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Neck Vertebral Level-specific Forces and Moments Under G_{-x} Accelerative Loading

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ABSTRACT

Introduction:

It is important to determine the local forces and moments across the entire cervical spine as dysfunctions such as spondylosis and acceleration-induced injuries are focused on specific levels/segments. The aims of the study were to determine the axial and shear forces and moments at each level under G_{-x} accelerative loading for female and male spines.

Methods:

A three-dimensional finite element model of the male head-cervical spinal column was developed. G_{-x} impact acceleration was applied using experimental data from whole body human cadaver tests. It was validated with experimental head kinematics. The model was converted to a female model, and the same input was applied. Segmental axial and shear forces and moments were obtained at all levels from C2 to T1 in male and female spines.

Results:

The time of occurrence of peak axial forces in male and female spines ranged from 37 to 41 ms and 31 to 35 ms. The peak times for the shear forces in male and female spines ranged from 65 to 86 ms and 58 to 78 ms. The peak times for the bending moment ranged from 79 to 91 ms for male and 75 to 83 ms for female spines. Other data are given.

Conclusions:

All metrics reached their peaks earlier in female than male spines, representing a quicker loading in the female spine. Peak magnitudes were also lower in the female spines. Moments and axial forces varied differently compared to the shear forces in the female spine, suggesting that intersegmental loads vary nonuniformly. Effects of head inertia contributed to the greatest increase in axial force under this impact acceleration vector. Because female spines have a lower biomechanical tolerance to injury, female spines may be more vulnerable to injury under this load vector.

INTRODUCTION

The human segmented subaxial cervical spine spanning from the axis to the first thoracic vertebra resists the externally applied loads through deformations and internal load sharing across each level. Because the anatomical and structural characteristics are not identical across each segmental level, forces and moments vary across the entire column. From a biomechanical perspective, considerable research has been expended to determine the responses of the head-neck complex under low level physiological loads, and studies have used mainly using pure moment and quasi-static loading protocols. Laboratory studies have relied on testing individual spinal units or segmented columns from human cadavers under flexion, extension, lateral bending, and axial rotation moments, and in fewer cases, head-neck complexes and whole-body human cadaver subjects have been used.[1] Testing of segments of the spine eliminates effects of curvature that is present in vivo. Although whole-body tests can incorporate such effects, it is not possible to record loads at each level of the cervical spinal column.[2] Because of the constraints for citations per instructions to the authors, all pertinent articles are not referred in this paper.

Although the automotive literature has investigated the cervical spine biomechanics under impact loads, they are primarily focused on the determination of the forces and moments at the occipital condyles.[3] The Hybrid III anthropomorphic test device is commonly used to extract these loads in the military and civilian crashworthiness studies to improve human safety. Injury metrics are specified in the standards only at the upper head-neck junction, described in the earlier cited reference. It is important to determine the local forces and moments at different vertebral levels because the spinal column is segmented, and dysfunctions such as spondylosis and injuries are focused on segments instead of the whole column.

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OBJECTIVES

The objectives of the present study are, therefore, to determine the axial and shear forces and bending moments at each vertebral level using three-dimensional finite element models of male and female head-neck complexes under G_{-x} accelerative loading and compare the responses of female and male spines.

METHODS

The research study received the Institutional Review Board approvals, and no Clinical Trial information was required/used in this investigation.

Finite Element Model

A detailed finite element model of a male cervical spinal column was developed with a focus on the accurate representation of the geometry and material properties at the tissue level. The original developmental details are given.[4] The intervertebral disks, articular cartilage, ligaments, and passive musculature were incorporated based on their anatomic definitions from the literature. The vertebrae and articular cartilage were modeled using solid elements. Inertial properties were incorporated by assigning the solid vertebral elements with the density of cancellous bone and surrounding them with a shell representing the cortical bone. The upper cervical spine, consisting of the axis, atlas, and basilar part of the occipital bone, is a complex joint stabilized only by the upper cervical ligaments. A layer of solid elements representing cartilage was modeled on each articular surface, including the occipital condyles, within the upper cervical spine. Each intervertebral disk included discrete representations of the nucleus pulposus, ground substance of the annulus fibrosus, fiber laminae, and cartilaginous endplates. The nucleus occupied approximately one-half of the cross-sectional area of the entire disk. Both the nucleus and annulus ground substance were modeled using solid hexahedral elements, while the annulus fiber laminae and cartilaginous endplates were modeled using quadrilateral shell elements.

