MODIFICATION AND VALIDATION OF HUMAN NECK MODEL UNDER DIRECT HEAD LOADING

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Paper Number 333

ABSTRACT

In the present study, we investigated the effects of cervical spine initial alignment, muscle activity, and the selected properties of trachea, vena cava and inactivated muscles on the responses of a finite element model of a cervical spine (*H-Model* by the ESI Group) under direct head loading. The modeling results were compared with those of volunteers. The present results suggest the following: 1) Properties of inactivated muscles, vena cava and trachea greatly affect the general head motion; 2) Maximum isometric force plays an important role in resisting neck extension; and 3) Reflex time determining the start of generation of active muscle force considerably affects the angular displacement of cervical vertebrae. Based on these findings, the modifications of the cervical spine *H-Model* were introduced. The modified model well represented the head-C7 motion, although it underestimated the head-atlas extension.

INTRODUCTION

It is commonly accepted that any mathematical model of a cervical spine for understanding injury mechanisms such as whiplash, associated disorders, and cervical cord injuries without bony damage should well simulate the relative motion between head and atlas and the relative motion between two adjacent vertebrae. However, numerous attempts to develop mathematical models of the cervical spine have primarily focused on simulation of the general motion of the head-cervical spine complex. Furthermore, experimental studies conducted on volunteers have indicated a large dispersion of data regarding such motion, which can be related to differences in the properties of tissues, muscle action and initial alignment of the head-cervical spine complex.

The present study complements previous efforts to create a model of the human cervical spine for understanding soft tissue injury mechanisms since it focuses on the simulation of not only general motion of the head-cervical spine complex but also the angular displacement of cervical vertebrae. The present attempt to facilitate such simulation consists of two steps. In the first step, we conducted a parametric study to investigate the effects of the muscle action, cervical spine initial alignment (pronounced lordosis and "upright" position), and the selected properties of the trachea, vena cava and inactivated muscles on the responses of a finite element model of the human cervical spine under direct loading applied to the head. The cervical spine model from the *H-Model* of a car occupant previously developed by Hong Ik University and ESI Group [1] was used. When conducting the present parametric study, the results obtained using this model were compared with those of volunteers. In the second step of the present analysis, the conclusions derived from the parametric study were applied to modify the cervical spine of the *H-Model* in order to improve its bio-fidelity.

METHODS

Model of Cervical Spine

In this study, we adapted the finite element model (referred to as the *H-Model*) of a car occupant developed by the Hong IK University and the ESI Group and implemented using the PAM-SAFE explicit finite element code. The *H-Model* features the entire spine, and in the present study, only its head-cervical spine complex and the first thoracic vertebra T1 were used. As the torso and upper extremities were deleted, the inferior-most ends of cervical muscles and ligaments were assumed to follow the T1 motion. The model consisting of the *H-Model* head-cervical spine complex and T1 is referred to as *H-Neck*.

The *H-Neck* model of the cervical spine consists of rigid vertebrae (shell elements) connected by the kinematic joints that simulate the inter-vertebral discs. Ligaments and joint capsules are simplified by means of nonlinear springs, and layered membrane elements are used for tectorial and cruciform ligaments. The muscle effects are represented with two distinct structures: 1) Hill-type multi-bar elements to simulate the active muscle action (Fig. 1); 2) "Flesh" built of solid elements to simulate volumetric responses of skeletal inactivated muscles, such as resistance to externally applied compression (Fig. 2).

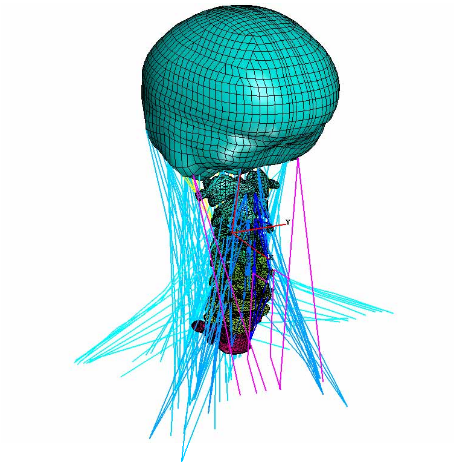

Fig. 1 Cervical muscles in *H-Neck*.

