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Vertebral Level-dependent Kinematics of Female and Male Necks Under G_{+x} Loading

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ABSTRACT

Introduction:

Size-matched volunteer studies report gender-dependent variations in spine morphology, and head mass and inertia properties. The objective of this study was to determine the influence of these properties on upper and lower cervical spine temporal kinematics during G_{+x} loading.

Methods:

Parametrized three-dimensional head-neck finite element models were used, and impacts were applied at 1.8 and 2.6 m/s at the distal end. Details are given in the article. Contributions of population-based variations in morphological and mass-related variables on temporal kinematics were evaluated using sensitivity analysis. Influence of variations on time to maximum nonphysiological curve formation, and flexion of upper and extension of the lower spines were analyzed for male-like and female-like spines.

Results:

Upper and lower spines responded with initial flexion and extension, resulting in a nonphysiological curve. Time to maximum nonphysiological curve and range of motions (ROMs) of the cervical column ranged from 45 to 66 ms, and 30 to 42 deg. Vertebral depth and location of the head center of gravity (cg) along anteroposterior axis were most influential variables for the upper spine flexion. Location of head cg along anteroposterior axis had the greatest influence on the time of the curve. Both anteroposterior and vertical locations of head cg, disc height, vertebral depth, head mass, and size were influential for the lower spine extension kinematics.

Conclusions:

Models with lesser vertebral depth, that is, female-like spines, experienced greater range of motions and pronounced nonphysiological curves. This results in greater distraction/stretch of the posterior upper spine complex, a phenomenon attributed to suboccipital headaches. Forward location of head cg along anteroposterior axis had the greatest influence on upper and lower spine motions and time of formation of the curve. Any increased anteroposterior location of cg attributable to head supported mass may induce greater risk of injuries/neck pain in women during G_{+x} loading.

INTRODUCTION

The head-neck complex is different between men and women. From an anatomical perspective, the cartilage in the bilateral facet joints is thinner and has a shorter cover for the subchondral bone in women.[1] The muscular anatomy is thinner in women than men for comparable anthropometry.[2] It is also known that the curvature of the cervical spine in a vehicle sitting posture without the personal protective equipment is less lordotic in men.[3] In size-matched populations, the mass of the head is lower in females than males.[4] A large study that examined vertebral body depth, spinal canal size, and facet angle from computed tomography images from 750 subjects, comprising 456 males and 294 females, reported that the mean vertebral body anteroposterior depth is greater in males by an average of 13% to 16% than female subjects.[5] Another study that analyzed the vertebral geometry as a function of age, gender, and stature from conventional radiographs of 180 seated subjects attributed the greater average vertebral measurements in males to differences in the distribution of stature between the genders, with the average male being taller than female.[6]

Using stature-matched males and females, one study reported that dimensions of female vertebrae in the anteroposterior direction is significantly ($P < 0.05$) smaller than male at all spinal levels.[2] The vertebral dimensions in the medial to lateral direction were also smaller, although they were not significant, and the vertebral superior to inferior height was smaller in females at some levels. In another size-matched

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study, measurements of vertebrae were made using computed tomography images of sitting height-matched and head circumference-matched human volunteers.[7] The anteroposterior combined depth of vertebral body and facet joint was significantly ($P < 0.05$) smaller in females in the sitting height-matched study. In the head circumference-matched study, all measured vertebral dimensions were smaller in females, indicating that the cervical vertebral column is smaller in women with a similar head circumference as that of men. Thus, under the same type of loading to the head-neck complex, biomechanical responses are expected to be different between men and women, and the ensuing spinal disorders as a result of mechanical loading may also be different.

Although anatomical studies provide information on parameters such as spinal morphology, the application of impact loading to the spine results in deformations, characterized by the range of motion (ROM) in clinical environments. To determine the magnitudes of such parameters, biomechanical studies are necessary. Static loads applied to human cadaver head-neck complexes and whole-body specimens are inadequate, as the effects of inertia are not included in this experimental model. Likewise, dynamic loading to cervical spine specimens are also inadequate because of the omission of the head in the experimental model.[8] Finite element models offer a unique advantage to simulate head-neck complexes with varying morphological characteristics, to apply dynamic impact loading, and to quantify the role of the geometric- and mass-related variables on the biomechanics of the head-neck complex. Delineations of parameters such as the ROMs of spinal regions within the cervical column and the timing of the occurrence of the peak motions/kinematics can be used to explain clinical injury mechanisms such as neck pain reported in the military. The objectives of this study were, therefore, to quantify the above biomechanical variables using three-dimensional finite element models (brief description given in Appendix 1) of male-like and female-like head-neck complexes under G_{+x} loading using a parametrized approach.