The ground substance was a simulated as a hyper-elastic foam and defined using the Hill strain energy function.[5] The annulus fibers of the disks were defined using membrane elements with tension-only directional fibers embedded in the ground substance.[6] The fibers in the anterior annulus region were defined in a crisscross manner, while fibers in the posterior region were defined in the vertical direction. The anterior annulus fibers did not form a continuous ring with the posterior annulus fibers, accounting for the bilateral uncovertebral anatomy of the spine. The nucleus was meshed with a posteriorly displaced position to account for the anteroposterior asymmetry of the annulus of the disk.[7] An incompressible fluid with a bulk modulus of 1,720 MPa was used to simulate the nucleus material of the cervical intervertebraldisk.[8]

The upper and lower cervical spine ligaments were attached to the vertebrae and skull based on published anatomical studies using membrane elements. The ligaments were characterized using normalized nonlinear stress-strain data obtained from literature. The muscle groups were incorporated using the origin-insertion location details from anatomical and clinical studies. Twenty-five muscle groups were represented, with each muscle group divided based on their attachment points. This resulted in 87 symmetrical muscle pairs, with the muscle divided into a series of beam elements. The skull was modeled as a rigid body connected to the subaxial spine by the atlantal-cranial ligaments, and the head mass was defined as a point mass with principal moment of inertia obtained from literature. Figure 1 shows the finite element model. Mapping block-based mesh morphing approach with used to develop the size-scaled female model.[9] The model was exercised using whole body human cadaver sled experimental test data and validated with its head kinematics outputs. With 12 cores and 64 GB ram, the run time was approximately 15 hours for each simulation.

Experimental Data

Previously published in house sled test data on the kinematics of the first thoracic vertebra was used in the present study. Briefly, the human cadaver specimen was seated in a rigid seat with a seat pan angle of 15 degrees with respect to the horizontal axis. The seat back angle was 25 degrees with respect to the vertical axis. Figure 2 shows a schematic of the experimental test setup. The Frankfort plane of the subject was maintained parallel to the horizontal plane. A tetrahedron-shaped nine accelerometer package was secured to the head of the subject.[10] The three linear accelerometers along three orthogonal axes were fixed to the vertex, and two linear accelerometers were placed on the other vertices. These sensor devices recorded the head kinematics in three dimensions. A custom mount was attached to the first thoracic vertebra. Its two brackets were attached bilaterally with minimal disruption to bilateral soft tissues. A triaxial accelerometer with a triaxial angular rate sensor was attached to a lateral bracket. To determine the kinematics, a set of three noncollinear retroreflective targets were secured to the first thoracic vertebra using the accelerometer mount. A similar array of targets was attached to the skull. After the test, the head was isolated, and its center of gravity was computed. The three-dimensional locations of the center of gravity of the head and occipital condyles were determined using a three-dimensional coordinate measuring machine. The moments of inertia were measured using a torsion pendulum.

The biomechanical signals from the accelerometers were captured according to the Society of Automotive Engineers J211b specifications, standard in impact biomechanics. The data from the retroreflective targets were gathered using an optoelectronic stereophotogrammetric 20-camera motion tracking system at 1000 frames/second. The linear and angular accelerations of the head were computed using the recorded set of nine accelerations on the vertex and head geometry. The three-dimensional kinematic model was used