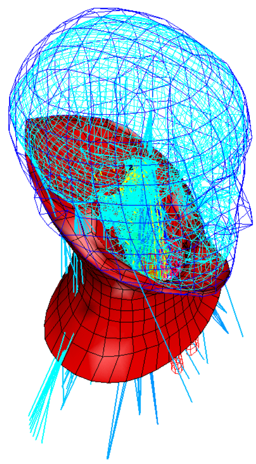

Fig. 2 Flesh in the *H-Neck*.

Cervical Spine Model Validation and Parametric Study

Validation of a cervical spine model against the results obtained using volunteers requires taking into account neuro-muscular reaction, which greatly increases the model complexity. Therefore, the present parametric study and validation of the *H-Neck* cervical spine model were done in two steps. First, the cervical model was validated against the published result of experiments conducted under a simplified set-up in which the cervical spine specimens obtained from human cadavers were subjected to either direct compressive impacts (Nightingale et al. [2]) or rear-end acceleration impacts (Yoganandan et al. [3]). Thus, no active muscle function was modeled when simulating these experiments. Second, the cervical spine model has been validated against the results of experiments in which a dynamic load was directly applied to volunteers' heads (Ono et al. [4]). These experiments were jointly conducted by Japan Automobile Research, Tsukuba University and TIT.

Model Validation of Results of Experiments on Cervical Spine Specimens

The model validation of the results of experiments on direct compressive impact to the head according to the set-up of Nightingale et al. [2] required removal of all the muscle and flesh (Fig. 3). On the other hand, when validating the rear-end acceleration impacts by Yoganandan et al. [3], the passive function of the cervical muscles (i.e., Flesh) was simulated using the *H-Neck* flesh (Fig. 4). Furthermore, the experiments by Yoganandan et al. [3] were conducted under "upright" alignment of the cervical spine specimens, whereas the *H-Neck* model exhibits pronounced lordosis (Fig. 5). For this reason, the positions of vertebrae in the model were modified to reduce the lordosis.

Comparison of the responses of the *H-Neck* model with those of human spine specimens reported by Nightingale et al. [2] and Yoganandan et al. [3] were used as a basis for understanding how accurately the overall properties of passive components of the human cervical spine, such as ligaments, inter-vertebral discs, facet joints, vena cava, trachea and inactivated muscles (i.e., flesh), are simulated in this model. Since discrepancies were observed between the *H-Neck* model behavior and the experimental results of Nightingale et al. [2] and Yoganandan et al. [3], the material properties used in this model were compared with those reported in the literature [5], [6], [7], and [8]. Then, the model properties were modified according to the literature data to minimize the differences between the model and specimen responses. These modified properties were used when simulating the experiments on volunteers and conducting the parametric study.

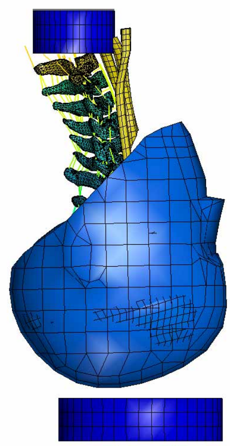

Fig. 3 Model of the experiments by Nightingale et al.

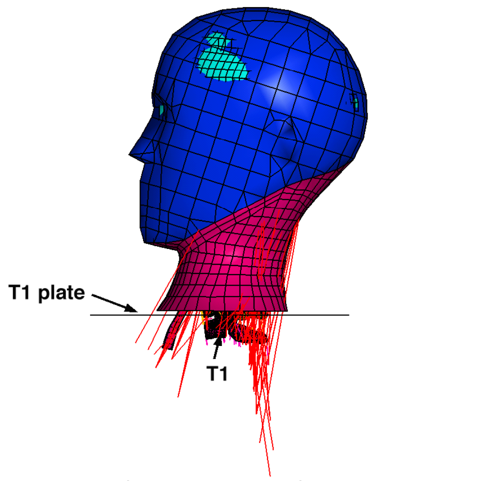

Fig. 4 Model set-up for simulation of experiments by Yoganadan et al.