METHODS

Head-Neck Finite Element Model

This study used a previously developed and validated three-dimensional finite element model of the sub-axial spinal column.[9,10] The C2-T1 osteoligamentous model included the following components: (a) the cancellous core and cortical shell of the bodies, their posterior elements, that is, laminae, pedicles, and spinous processes, and dens of the axis; (b) the intervertebral disc fibers, ground substance, and nucleus pulposus; and (c) the anterior and posterior longitudinal ligaments, joint capsules of the lateral mass, ligamentum flavum, and interspinous ligaments. The spinal column was meshed with hexahedral elements. The anterior region of the annulus fibrosus consisted of 16 layers, and the posterior region consisted of 8 layers. The anterior annulus fibers did not form a continuous ring with the posterior annulus; however, a gap was formed bilaterally at the uncovertebral clefts. Twenty-three pairs of cervical spine muscles were simulated, and the definitions of the muscle point of attachments and physiological cross-sectional areas were based on dissection studies.[11,12] Nonlinear rate-dependent stress-strain relationships were used for the anterior and posterior ligaments, the Hill simulation was used for muscles, elastic properties were used for the cortical and cancellous components of the vertebral body, posterior complex, and endplates, and viscoelastic properties were used for discs. Appendix 2 shows the elements and the material properties of the spinal components used in the study. Fig. 1 shows the head-neck finite element model used in the study.

Loading, Parametrization, and Analysis

The model was subjected to changes in velocities of 1.8 and 2.6 m/s, corresponding to human cadaver head-neck complex experiments conducted by the authors in an earlier study.[13] The explicit finite element code LS-DYNA (Ansys Corporation, Canonsburg, PA) was used as the solver. The segmental ROMs at all levels from the model were compared to the response corridors reported in the cited study for validation purposes. Following the validation of the model at these two velocities, parametrization was done to incorporate population-based geometrical and physical property variations.

Mapping block-based mesh morphing was used for the parametrization. The head mass property variations parametrized were the head weight, moments of inertia, and center of mass location in the sagittal plane, that is, anteroposterior, x-axis and vertical, z-axis; the spine morphology variations included the intervertebral disc height, vertebral depth, and segmental size, and they were based on literature.[4,14] These are shown in Appendix 3. A D-optimal experimental design was used to generate various finite spine head-neck models with the above parametric variations in the spine and head geometry- and head mass-related variables. A linear polynomial regression-based analysis of variance was used to identify the contribution of parametric variations on the model responses. The model generation and parametric analysis was performed using the design optimization suite LS-OPT. The head-neck model was divided into the upper and lower spinal regions to obtain the maximum nonphysiological/S-curve formation (maximum flexion before retraction of the head) in the upper spine, time to maximum nonphysiological curve of the upper spine, and the maximum extension in the lower spine. They were used to delineate the differences between male-like and female-like spines. Kinematics and motion are used synonymously in the article.

RESULTS

Ranges of Output

The time to maximum nonphysiological curve ranged from 50 to 66 ms at the velocity of 1.8 m/s and from 45 to 58 ms at the