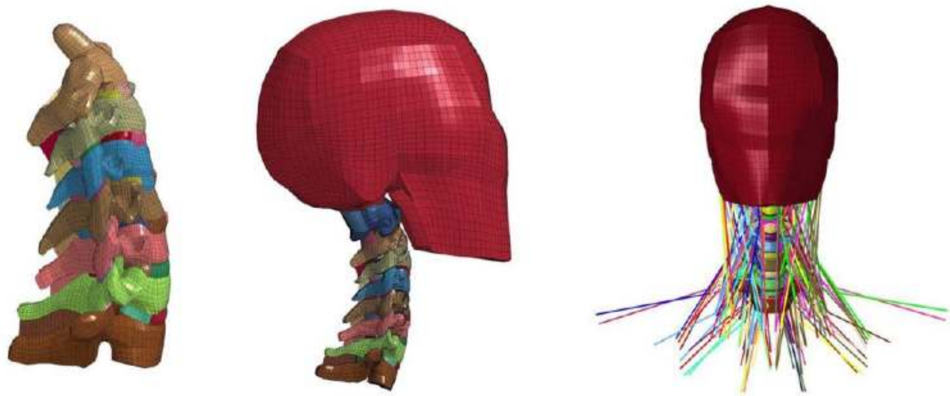

FIGURE 1. Finite element model of the head-neck complex showing the subaxial spine (left), head-spine, and head-spine with muscles.

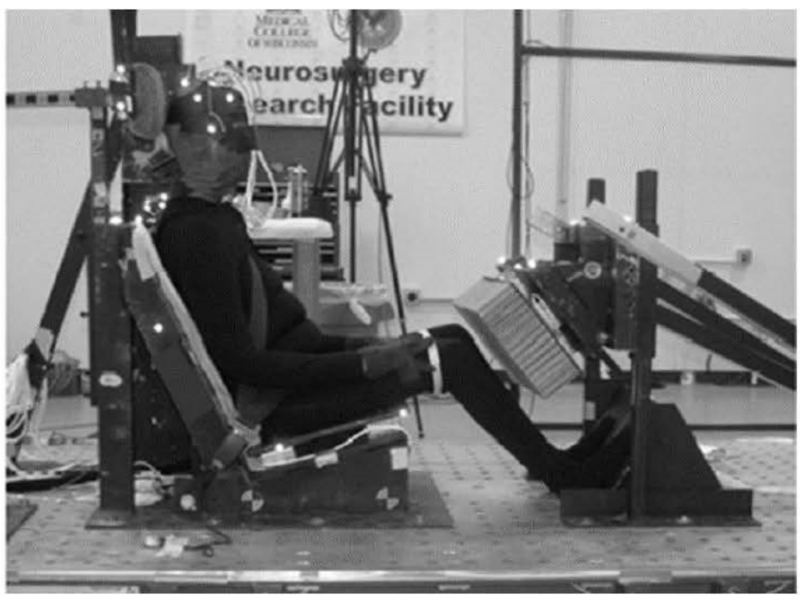

FIGURE 2. Test setup used to obtain input data at the first thoracic vertebra.

to determine the position and orientation matrix of the head with respect to the thoracic vertebra on a temporal basis.

The input data to the finite element model consisted of three linear accelerations and three angular velocities of the base of the neck along the anatomical *x*-, *y*-, and *z*-axes (Appendix 1). The finite element model was validated with the experimental kinematics of the head. The model size was then decreased by 10%, reflecting a smaller size anthropometry, representing the female head-neck model. The responses of the female and male models were analyzed to meet the objectives of the study, i.e., segmental forces and moments and their times of occurrence and comparison between the two spines.

RESULTS

The age, stature, and total body mass of the Post-Mortem Human Surrogate (PMHS or human cadaver) specimen was 83 years, 174 cm, and 86 kg, respectively. The input time histories applied to the first thoracic vertebra is shown in Appendix 1. A comparison of the model-predicted head kinematics with experimental data is also shown in Appendix 2.

Peak Magnitudes of Force and Moments

For the male spine, the segmental axial forces at the C2-C3, C3-C4, C4-C5, C5-C6, and C6-C7 levels were 1.10, 0.89, 0.99, 0.97, and 0.87 kN, respectively. The segmental shear forces at these levels were 0.52, 0.31, 0.38, 0.25, and 0.45 kN, respectively. The bending moments were 13.0, 22.1, 27.8, 38.8, and 40.2 Nm, respectively. For the female spine simulations, the segmental axial forces decreased by an average of 7.2%, moments by an average of 4.8%, and shear forces by an average of 3.3%, across all segmental levels. Figure 3 shows the peak axial and shear forces and bending moments for both male and female spines.