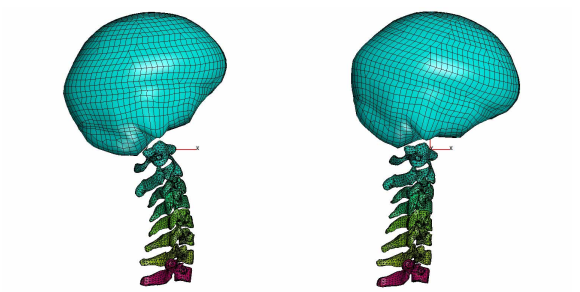

Fig. 5 Original cervical spine model with pronounced lordosis and modified model applied when simulating the experiments by Yoganadan et al.

Validation Against Results of Experiments on Volunteers and Parametric Study

Description of simulated experiments The validation was done against the results of experiments aimed at determining the mechanical properties of the upper neck under direct head dynamic loading. In these experiments, load was applied to the head of sitting volunteers in the following way. A fabric band connected with a steel cable was strapped around the head of each volunteer, and a 2-kg weight was then attached to the cable. To generate a dynamic load, the weight was freely dropped along a guide from a height of around 40 cm. The impact load was measured by the load-cell attached to the cable. Although several load directions were used by Ono et al. [4], the present study focused on "upward chin" loading conditions in which an upward-oriented force was applied to the volunteer's chin (Fig. 6). In the experiment, the cervical vertebrae rotation and facet joint kinematics were determined from X-ray cine-radiography recordings at 60 frames per second. These experiments were done under two distinct muscle tension conditions: volunteers were asked either to relax or tense their muscles. The activity of the sternocleidomastoid (SCM) and paravertebral (PVM) muscles was monitored using surface electromyography (SEMG). When the volunteers kept their muscles relaxed, the reflex time of SCM muscle was estimated to be around 70 ms. When the volunteers tensed their muscles, SCM activity started to increase immediately after the load was imposed on the head.

Simulation of experiments on volunteers The force measured by the load cell (Fig. 7) was applied to the head at a node coinciding with the point where the cable was attached to the fabric band strapped around the head. As the torso was not modeled, the inferior-most nodes of torso-neck muscles, vena cava, trachea, and neck flesh were attached to the T1 rigid body. This rigid body was fully constrained as seen from the video recordings taken during the experiments, and it was verified that the torso motion was very small and could be disregarded. Only the experiments in which volunteers kept their muscles relaxed were simulated here, so the reflex time of 70 ms was assumed for all the modeled muscles. To maintain the initial equilibrium of the cervical spine model, the initial active muscle force was designated as 0.01 and 0.005 of its maximum isometric value for the cervical flexors and extensors, respectively. When validating the results of experiments on volunteers and parametric study, the calculated angular displacements of the head and C1-C6 vertebrae in relation to the seventh cervical vertebra C7 were compared to those experimentally obtained by Ono et al. [4] using cine-radiography. The data obtained in only one experiment were available for comparison. The key input parameters used when simulating the experiments by Ono is given in Table 1.

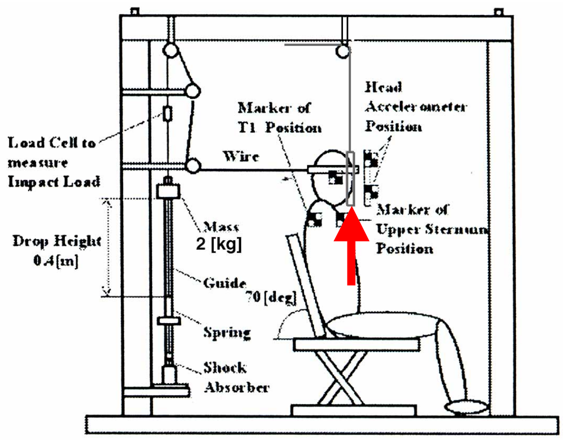

Fig. 6 Set-up of the experiments by Ono et al. [4]. "Upward Cine" load-condition is indicated by the arrow.

