BIOMECHANICAL IMPACT TOLERANCE CHARACTERISTICS OF THE HUMAN NECK

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INTRODUCTION

Cervical spinal column injuries secondary to vehicular crashes can be severe and costly to the individual and to the society as a whole. Injuries involve bony damage in the form of fractures with or without dislocations and/or soft tissue ruptures such as intervertebral disc disruption and ligament tear. Our understanding of the mechanism and the biomechanics associated with these injuries comes from an analysis of epidemiological, clinical and experimental research [1, 4, 6-11, 14, 17]. Epidemiological studies have classified these injuries in a vehicular environment based on factors such as incidence, type of impact and occupant seating location. Databases such as the National Automotive Sampling System and Fatality Analysis Reporting System have been traditionally used to further analyze injuries. Clinical studies have included the retrospective evaluation of the patient using modalities such as radiography, computed tomography and magnetic resonance imaging. These studies can provide important information regarding the physiological and anatomical status of the patient, and the determination of the mechanisms of injury on a retrospective basis. However, from these studies it is difficult to quantify the actual load vector responsible for the production of the injury and the associated biomechanical variables. Depending on the extent and severity of the external load vector applied during the crash event, different types of injuries can occur to the human neck structure. Commonly encountered cervical injuries are classified into noncontact related (inertial loading) and contact related (with head impact) trauma. For example, cervical spine injuries resulting from a low speed, rear-end vehicular-collision caused by inertial loading are often considered to be of the noncontact type. In contrast, injuries arising from contact of the human head with the vehicular interior or the exterior surfaces belong to the contact category. Bony damage such as burst and wedge fractures associated with the disruption of the posterior ligaments are typical examples of contact induced neck injuries in a motor vehicle environment. This paper focuses on the correlation between the loading mechanisms and biomechanical quantities associated with cervical spine injury due to head impact.

MATERIALS AND METHODS

Unembalmed human cadaver head-neck complexes were used in the study. The specimens were selected through an evaluation of medical records and radiographic examination to eliminate bone disease, spinal disease or cancer. The subjects were screened for HIV; and Hepatitis A, B and C. Standard guidelines and laboratory practices were adopted in the biomechanical study. The demographics of the subjects were obtained which included documentation of age, height and weight. After procurement and selection, the head-neck complexes were isolated by transecting at the T2-T3 intervertebral disc space. Radiographs of the specimen in the frontal and lateral projections were obtained. Two-dimensional computed tomography (CT) images were obtained in the axial and sagittal planes (High-Speed Advantage, General Electric, Waukesha, WI). The head-neck complexes were sealed in double plastic bags and kept frozen at -70 degrees Centigrade. Handling and storage of human cadaver material in this manner, routinely used in biomechanical investigations, does not alter the material characteristics of the bone and soft tissues including ligament and cartilage [15-18, 20]. The cranium and its contents were left intact. The inferior end of the preparation was fixed in polymethyl-methacrylate. The distal end of the fixation was rigidly mounted to a six-axis load cell and firmly affixed to the platform on an electrohydraulic testing apparatus. The head was held in place using pulleys and dead-weights or masking tape to achieve the initial head-neck orientation. A flat metallic plate covered with an Ensolite padding was fixed to the piston of the electrohydraulic testing device. This served as the impact surface for contacting the preparation during dynamic loading. A schematic of the experimental set up is included in Figure 1.