The time of occurrence: The time of occurrence of the peak axial forces in the male and female spines ranged from 37 to

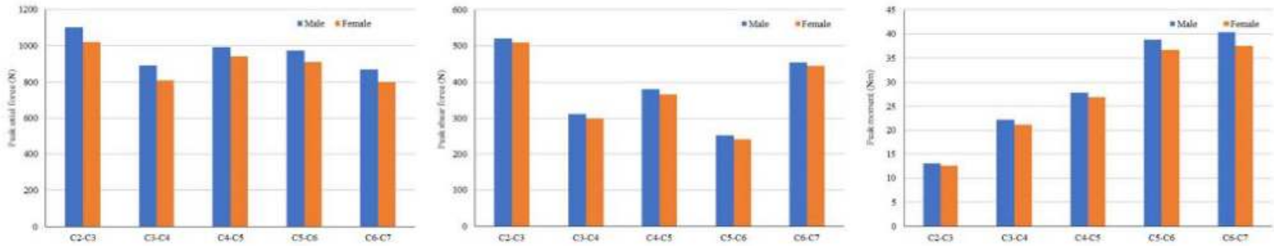

FIGURE 3. Comparison of male and female responses at segmental levels: left—axial forces, middle—shear forces, right—bending moments.

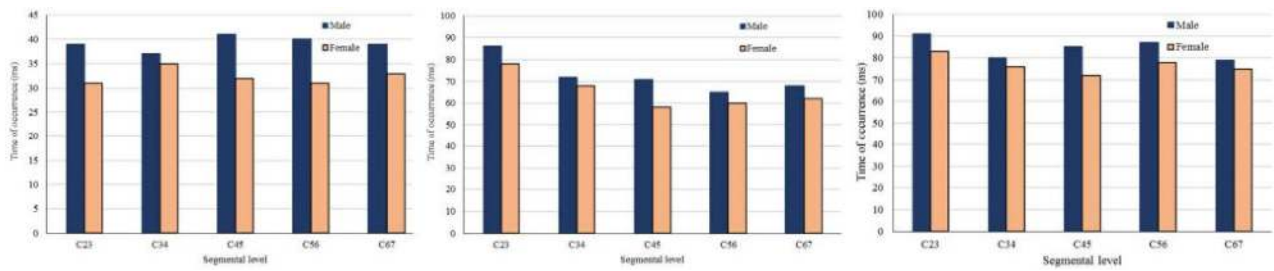

FIGURE 4. Comparison of male and female response times of occurrences of peak force and moments segmental levels: left—axial forces, middle—shear forces, right—bending moments.

41 ms and from 31 to 35 ms, respectively. The peak times for the shear forces in the male spine simulations ranged from 65 to 86 ms, and in the female spine simulations, it ranged from 58 to 78 ms. The peak times for the bending moment ranged from 79 to 91 ms for the male and 75 to 83 ms for the female spine simulations. Figure 4 shows a comparison of these times for all biomechanical metrics and both sexes.

Representative time histories: Depictions of the temporal segmental responses at all levels of the subaxial column for the male and female spine simulations for the axial force, shear force, and bending moment are given (Appendices 3 and 4). The morphologies of all the responses are similar, and a lag can be observed in the male head-neck complex response. This journal allows only four figures in the main article.

DISCUSSION

As stated in the introduction section, the objective of the study was to determine the segmental forces and moments of male and female spines using a finite element model. There is a need to understand the load sharing process in the spine by quantifying the vertebral level-specific loads on the cervical spinal column because of its segmented nature; in other words, the human spine is a heterogeneous column with differing constituents (bone and ligaments, for example) and widely varying material properties (annulus of the disk to cortical bone, for example). Because injuries occur more commonly at the lower spine, and procedures such as anterior cervical diskectomy and fusion, and artificial disk replacement are mostly done for lower spine disorders such as herniation, spondylosis, and diskogenic pain, load transmission at the lower levels of the spine are relevant.[11–13] Computational finite element models have the unique ability to quantify such loads. Thus, the present hybrid approach of coupling experiments with finite element modeling and using specimen-specific outputs in the form kinematics at the first thoracic vertebra (input to the head-neck complex finite element model) and comparing the kinematics of the head center of gravity for validation of the finite element model is effective to delineate the internal load distribution within the neck structures. This is a uniqueness of the present study.