Table 1 Summary of material properties used in the cervical spine model.

Spine Componet	Properties
Flesh bulk modulus	0.110 MPa
Vena cava and Trachea	Deleted
Aligment	Neck alignment modified to decrease the lordosis.
Muscle	σ_{max}= 0.2 MPa. Reflex time of neck flexor; 70 ms. Neck extensor inactivated.

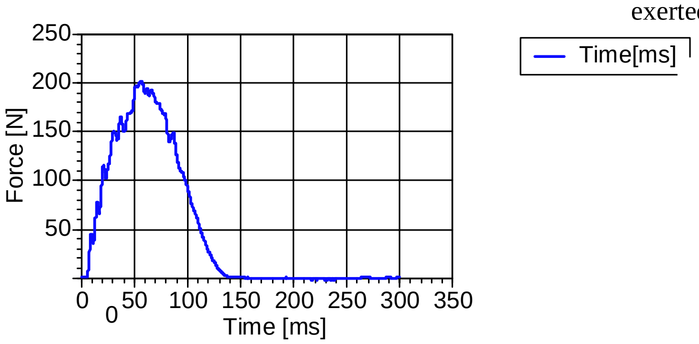

Fig. 7 Cable force measured by load-cell in the present model of "upward chin."

Parametric Study

The effects of Young's modulus the *H-Model* flesh, trachea and vena cava as well as assumptions regarding the maximum isometric muscle forces on the calculated kinematics of the head and cervical vertebrae were investigated. In conducting the parameter study, the experiments on volunteers by Ono et al. [4] were simulated.

Effect of Flesh The PAM-SAFE Hill-type multi-bar muscle elements are deficient when simulating the effects exerted on the head and cervical spine motion by inactivated cervical muscles subjected to compression. In the *H-Model*, these effects were simulated using a structure referred to as flesh built of visco-elastic solid elements connecting the head and torso (Fig. 1). The main problem with this approach is that this structure represents lumped "passive" effects of the entire cervical musculature rather than providing a simulation of the actual muscles. Therefore, the properties of flesh must be calibrated because they cannot be directly measured.

In the original version of the *H-Neck*, a flesh bulk modulus of 0.415 MPa was used. However, when validating this model against the results of Yoganandan et al. [3], it was found that the modulus of 0.110 MPa, yields results closer to human specimen responses than the original 0.415 MPa. Therefore, when simulating the experiments on volunteers in the parametric study, values of both the flesh bulk modulus of 0.415 and 0.110 MPa were used.

Effect of Vena Cava and Trachea In the *H-Neck*, the vena cava and trachea are directly attached to the head (Fig. 8). This oversimplification can obviously overestimate the effects on the cervical spine responses. Indeed, when validating the *H-Neck*, the vena cava and trachea's Young's modulus of 10 MPa originally used in this model, lead to clear underestimation of the angular displacements of cervical vertebrae. Therefore, when conducting the parametric study, two extreme cases were analyzed; 1) Parameters of vena cava and trachea as in the original *H-Model*; 2) Vena cava and trachea defined using null shells (i.e., no internal forces calculated). The latter case implies virtual disregard of the forces exerted by the trachea and vena cava on the head.