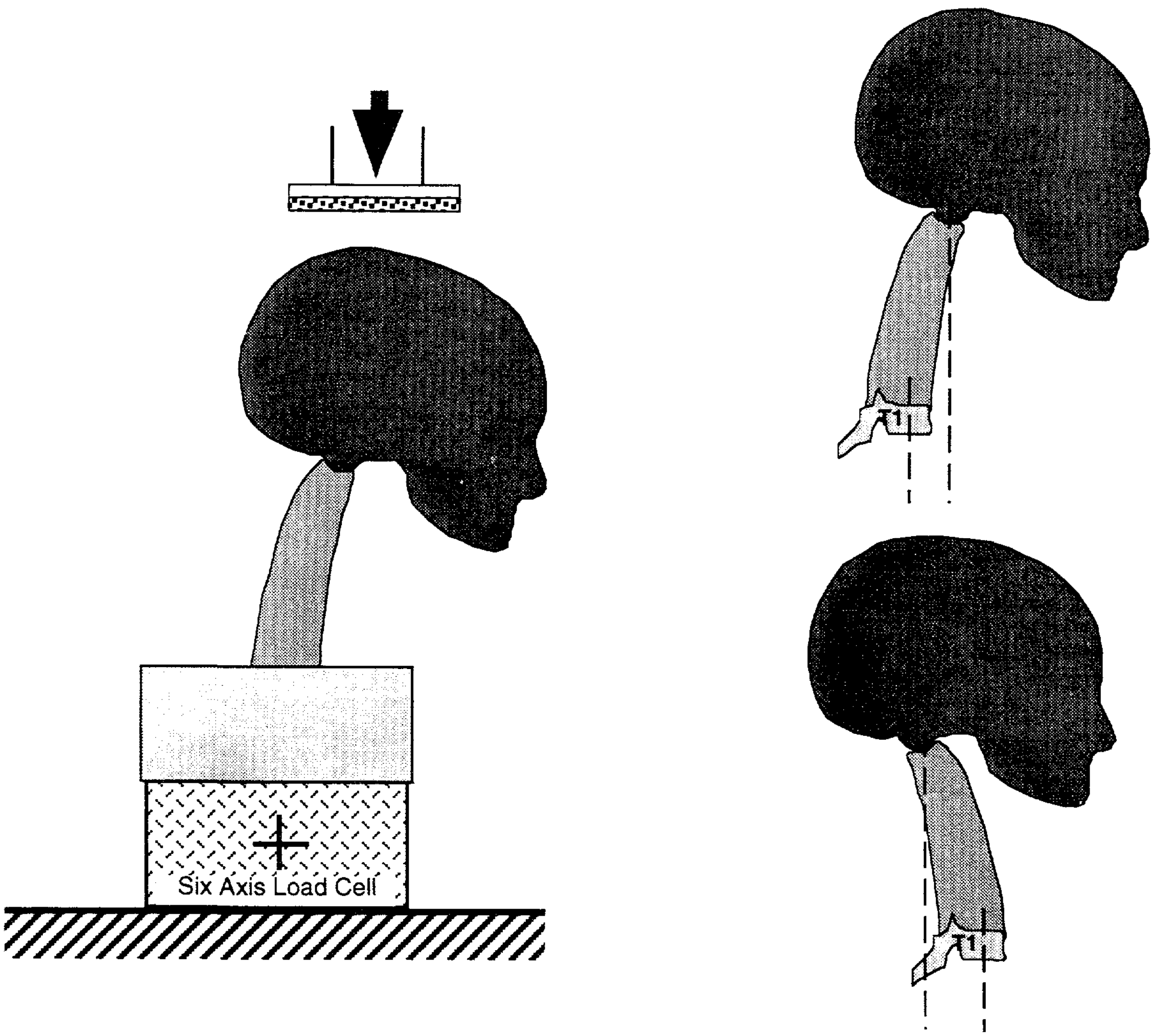

Figure 1: Schematic of experimental setup (left). Eccentricities measured from the occipital condyles to the thoracic vertebra; positive eccentricities (right top), negative eccentricities (right bottom).