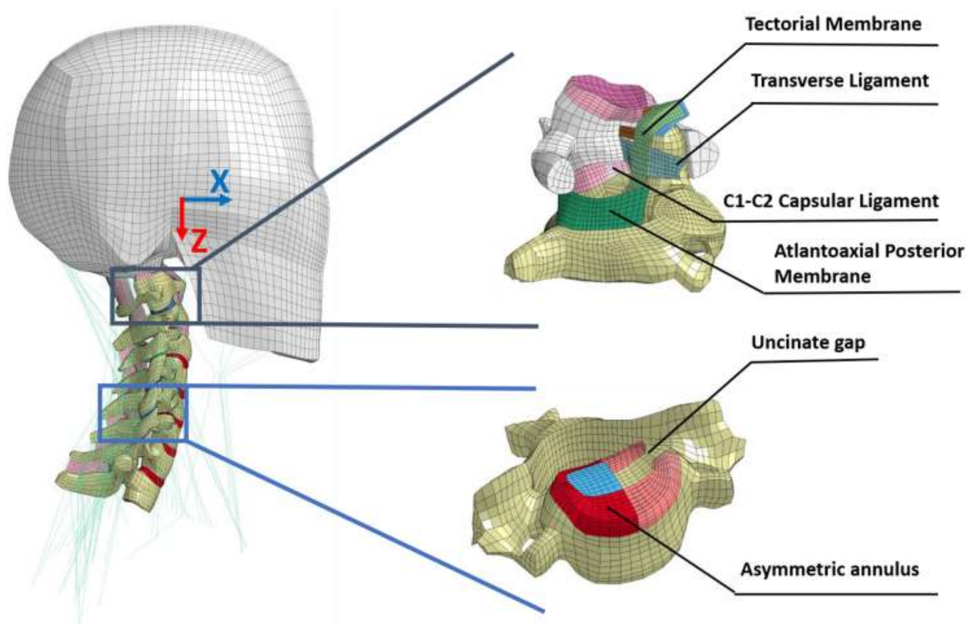

FIGURE 1. Finite element model.

velocity of 2.6 m/s. The total extension at 100 ms in the upper spine at the lower velocity ranged from 10.4 to 15.9 deg, and in the inferior spine, it ranged from 18.6 and 24.4 deg. The total extension of the sub-axial cervical column ranged from 29.6 to 40.3 deg. For the greater change in velocity, the total extension in the superior and inferior spines ranged from 10.8 to 18.0 deg and from 20.4 to 26.2 deg. The total extension of the sub-axial spinal column ranged from 31.7 to 42.2 deg at the higher velocity.

Flexion Response at the Upper Spine

Motion: The decrease in the vertebral depth and segmental size produced an increase in the flexion ROM at the upper spine at the time of maximum nonphysiological formation, and this was true at both impact severities. This represented a negatively related response. In contrast, greater disc height and greater distance of the head center of gravity along the anteroposterior axis resulted in increased ROMs at both severities for the upper spine at the time of maximum nonphysiological curve formation. This represented a positively related response. The relative roles of the other parameters, that is, head mass, head moment of inertia, and distance of the head center of gravity along the vertical axis were generally positive, but less compared to the roles of the disc height and other head mass location. These observations were true at both severity impacts (Fig. 2).

Temporal Analysis

The time to the development of the nonphysiological curve was such that a greater distance of the head center of gravity along the anteroposterior axis resulted in a decreased time. All other parameters responded positively, that is, increased times with greater magnitudes of the variables (exception, vertebral depth). These observations were true at both velocities (Fig. 3).

Extension Response at the Lower Spine

The maximum extension ROM of the lower spine was such that greater distances of the head center of gravity along both axes and moment of inertia produced greater motions, while a negatively related response was found for the vertebral depth, size, and head mass. These observations were found to be true at both impact velocities. The disc height had a negative relation at the higher severity, while it was positive at the lower severity (Fig. 4).

Sensitivity/Influence of the Variables

Regarding the sensitivity of these parameters, vertebral depth and location of the head center of gravity along the anteroposterior axis were the most influential variables for the maximum nonphysiologic curve formation. For the time factor, the location of the head center of gravity along the anteroposterior axis had the greatest influence. For the lower spine extension, both locations of the of the head center of gravity, disc height, vertebral depth, head mass, and size were influential, in the cited sequence.

DISCUSSION

As stated in the introduction section, the objectives of this study were to describe the kinematic response under G_{+x} impacts, that is, to determine the ROMs of lower spines in

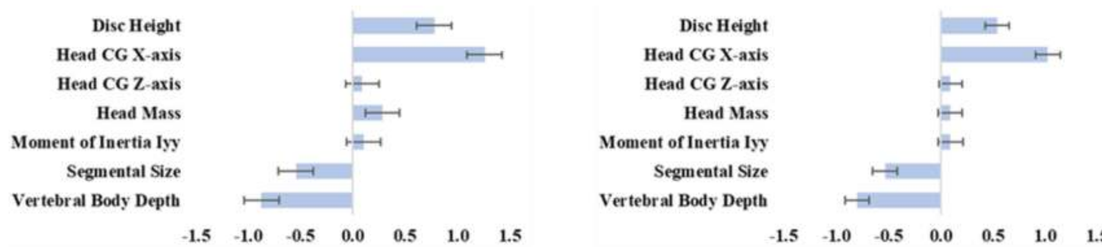

FIGURE 2. Flexion response at the upper spine with different variables.