The quantifications of these types of intrinsic segmental forces and moment responses are extremely difficult, if not impossible, to obtain from experiments without compromising the integrity of the anatomical structure. This is because any intervention to measure segmental forces and moments via a load cell introduced into the joints alters the load path and its distribution in the spine, and furthermore, the load-cell(s) introduced experimental model produces results that do not mimic the intact human spinal column. In addition, the required excessive invasion of the human cadaver during the preparation process may render the head-neck unstable. This issue can only be solved using an anatomically accurate and validated finite element model, and hence, this process was followed in the present investigation.

Model Validation

The present model was validated using human cadaver sled test data from our laboratory. The input kinematic data consisted of three linear accelerations and three angular velocities of the first thoracic vertebra along the anatomical *x*-, *y*-, and *z*-axes (Appendix 1). These data were derived using the kinematic targets and sensors attached to the first thoracic vertebra. Simple duplication of experimentally gathered sensor signals is not appropriate as mounted sensors were not actually to the first thoracic vertebra. This is an experimental issue. Although it is possible to introduce sensors on the vertebral body, invasive procedures are complex and not well developed, disrupt the integrity of the spine, and may lead to an unrealistic response. Use of multiple sensors (linear and angular devices) are challenging to any experimentalist to attach to the first thoracic vertebra of a whole-body human cadaver. Thus, a custom mount was used, which placed these sensors away from the vertebra, while the mount was rigidly attached to the vertebra. Hence, appropriate transformations of the recorded signals using the locations of the sensors and kinematics were needed to obtain the input parameters to be applied at the first thoracic vertebra. A comparison of the morphology of the kinematics of the head, the farthest region from the region of delivery of impact acceleration, showed reasonably good agreement between the model and experiment (Appendix 2). This agreement provided confidence in the further analysis, i.e., determination of segmental forces and moments across all subaxial levels of the cervical column. Because the actual geometry of the spine was not (available) incorporated in the present simulation, the findings from the study represent a first step in the understanding of the male and female spine responses to G_{-x} impact accelerative loading.

For validating the segmental forces and moments, results were used from a different head-neck finite element model.[14] At all segmental levels, the axial forces were greater for the male model and lower for the female model compared to the literature results. The shear forces and moments were within 60 N and 3 Nm between the male or female model with the literature results. Acknowledging differences in the modeling processes between the literature and present study, the authors consider the models as validated. Data are shown in Appendix 5.

Rationale for the Method Used for the Female Model

This study quantified the responses of male and female spines using a scaling approach and with the assumption that both belong to the same age group. In this context, it is worth noting that the automotive studies have largely focused on the male, as standards first evolved for the male occupant. The current upper neck injury criteria for the frontal impacts (G_{-x} loading) is applied to the Hybrid III family of dummies, and the criteria are all scaled from one dummy to another: small female to midsize male to large male anthropometry. Although the cervical spine components of men and women are the same, female spines are not scaled down versions of male spines.[15] Automotive safety standards around the world have used, and continue to adopt, the simple linear scaling approach to specify injury metrics and criteria for men and women.[3] This is because female-specific data such as vertebral and disk geometries, material properties, and impact biomechanical experiments conducted solely using this group of the population were, and continue to be, sparse. Paralleling the current state of the art in this field, the present study, therefore, obtained the female model by linearly size scaling the male model. From this perspective, the present results are considered as a first step in the delineation of the segmental responses of female head-neck complexes under G_{-x} impacts.

Rationale for the Output Variables

The choice for the three biomechanical variables is as follows. The chosen loading mode, G_{-x} impact, is focused on the sagittal plane, and in this mode of loading, off-axis forces and moments, i.e., lateral shear, lateral flexion, and axial twist play a secondary role. Thus, the axial force that essentially distracts the head by stretching the cervical column in this mode, developed shear force along the anteroposterior axis, and bending moment in the sagittal plane are the three most important loads acting on each segment of the subaxial cervical spine. The present neck injury criteria used in automotive crashworthiness studies are also based on the axial force and the bending moment variables.[3] The addition of the shear force adds to the simple axial force and moment components of the sagittal loading mode in G_{-x} impact scenarios, and better described the segmental load sharing in the column.

