Effect of Muscle Activation on Head-neck Complex under Simulated Frontal Car Impact

Qing Hang Zhang, Ee ChoTeo
(School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore 639798)

Abstract: A detailed three-dimensional FE model of the whole head-neck complex was developed to investigate the effect of muscle activation on the biomechanical responses of the head and cervical spine under simulated frontal impact. For the simulation, horizontal acceleration of a half-sine-wave pulse with peak value of 5G and duration of 100ms was applied on the inferior surface ofT1 vertebral body. The results showed that the muscle force began to take effect from the 120ms, which reduce the peak flexion of all the motion segments. The maximum reduction occurred at C0-C1, in which the angle was reduced by more than 25%. The effect of muscle activation force should not be ignored under such condition.
Keywords: finite element, muscle activation, frontal, cervical spine

1 Introduction

Advances in computing technology and analysis software have enabled the development of sophisticated finite element (FE) models that have the potential to provide a more comprehensive understanding of human impact response, injury mechanisms, and tolerance. To date, many finite element models with complex geometry and multiple material compositions representing head-neck system have been developed to study the kinematics of cervical spine under various impact conditions[1]-[4]. Due to the short onset and duration impact time, the active muscle force was not included in most of these models, and only a few numerical models have implemented the passive neck muscles. In current study, a previously developed C0-T1 finite element model was modified to include muscle element with both passive and active properties. The global and segmental rotational response would be used to analyze the effect of muscle activation on the kinematics of head-neck complex under frontal car impact condition.

2 Method

The three-dimensional FE models of the skull and C1-T1 vertebrae were developed with geometrical data based on the actual geometry of a 68 year-old male cadaver specimen. Detailed examination was performed to ensure the absence of any physical abnormalities in the specimens. A flexible digitizer was used to capture the coordinates of the surface profile of the bony structures (head, C1-T1) continuously and these data were subsequently processed for the FE mesh generation. For the modelingof the intervertebral discs (IVDs), the basic geometries taken from the average values reported in literature[5] were used. Furthermore, the major ligaments and muscle groups associated with the cervical spine were also incorporated in the model, for which the attachment points were determined from literatures[6]-[11]. The modeling, meshing and analysis were performed in ANSYS 10.0 and LS-Dyna. Figure 1 shows the final C0-T1 FE model consists of 27,712 elements and 31,749 nodes and the global XYZ coordinate system. The detailed process of finite element modeling of the head-neck complex was described elsewhere[12].

Figure 1 Finite element mesh of the C0-T1 complex model under lateral and posterior view.

The material properties of the elements representing the head and vertebrae were assumed to be elastic, homogenous and isotropic since stress contributions in the hard tissues was not the major concern under general whiplash conditions. In addition, the Flanagan-Belytschko stiffness form hourglass control with coefficient of 0.12 was used to minimize hourglass energy for the disc and endplate. The nonlinear stress-strain curve describing the material properties of various ligaments were derived from experimental data of Yoganandan et al[11]. The muscle properties were represented by basic Hill-type muscle model consisting of a contractile element and a parallel elastic element to provide the active and passive muscle force, respectively. The active muscle force was calculated by the following dimensionless form as:

$F_{active}=a(t)*F_{max}*f_{TL}(L)*f_{TV}(V)$

Where $a(t)$, $f_{TL}(L)$ and $f_{TV}(V)$are the functions describing the muscle active level-time, tension-length and tension-velocity relationships, respectively, which were derived from previous studies on muscle active state modeling[13],[14] and shown in Figure 2. is the muscle peak isometric force and was calculated by the initial muscle cross section area and a peak muscle stress of 250cmN[13]. The passive muscle force was determined directly from the current length of the muscle using an exponential relationship. [14]