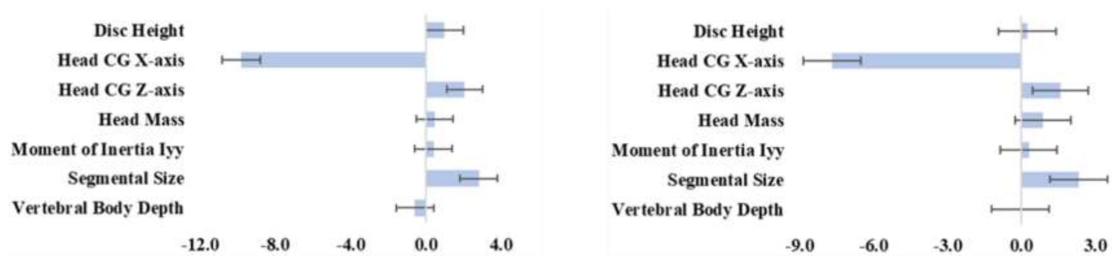

FIGURE 3. Temporal analysis with different variables.

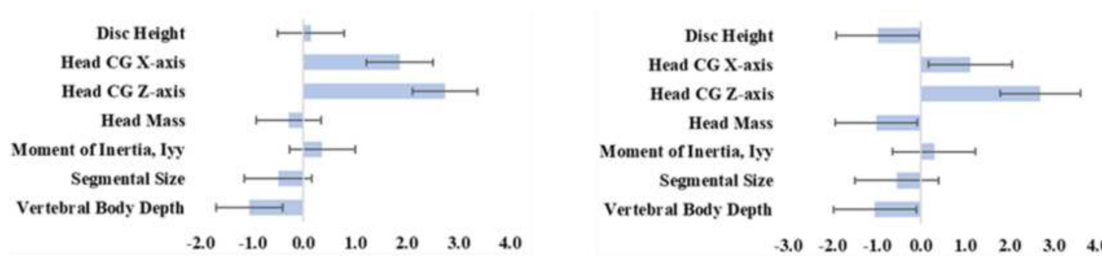

FIGURE 4. Extension response at the lower spine with different variables.

extension and ROMs of upper cervical spinal regions in flexion and their times of occurrence. This was accomplished using finite element models that incorporated variations in both spine and head properties. The ranges in the properties were obtained from published literature, all reported from the civilian population samples. The baseline finite element model was validated with the ROMs of all segments of the sub-axial column at both velocities against human cadaver experiments, a norm to perform parametric studies for impact biomechanics applications.[13] Likewise, the parametrized approach is the norm.[14] It allows to investigate the roles of multiple variables and determine the most sensitive parameter(s) on the response of the structure, in this case, ROMs of the upper and lower spines and the time to the formation of the nonphysiologic curve, that may have clinical implications (discussed later in this section).

Human volunteer and human cadaver studies have shown that the female spine less stiff than the male spine.[15,16] This implies greater ROMs in female spines for the same loading or impact severity. Decreasing the size of the spine and decreasing the vertebral body width are associated with female-like spines.[5] The results from this study showing that a decrease in the vertebral body width and size leads to increased motions (negative relationship) for both the upper spine and lower spine matches with literature. Although the published literature is silent on the relative influence or sensitivity of these factors, the present results delineated the hierarchical roles among the variables and for different biomechanical outputs: ROMs in flexion and extension at the upper and lower spines and time of nonphysiological curve formation in the neck. The consistent identification of the greatest contribution of the vertebral body depth for all the output parameters at both velocities indicate the importance of the osseous component in contrast to the soft tissue component. The decrease in disc height, representing female-like spines produced greater motions in extension of the lower spine, especially at the higher velocity severity. These hard and soft tissue parameters underscore the coupled roles of the bone and disc in affecting the female-male spine kinematic biomechanical responses under G_{+x} impacts.

