1 and fractures through the pars interarticularis of C2 (hangman's fracture). Milder forms of cervical injury include the so-called whiplash syndrome caused by a rear-end collision. Although clinical literature frequently describes it as a real injury, the picture is confused by a multitude of claims of an injury for which the etiology is unknown.

This chapter summarized research aimed at providing some biomechanical characteristics of the neck in a form that will be useful in the design of protective systems and in the development of societal strategies to reduce the number of cervical spine injuries. To this end, various biomechanical models (i.e. animal/human models and numerical models), injury mechanisms and injury tolerance of the neck were presented.
Much work remains to be done. Improvements in numerical models of neck dynamics has been dramatic and remains promising. Future research to develop injury prevention strategies that capitalize on these results will be required. Understanding the effects of the spinal musculature, tensile neck injuries, and the unique features of the pediatric spine also remain as goals for the next decade.

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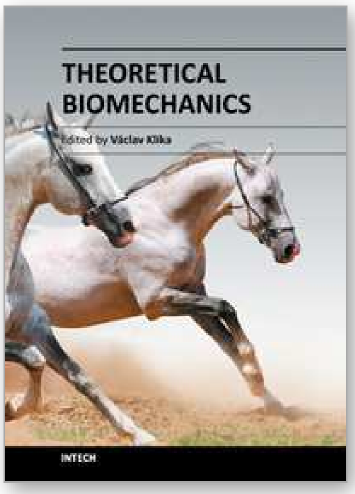

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