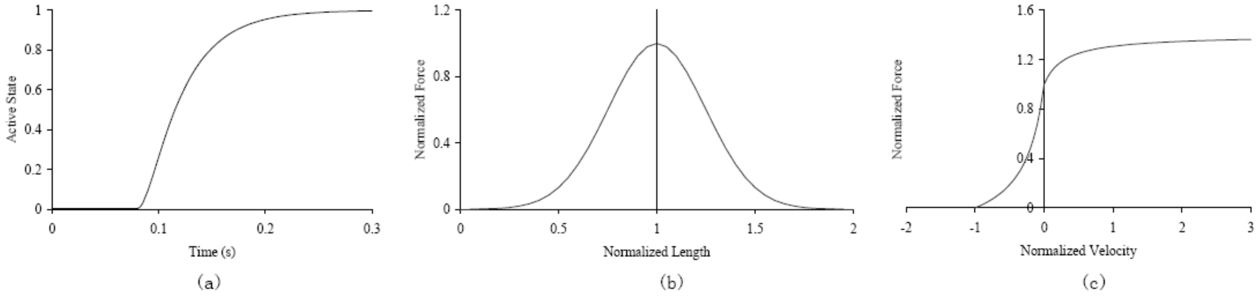

Figure 2 Curves used to control the behavior of activated muscles during simulation of the pilot ejection.

(a) The activation level of muscle as a function of time; (b) Normalized tension of activated muscle as a function of normalized length (*inill*), is the initial length of muscle; (c) Normalized tension of activated muscle as a function of normalized velocity (*inil* max*vv*) , is the maximum shorting velocity.

For the simulation, horizontal acceleration of a half-sine-wave pulse with peak value of 5G and duration of 100ms was applied on the inferior surface of T1 vertebral body, during which the T1 inferior surface was constrained to move only in horizontal direction. The predicted overall and segmental rotational angles with and without consideration of muscle activation force during 250ms after impact were compared.

3 Results

Different segments experienced different peak extension or flexion angular rotational motions at different times after impact. During the first 10ms after impact, most of the motion segments were in extension. After that, the lower segments (between C3 to C6) turned to flexion motion with different angulations while the upper segments (between C0 to C3) maintained the extension motion for much longer duration before turning to flexion motion. During the 30-80ms period, the whole C0-C7 structure formed a S-shaped curvature with extension at the upper levels and flexion at the lower levels (Figure 3). This S-curvature caused the head-lag phenomenon, i.e. the head translated posteriorly with respect to C7, with limited flexion rotation. After 80ms, all the motion segments were in flexion, which resulted in the increase of the flexion angulation of C0 with respect to C7 and the entire cervical spine formed a C-shaped curvature thereafter.

As predicted the muscle force began to take effect from 120ms, which reduce the peak flexion of all the motion segments (Figure 4). The maximum reduction occurred at C0-C1, in which the angle was reduced by more than 25% (Figure 3). The minimum reduction occurred at C6-C7, in which the angle was reduced by less than 1%. This result is accordance with the anatomic characteristics of the cervical spine segments. The C0-C2 complex is unique in that there is no disc in these two levels, and the links between the vertebrae are only muscle, ligaments and joint articulations. The lax connections in this region make it possible that the effect of the muscle force can be easier to be presented.