Dynamic loading to the cranium was applied by the piston at rates ranging from 3 to 8 m/s. The maximum piston excursions were set at 25 to 100 mm. The head-neck specimens were tested at varying eccentricities. Zero eccentricity was defined as the position of the occipital condyles aligned with respect to the center of the first thoracic vertebral body along the direction of loading. In this position the head was flexed forward to remove the Lourdes's of the cervical spine. Eccentricities were defined as positive when the occipital condyles were aligned anterior to the first thoracic vertebra (Figure 1). When the occipital condyles were posterior to the first thoracic vertebral body the eccentricities were considered to be negative. The eccentricities were measured using radiographs taken prior to head impact. Each specimen was impacted once with the above initial conditions. Following the dynamic impact, the specimens were macroscopically examined, radiographs were taken, and CT images were obtained in the sagittal and axial planes. The pathology was determined using these images.

The six-axis load cell placed at the inferior end of the preparation recorded the forces and moments in the three directions. The coordinate system of reference was such that the x, y and z axes referred to the posteroanterior, right-left lateral and inferior-superior directions. A load cell (Model 9251, Kistler Corp., Amherst, NY) was attached in series with the piston of the testing device to measure the applied forces. In addition, a built-in linear variable differential transformer recorded the input displacements as a function of time. All biomechanical data were collected according to the Society of Automotive Engineers SAE J211 specifications using a digital data acquisition system. High-speed photographic images were obtained using a 16 mm high-speed camera or a digital video camera.

The failure force and bending moment at the level of injury were determined using the generalized force histories at the inferior load cell, geometry data from the radiographs and/or highspeed images, equations of equilibrium, and pathological information from post test evaluation. The bending moment at the level of injury was computed for the time when the compressive force was at its maximum. In this initial study, the compressive force and bending moment sustained by the cervical spine were considered as the primary biomechanical parameters to quantify the cervical spine injury. In order to associate these two biomechanical parameters to the initial loading conditions, one factor ANOVA statistics were performed to determine the number of data grouping cases separated according to their corresponding eccentricities. The optimal number of groups and the range of eccentricities for each group were determined when the one factor ANOVA statistics gave the lowest p values for both force and moment. Student t-tests were performed to determine the differences (significance level was chosen to be $p < 0.05$) in the force and moment parameters between any two groups.

RESULTS

A total of 28 specimens were included in the present study. Table 1 includes a summary of data.

Pathology identified by radiography and CT images included bony fractures of the cervical vertebrae with and without dislocation of the joints. Bony injuries included wedge, burst, vertical and tear drop fractures. Ligamentous injuries ranged from tear of the posterior or anterior ligaments to disruption of the entire intervertebral joint. In a majority of cases the injuries were concentrated at one level of the cervical spine; this was primarily in the mid to lower spinal areas. In general, irrespective of the eccentricity of the external load vector, bony/soft tissue damage occurred during the loading process. With the piston dynamically contacting the head-neck complex, the cervical spine experienced deformations during the loading sequence. Following the completion of the loading process (maximum piston excursion), the cervical spine sustained additional deformations secondary to inertial effects of the head. These observations were made using the highspeed photographic images. For the entire ensemble, the peak load to failure measured by the inferiorly placed load cell ranged from 650 to 6431 N (mean: 3055 N ± 267). The moments at the level of injury ranged from -37 to 127 Nm (mean: 37 Nm ± 6).

Table 1: Summary of Data

Age (years)	Height (cm)	Weight (kg)	Sex (M/F)	Eccentricity (cm)	Mechanism
39 - 95	152 - 178	50 - 91	1 / 3	- 0.5 to -0.1	CE
29 - 76	152 - 183	41 - 98	8 / 4	0.0 to 1.0	VC
39 - 82	152 - 193	48 - 98	5 / 2	1.1 to 4.0	CF
46 - 61	153 - 178	64 - 102	4 / 1	4.1 to 11.0	HF