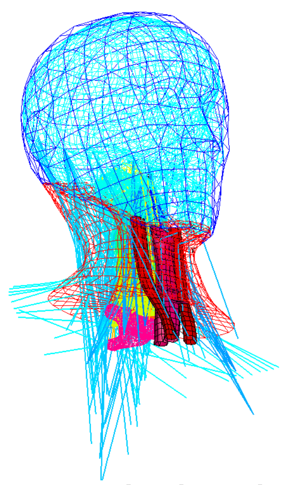

Fig. 8 Vena cava and trachea in the *H-Neck*.

Effect of Neck Alignment The original cervical spine model in *H-Neck* exhibits pronounced neck lordosis. However, in the analyzed experiments on volunteers [4], the Frankfort plane was oriented horizontally and the occipital condyles were aligned over the inferior thoracic vertebral body, which implies "upright" alignment of the neck. For this reason, the original cervical spine model was modified in the present study so as to align it almost vertically.

Effect of Muscle Activity and Discharge An important factor in predicting the muscle effects on the head-neck complex kinematics under transient load is the appropriate modeling of the force-time relation of the neck muscles, which requires information about muscle reflex time. In the current analysis, to verify the muscle effects on the head-neck complex, we utilize the reflex time obtained on the basis of the data on the direct head impact loading on volunteer. According from the SEMG data in "Upward", we could make assumption that the only flexors are activated and reflex time of the cervical flexors muscle discharge at 70 ms when simulating the test of relaxed case. To demonstrate the effectiveness of reflex time, we also calculated the muscle discharge at 10 ms.

Effect of Maximum Isometric Muscle Force The muscle effect strongly depends on the assumption regarding the maximum isometric muscle force. The typical method of estimating this force is based on knowledge of the muscle physiological cross-sectional area PCSA (e.g., Yamaguchi et al. [9]) and the maximum isometric muscle stress, $\sigma_{\max}$.

$$F_{\max} = \sigma_{\max} \times PCSA \tag{1}$$

Data on PCSA in the literature exhibit a large variation, which is related to the anthropometric differences between individuals. Similarly, the values of $\sigma_{\max}$ reported in the literature range widely from 0.25 MPa (Herzog, [10]) to 1.0 MPa (Ikai and Fukunaga [11]). It means that the PCSAs of different muscles were obtained from subjects who often differed in body size. For this reason, the present parameter study was undertaken in order to estimate the distribution of the maximum isometric force. Therefore, $\sigma_{\max}$ is determined to be 0.2 MPa from parameter study with the original PCSA which was already defined in *H-Neck* and data of Yamaguchi et al. In the current analysis we utilize the reflex time obtained from the data from direct head dynamic loading on volunteers to verify the muscle effects on the head-neck complex. According to the SEMG data in "Upward," we may assume that the only flexors are activated and that the reflex time of cervical flexor muscles discharge at 70 ms when simulating the test of a muscle relax case. The initial muscle active state was assumed to be 0.01 and 0.005 of its maximum value for the cervical flexors and extensors, respectively, to maintain the initial equilibrium condition of the cervical spine.

RESULTS

Effect of Flesh

The time history of angular displacements of the cervical vertebrae in reference to C7 was shown in Fig. 11. For the purpose of comparison, the experimental results and the results from original bulk modulus inserted in the *H-Neck* are also shown in Fig. 9 and Fig. 10, respectively. The decrease in the flesh bulk modulus from 0.415 MPa (Fig. 10) to 0.110 MPa (Fig. 11) appreciably increased the peak values of the angular displacement of each vertebra after 150 ms. Thus, this decrease improved the model bio-fidelity (Fig. 9 and Fig. 11).

Effect of Vena Cava and Trachea

The result obtained when simulating the experiments conducted on volunteers indicated important effects of properties of vena cava and trachea on the responses of the present cervical spine model. When the original properties from the *H-Model* were used (Fig. 10), the present model clearly underestimated the angular displacements of the cervical vertebrae measured in reference to the C7 vertebrae. The calculated and experimentally obtained time histories of this angle exhibited similar values when the present cervical spine model was modified by simplifying the vena cava and trachea using null shell material (Fig. 12). Thus, this modification was used when studying muscle effects. However, it should be noted that despite its high bio-fidelity in terms of the peak C0-C7 angle, the present cervical spine model with the vena cava and trachea simplified using null shell material clearly underestimated the C0-C1 extension. The magnitude of C1-C7, C2-C7, and C3-C7 angle time histories was clearly overestimated by the model. This indicates that problems related to the modeling of the vena cava and trachea were not the only source of differences between the responses of the present cervical spine model and the experimental results.