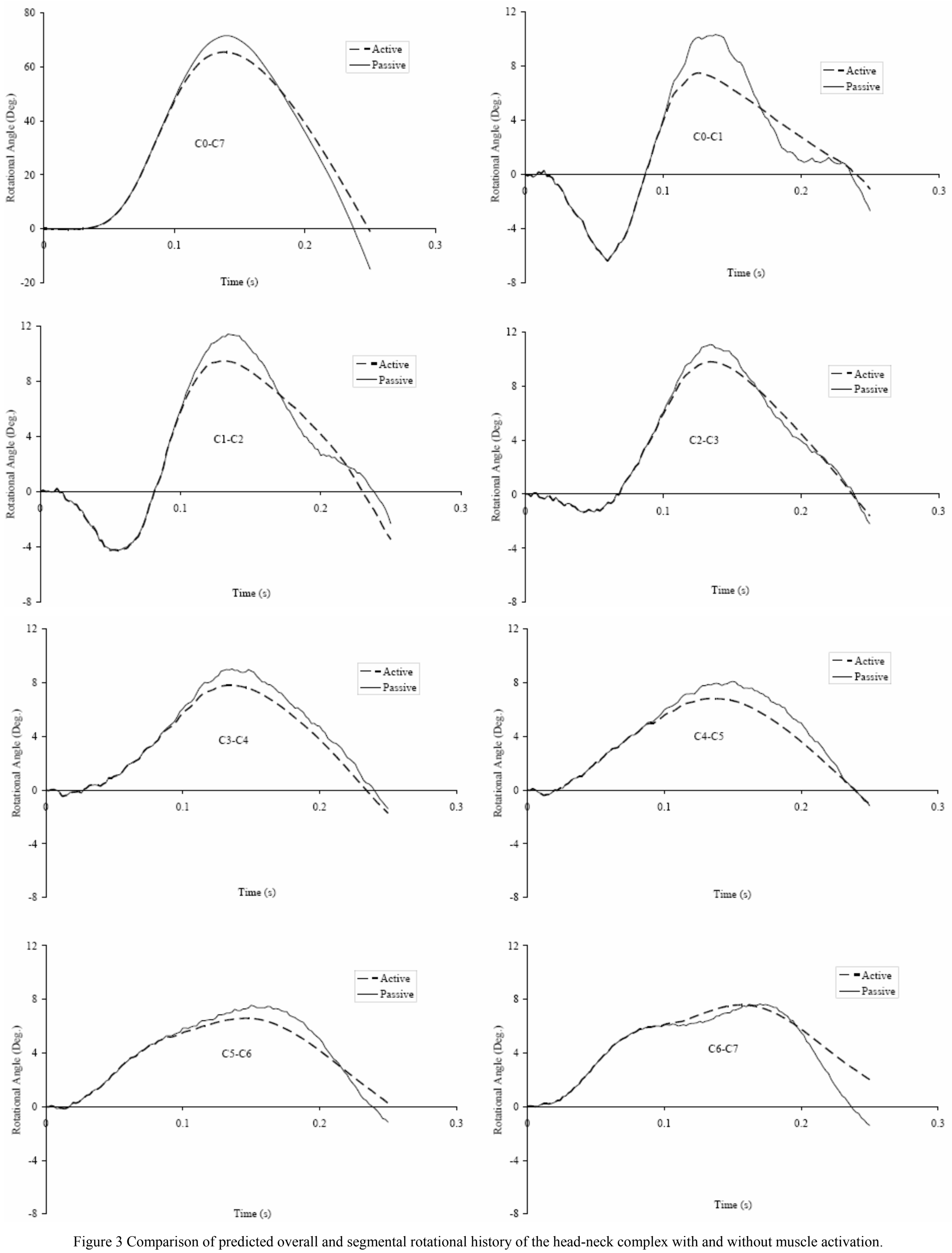

Figure 3 Comparison of predicted overall and segmental rotational history of the head-neck complex with and without muscle activation.

4 Conclusion

After 120ms during frontal impact, the activated muscle can effectively reduce the flexion of cervical segment, especially for the upper levels. The effect of muscle activation force should not be ignored under such condition.

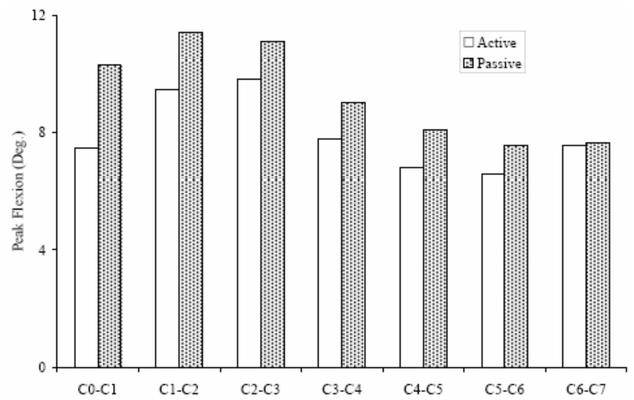

Figure 4 Comparison of peak segmental flexion with and without muscle activation.

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