Statistical analyses revealed the following groups to have significant ($p < 0.05$) differences in the biomechanical variables. Compression-extension (CE), vertical compression (VC), compression-flexion (CF), and hyperflexion (HF) were found to have the eccentricity in the range of -0.5 to -0.1, 0.0 to 1.0, 1.1 to 4.0, and 4.1 to 11.0 cm, respectively. The compressive force and moment were significant biomechanical factors for differentiating these groups (ANOVA factorial test). The vertical compression group sustained the greatest compressive force (mean: 3680 N ± 258; this force was significantly greater than the force sustained by the specimens in the compression-flexion (mean: 2786 N ± 182) and hyperflexion (mean: 1275 N ± 292) groups based on unpaired Student t-test. The force sustained by the specimens in the compression-flexion group were significantly greater than the forces sustained by the specimens in the hyperflexion group (unpaired t-test). However, the forces sustained by the specimens in the compression-extension group were not statistically different from that of the other three groups ($p > 0.1$). The mean bending moment at the injury level sustained by the specimens in the compression-extension group (-7 Nm ± 15) was significantly smaller than the mean moments sustained by the specimens in the vertical compression (30 Nm ± 4), compression-flexion (65 Nm ± 16,) and hyperflexion (47 Nm ± 11) groups. The bending moment sustained by the specimens in the vertical compression group was significantly lower than the bending moment sustained by the compression-flexion group.

The kinematics of the cervical spine in response to head impact had different patterns among the four groups. The spinal column in the vertical compression group generally deformed axially. In the compression-flexion group, the upper portion of the vertebral column deformed axially, while the mid or lower portion bent into flexion. The kinematic response in the hyperflexion group demonstrated a continuous increase in flexion in the cervical column. In the compression-extension group, the spine deformed axially accompanied by an increasing extension movement.

DISCUSSION

In order to quantitatively determine the biodynamics of the human cervical spine secondary to contact induced forces, several experimental approaches have been used. They include conducting dynamic tests using whole-body human cadavers employing drop techniques, pendulum impact methods, or applying loads with an electrohydraulic testing device [2, 3, 6-8, 11, 19]. Tests have been conducted using intact head-neck complexes (without the underlying human torso and extremities) employing drop techniques or loading with an actuator [3, 6, 11]. In addition, experiments have been conducted using isolated segments of the cervical spine [12]. These studies form the primary database on this topic. While the testing of intact cadavers provides a unique opportunity to include all load bearing structures in the human body, experimental difficulties exist particularly with regard to the consistent reproduction of clinically seen motor vehicle related trauma and the associated quantification of the biomechanical variables. The use of segmented regions of the cervical spine limits the extrapolation and applications of the experimental protocol since factors such as the effects of spinal curvature and orientation cannot be included in the model. Consequently, the use of the intact head-neck complex appears to be a viable alternative to produce clinically seen injuries and at the same time measure the appropriate biomechanical variables to quantify trauma. Because of these reasons, the present study used an intact head-neck model.

The fundamental mechanical parameters investigated in the present study to quantify injury and contribute to the determination of human tolerance included forces and moments. In order to measure such biomechanical variables, several controls have to be exercised while applying the dynamic load to the head-neck complex. In this study, the insult was applied to the intact cranium using the electrohydraulic testing device and the specimen was rigidly fixed at the inferior end. The vertical travel of the piston applied dynamic load to the specimen in vertical or preflexed positions. The loading condition was varied by adjusting the location of the occipital condyles with respect to the first thoracic vertebra, i.e., the eccentricity of load application. This was achieved by suitably orienting the cervical spinal column with respect to the head. All specimens were configured such that the occipital condyles were aligned at eccentricities ranging from -0.5 to 11.0 cm. The resulting forces and moments were measured by the inferiorly placed six-axis load cell. Due to the interspecimen variability of the specimen position with respect to the load cell, the bending moments measured at the load cell include the influence from the shear and compressive forces. Such moment contribution from the forces masks the real moment load experienced by the cervical spine. This effect was minimized by determining the bending moment at the injury level such that a comparison could be made in the biomechanical parameters. Cervical spine injury may be quantified by the axial compressive force alone (e.g., burst or vertical fractures of the vertebral body), by bending moment alone (e.g., tear of the posterior ligaments without fracture), or a combination of the force and moment variables.