Increased flexion ROM at the upper cervical spine in female spines in vivo will stretch the posterior complex more than the male spine, affecting the soft tissue connections to the cervical-occipital areas. It is known in medical literature that excessive distraction of upper head-neck complex may elicit suboccipital headaches, and this may occur more often in females as a result of their morphological characteristics under G_{+x} impacts, a finding supported from the results from this study.[17,18] Likewise, increased extension kinematics at the lower spine will stretch the posterior complex, that is, the facet joint anatomy (capsules and ligaments), and any excessive stretch may result in neck pain under this impact mode, also supported from clinical observations in whiplash patients in the civilian populations. Since this journal limits references to references that are younger than 10 years, they are not cited in this manuscript.

Greater roles for the distances of the head center of gravity along the anteroposterior and vertical axes than the mass of the head on the ROMs of the upper and lower spines indicate the importance of the geometric variables compared to the mass magnitude parameter. This may have implications in head supported mass (HSM) issues, although this study used only ranges in the in vivo head geometry and mass, that is, helmet weight was not considered.[4] The addition of the HSM increases the mass of the in vivo head; however, its effects on neck biomechanics are more influenced by the location of the center of gravity of the added mass itself in the sagittal plane. The use of head mounted devices (HMDs) such as combo and night vision goggles further alter the geometry of the HSM, and any increased distance, especially along the anteroposterior direction can have a negative impact, as motions increase with this parameter. Greater anteroposterior distance enhances flexion kinematics resulting posterior distraction, as discussed earlier, of the upper neck complex.

The early onset of the nonphysiological curve in the spine being most influenced by the location of the head center of gravity along the anteroposterior axis suggests that this parameter is more critical to control with the HSM, as early onset implies quick transformation of the natural curvature of the spine to a nonphysiological curve. The upper spine flexion, as discussed earlier, may lead to suboccipital headaches as seen in the civilian populations with G_{+x} impacts.

The present investigation incorporating population-based spine geometrical properties, and mass and geometrical properties of the head, and evaluating the combined responses of the head-neck under during G_{+x} impacts used a novel parametrized approach. It allowed a simultaneous examination of multiple factors. The use of ROM kinematics at the upper and lower regions of the spine, as biomechanical response variables, parallels clinical situations: this measure is routinely obtained from radiographs, including prescreening. Changes in ROMs overtime and with any associated neck dysfunction (pain and or early onset of spondylosis) that may be secondary to the use of the personal protective equipment (helmet with different types of HMDs) normally receive additional attention from operational and/or occupational perspectives. The location of the center of mass playing an important role in nonphysiological curve formation and the greater anterior displacement of the head center of mass (positive *x*-axis direction) resulting in a faster flexion at the upper cervical spine segments describe the local kinematics-related load-sharing paths in the head-neck complex. As the head is supported mainly by the upper cervical muscular complex, any disturbance in the activity of these muscles as a result of accentuated osteoligamentous column motions (flexion in this case) can add to the fatigue process over time. Such responses over time may elicit pain, and it may also have implications in operational performances. Anterior shifts in the location of the center of mass attributable to HSM may impart early response changes to military personnel that may be chronic in nature. Female head neck characteristics are more prone to accentuated motions and perhaps head-neck dysfunction from G_{+x} impacts.

It should be noted, however, that this study did not directly include the effects of HSM (e.g., its geometric and mass properties), and the earlier discussion on the potential role of the HSM should be confirmed by additional studies. A plausible design of such a study would be to incorporate the geometry of the actual helmet (outer shell, inner lining, etc.), mass, material properties of the components, and any helmet mounted devices, and then, analyze spinal kinematics. This is a future parametric study.

CONCLUSIONS

Using the population-based variations in the geometrical and mass-related properties of the spine and head, this study delineated their roles on the biomechanics of injuries from G_{+x} impacts. Models with greater disc height representative of male-like spines experienced more ROM in the upper spine and less ROM in the lower spine in extension. The models with female-like head-neck complex with lesser vertebral depth, however, experienced more flexion and extension and pronounced nonphysiological curve. The location of the center of the gravity of the head along the anteroposterior axis had the greatest influence on both motions and the time of formation of the nonphysiological curve. As women tend to have spines with lower vertebral depth and disc height, any increased anteroposterior location of the center of gravity as a result of HSM (helmet) and its components (HMDs) may lead to higher risk for injuries under G_{+x} impact loading to mounted personnel.

SUPPLEMENTARY MATERIAL

Supplementary material is available at *Military Medicine* online.

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