Effect of Neck Initial Alignment

Changing the initial cervical spine alignment from the lordotic one originally used in the *H-Neck* to the upright one clearly decreased the peak value of the angular displacement of cervical vertebra in reference to C7 (Fig. 12 and Fig. 13). Thus, this modification alone did not improve the model bio-fidelity.

Muscle Effect

Effect of Muscle Activity and Discharge The C0-C1 extension appreciably increased when the computation was done under the assumption that cervical flexor muscles started to increase their activity at 70 ms rather than at 10 ms (Fig. 14 and 15). Therefore, reflex time of the active muscles is an important factor for improving the bio-fidelity of the present cervical spine model.

Effect of Maximum Isometric Muscle Force It is clear that the maximum isometric force affected the peak angular displacement of each vertebra. As shown in Fig. 16, the respective peak values of the calculated C0-C7, C1-C7, and C2-C7 extension angle were virtually the same as in the experimental results. Furthermore, the C0-C7 angular displacement is larger than for other vertebra. This phenomenon is the hyper-extensional motion of the cervical spine which is also confirmed in the experiment on volunteers. However, the angular displacement of the head-atlas extension angle is smaller than in the experimental results. One possible reason for this decrease is that when simulating the experiment, the inferior most nodes of these structures were attached to the fully constrained T1 rigid body, whereas in the human body they are connected with the other deformable soft tissue which could absorb the rotational energy. This difference in boundary condition has some effect on the angular displacement of the vertebra.

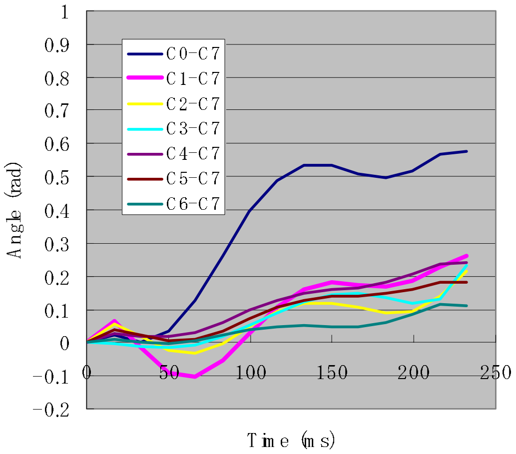

Fig. 9 Angular displacement of cervical vertebrae in reference to C7 obtained in the volunteer test (cervical muscles relaxed)

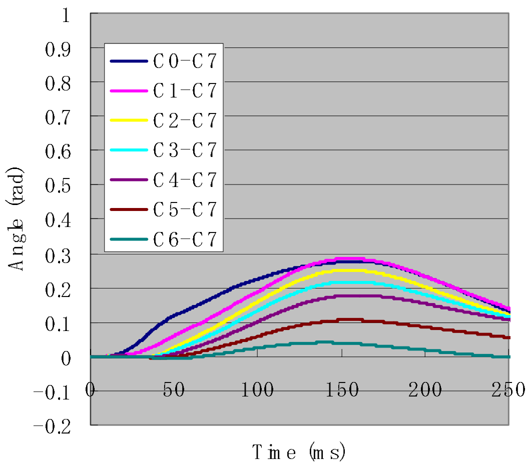

Fig. 10 Original cervical spine model (flesh bulk modulus: 0.415 MPa)