The effects of the initial and boundary condition on the cervical spine responses have been observed by a number of studies in literature. The alignment conditions were found to affect the loads and injuries sustained by the cervical spine [8, 11, 13]. For example, the positions of the head, neck and torso with respect to the loading direction affected the strains sustained by the cervical spine [8]. The injury outcome was found to be dependent on the position of the occipital condyles with respect to T1 vertebra [11]. Other studies reported strong influences of the boundary conditions on the resulting neck forces and injury outcomes [5, 6, 19]. The force and injury severity sustained by the cervical spine increased with head restraint in drop testes [6, 19]. Although these previous studies provided evidence of such effects, quantitative correlation between the head-neck alignment condition and the resulting load vector sustained by the cervical spine has not been reported. The present study quantified such correlation between the geometric parameter and the load vector using experimental data from 28 head-neck cadaver specimens. Statistical analyses of the forces and moments demonstrated a strong correlation between the biomechanical variables and eccentricity. The specimens with eccentricities of 0.0 to 1.0 cm sustained the greatest compressive force (mean: 3680 N ± 258) while those with much larger eccentricities (4.1 to 11.0 cm) sustained the least force (mean: 1275 N ± 292). The compressive force was determined to be a biomechanical parameter that significantly differentiated the three test groups according to their pre-test eccentricities in the range of 0.0 to 11.0 cm. The compressive force was not effective however, in differentiating between groups with negative and positive eccentricities. The bending moment, on the other hand, was determined to be sensitive to the sign of the eccentricity and differentiated the test group with negative eccentricities from any of the other three groups with positive eccentricities.

A measurement of compressive force and bending moment therefore is required to determine the loading condition in terms of the eccentricity. For example, a 3000

N compressive force sustained by the cervical spine may indicate any of the following: vertical compression, compression-extension, or compressive-flexion. The sign of the bending moment can help to select or eliminate the compression-extension group. The moment magnitude can help to differentiate the vertical compression from compression-flexion groups. Similarly, a 65 Nm bending moment may suggest compression-flexion or hyperflexion. A compressive force greater than 2000 N can then indicate compression-flexion as a more likely loading mechanism than hyperflexion. The demarcation between the four groups for the forces, moments and eccentricities will become more definitive when the experimental sample size increases to cover the range of testing conditions. Additional experiments are needed to increase the accuracy of the grouping demarcations and extend the results into greater negative eccentricities. Likewise, additional tests are needed to include the effects of parameters such as age, bone condition and gender on the biomechanics of cervical spine injury. Nevertheless, this study has provided an important framework to guide future studies of cervical spine injury.

The results in this study suggest that a general tolerance criterion for the human cervical spine injury due to head impact should include force and moment parameters, and different tolerances for different mechanisms. The present study has demonstrated that the forces and moments sustained by the cervical spine are strongly dependent on the eccentricity. Four distinct groups were identified according to their eccentricities in the range of -0.5 to 11.0 cm. Each group was associated with a particular pattern of kinematic responses. Therefore, such groupings may be associated with different loading mechanisms of the cervical spine. A single force parameter may be adequate to quantify the tolerance under vertical compression where vertebral body fractures are predominant [11]. For hyperflexion injuries however, where bending of the spine is the predominant response, the same single force parameter may not only be too high, but may also not represent the resulting injury pattern. A single bending moment may effectively quantify hyperflexion or hyperextension injuries where only the ligaments are torn. A universal moment tolerance criterion derived from hyperflexion injuries will not effectively quantify vertical compression injuries. The compression-flexion and compression-extension injuries may involve ligamentous disruption and vertebral fracture as primary structural failure and therefore, may need to be quantified with both force and moment parameters. The application of such a tolerance criterion to a particular type of cervical spine injury case involves the determination of the loading or injury mechanism, and selection of the appropriate tolerance parameters and values according to the mechanism.

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