Time of Occurrence of the Forces and Moments

The time of occurrence of the peak forces and moments were such that the female spine peaked early, i.e., for all the variables, female spine responded with earlier times to reach their maximum magnitudes, regardless of the segmental level or the load type: axial force, shear force, or bending moments. This suggests that the female spines sustain a greater segmental loading rate response than the male spine for an identical G_{-x} impact acceleration. The lag for the male response is as a result of its larger mass of the head. Together with its increased moments of inertia will take more time for the pulse to kinematically activate the head-neck complex, while the relatively smaller sized female spine responds more quickly to the imparted accelerative loading. This suggests that the female spines absorb/manage the energy input differently from the male spine, and as discussed later, the peak magnitudes are also different.

Forces and Moments

The temporal responses of female and male spines at all segmental levels and for all three biomechanical variables showed a similar morphology, suggesting that the two spines react uniformly to the applied G_{-x} impact accelerative loading. There were differences in the peaks and times of occurrences, however. The axial forces across all spinal levels were greater than shear forces, and this is to be expected because of the impact vector. This was true for both male and female spine models. Although the forces did not exhibit considerable variability across all levels (coefficient of variation 0.19 for shear and 0.25 for axial force), moment exhibited a larger variation (coefficient of variation 0.31). The similar distribution of the axial force implies that all levels/intervertebral joints of the cervical spine stretch more uniformly than the other two load components. This is to be expected because the driving force is inertial, i.e., head inertia as a result of the G_{-x} accelerative loading pulls the neck in tension, and this distractive force is uniform along the axis of the column. The larger variations in the bending moment across all levels shows the sensitivity of this biomechanical metric associated with the load vector, G_{-x} impact acceleration. Although both forces and moments decreased in the female spine compared to the male spine, the decrease was larger for the axial force and moment metrics than the shear force metrics, suggesting that the head inertia plays a role in the development of segmental cervical spine loads under this accelerative vector.

A comparison of the forces and moments across all levels showed that the male spine responded with greater magnitudes than the female spine, and the changes were more pronounced in the axial force than the shear force or bending moment. This suggests that the axial force is a primary variable in this mode of impact acceleration. Any increase in the head inertia properties as a result of the presence of the head supported mass, helmet, and combos such as night vision goggles adds to the axial force, and it is likely that these magnitudes increase segmental axial loads. In this study, such effects were not considered as the objectives were to examine the primary/fundamental responses of female and male spines to G_{-x} impact acceleration. Because female spines have different magnitudes of force and moment profiles, and the times of occurrences precede the male spine, they may impart increased rate of loading at all segmental levels than the male spine, and the investigations of the factors such helmet mass and inertial properties are considered as a future research.

Potential Applications

The present knowledge-based product study quantifies the biomechanical loads at different segments of the spinal column, a dataset that is not possible to directly gather from human volunteer and human cadaver experiments. As human body models are gaining more popularity in the safety engineering community, understanding the pattern of the axial and shear forces and bending moments in the various segments of the spine due to G_{-x} loading may help in whole-body computational modeling efforts, via the benchmarking process. It may also assist in the development of more biofidelic anthropomorphic sex-based manikins. Although not the objective of the present study, these data may be used to conduct parametric studies for clinical applications in which certain procedures places additional loads on the adjacent segments stemming from fusing one segment of the spinal column.

CONCLUSIONS

Using male and female finite element models of the human head-neck complex, this study quantified the segmental axial and shear forces and bending moments in the sagittal plane under G_{-x} impact acceleration loading. All three metrics reached their peaks earlier in female than male spines, representing a quicker loading for the female spine. In addition, all three metrics were lower in female spines than male spines. Moments and axial forces accentuated approximately eight and five percent in the male spine, suggesting that the intersegmental loads vary nonuniformly among the three metrics. Effects of head inertia contributed to the greatest increase in axial force under this impact acceleration vector. Because female spines have a lower biomechanical threshold to injury, female spines may be more vulnerable to injury under this load vector.

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SUPPLEMENTARY MATERIAL

Supplementary material is available at *Military Medicine* online.

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