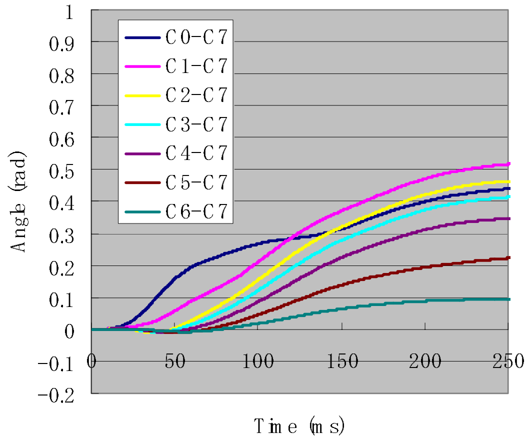

Fig. 11 Effect of flesh on the response of the present cervical spine model (flesh bulk modulus: 0.110 MPa)

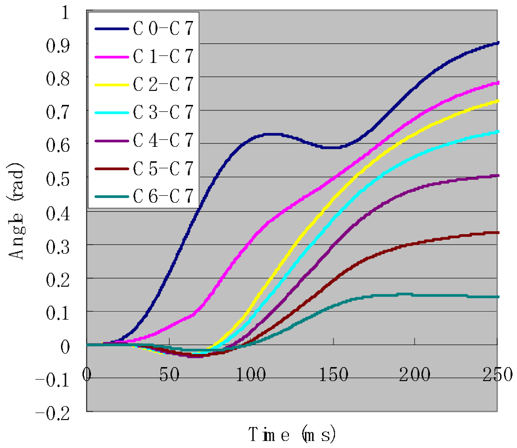

Fig. 12 Effect of vena cava and trachea on the response of the present cervical spine model. Vena cava and trachea using null shell material model. Flesh bulk modulus: 0.110MPa.

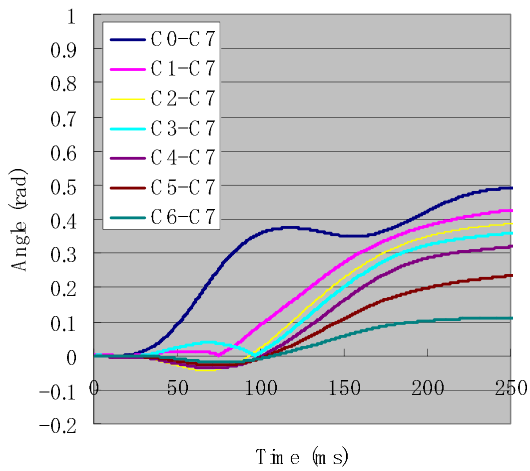

Fig. 13 Result obtained using the cervical spine model with the upright alignment. Flesh bulk modulus is 0.110 MPa, and vena cava and trachea are null shell material.

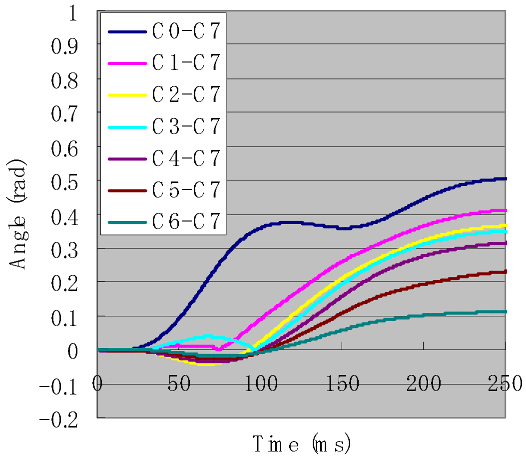

Fig. 14 Result obtained when simulating volunteer with muscles relaxed. Flexor muscles are assumed to start increasing their activity at 10 ms.

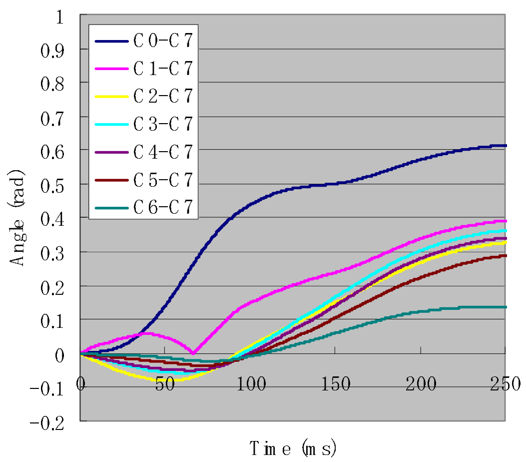

Fig. 15 Result obtained when simulating volunteer with muscles relaxed. Flexor muscles are assumed to start their activity at 70 ms. σ_{max} : 0.1 MPa

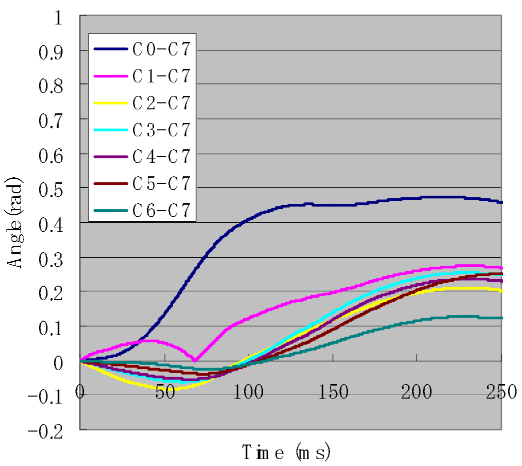

Fig. 16 Result obtained when simulating volunteer with muscles relaxed. Flexor muscles are assumed to start increasing their activity at 70 ms. σ_{max} :0.2 MPa

DISCUSSION

The simulation of the experiments on volunteers indicated that assumptions regarding the material properties of the vena cava and trachea greatly affect the responses of the present cervical spine model. The properties originally used in the *H-Neck* lead to serious underestimation of the C0-C7 extension angle magnitude. The following two explanations of this phenomenon are suggested. First, the vena cava force-elongation relationship exhibits a toe zone within which the force only slightly increases with elongation. On the other hand, in the *H-Neck*, the vena cava and trachea were assumed to exhibit linear-elastic behavior, which can lead to overestimation of the force at little elongation. The second possible explanation for the large constraining effects of the trachea in the present model of the cervical spine can be that the trachea resistive force may have been mistakenly taken into account twice: 1) When modeling trachea as such; and 2) When determining properties of the lumped hyoid muscle. However, even were one to disregard the forces exerted on the head by the vena cava and trachea, the present cervical spine model considerably underestimated the C0-C1 extension.
This extension increased when the model was computed under the assumption that only flexor muscles started to increase their activity at 70 ms and maximum muscle stress is determined to be around 0.2 MPa. These values are based on the experimental result by Ono et al. [4] and parameter study with maximum muscle force. Finally, the modified model results were closest to the experimental ones.
Thus, it can be suggested that inaccuracies in identifying the material properties of the vena cava and trachea were not the only source of the discrepancies between the responses of the present cervical spine model and experimental data on volunteers in the present study. Therefore, the following must be elucidated in further studies:

a) Physical properties of the vena cava and trachea, including the nonlinear force-elongation relationship.
b) Thickness of the vena cava and trachea walls and their anatomical attachment points on the head.
c) Values of the assumed maximum isometric forces of cervical muscles and the ratio of these forces in different muscles.
d) Assumptions made when simulating neural control of muscle activity. This can include varying the reflex time, stretch reflex threshold, and initial active state values of different muscles.

CONCLUSION

The present study suggests that the physical properties of the flesh, vena cava and trachea greatly affected the response of the cervical spine itself. The properties originally used in *H-Neck* lead to considerable underestimation of the C0-C1 extension angle. When neck flexor muscles begin to generate their active force at around 70 ms, C0-C1 extension angle is increased. The responses of the head-neck model with the muscle elements are consistent with the experimental result of direct impact head loading.

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