Journal of Biomechanics 49 (2016) 558–564

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Impact responses of the cervical spine: A computational study of the effects of muscle activity, torso constraint, and pre-flexion

Roger W. Nightingale*, Jake Sganga, Hattie Cutcliffe, Cameron R. 'Dale' Bass

Duke University, Department of Biomedical Engineering and Division of Orthopaedic Surgery, Box 90281 Durham, NC 277080281, United States

A R T I C L E I N F O

Article history:
Accepted 8 January 2016

Keywords:
Biomechanics
Cervical spine
Bilateral facet Dislocation
Buckling
Muscle
Initial conditions
Compression
Pre-flexion
Preflexion
Alignment

A B S T R A C T

Cervical spine injuries continue to be a costly societal problem. Future advancements in injury prevention depend on improved physical and computational models, which are predicated on a better understanding of the neck response during dynamic loading. Previous studies have shown that the tolerance of the neck is dependent on its initial position and its buckling behavior. This study uses a computational model to examine three important factors hypothesized to influence the loads experienced by vertebrae in the neck under compressive impact: muscle activation, torso constraints, and pre-flexion angle of the cervical spine. Since cadaver testing is not practical for large scale parametric analyses, these factors were studied using a previously validated computational model. On average, simulations with active muscles had 32% larger compressive forces and 25% larger shear forces—well in excess of what was expected from the muscle forces alone. In the short period of time required for neck injury, constraints on torso motion increased the average neck compression by less than 250 N. The pre-flexion hypothesis was tested by examining pre-flexion angles from neutral (0°) to 64°. Increases in pre-flexion resulted in the largest increases in peak loads and the expression of higher-order buckling modes. Peak force and buckling modality were both very sensitive to pre-flexion angle. These results validate the relevance of prior cadaver models for neck injury and help explain the wide variety of cervical spine fractures that can result from ostensibly similar compressive loadings. They also give insight into the mechanistic differences between burst fractures and lower cervical spine dislocations.

Introduction

Cervical spine injury continues to be a costly social problem. There are an estimated 200,000 people living with spinal cord injury in the United States (Sekhon and Fehlings, 2001), with total annual medical costs estimated between $3 and $6 billion (Bernhard et al., 2005; French et al., 2007). Not included in this estimate is the loss of income and productivity for both the surviving victims and the estimated 30% of victims who die prior to hospitalization. More than 50% of this cost can be attributed to cord injury at the cervical level (French et al., 2007). While there have been dramatic advances in automotive safety systems over the past 30 years, there does not seem to have been a corresponding decrease in the prevalence or incidence of spinal cord injury (Wyndaele and Wyndaele, 2006). Future advancements in injury prevention are predicated on a better understanding of the neck responses to dynamic loading. Because of the inherent limitations of cadaver testing, some of this understanding will necessarily come from advanced physical and computational models.

The human neck, with its seven cervical vertebrae and nearly 100 individual muscles, is one of the most complex mechanical systems in the human body. It provides impressive mobility for the head while protecting vital vascular, neurological, and respiratory pathways. When subjected to compressive impact, the cervical spine acts as a segmented beam column that can express multiple failure modes due to complex buckling kinematics that depend on the initial conditions. This buckling is affected by all the parameters important in stability theory, including mass, rate, geometry, and constraints (Crisco and Panjabi 1992; Harrison et al., 2001; Liu and Dai, 1989; Nightingale et al., 2000; Panjabi et al., 1998). However, there has been no systematic investigation of the effects of muscle activation, torso end-condition, and initial pre-flexion position on cervical spine loads. These effects are important to understand because the cadaver tests that form the basis of our current understanding of cervical spine injury do not have muscles, and the instrumentation and control needed to acquire the biomechanical data disrupts the in-vivo boundary conditions. It has been assumed that muscles do not significantly affect the cervical spine response in compression, but this has not been

* Corresponding author. +1 919 6605451.
E-mail address: rwn@duke.edu (R.W. Nightingale).

http://dx.doi.org/10.1016/j.jbiomech.2016.01.006
0021-9290/

verified. We also know that the initial orientation of the cervical spine is one of the most important factors in determining injury type and risk (Nusholtz et al., 1983, 1981).

The goal of this study was to use a validated computational model to test the following hypotheses regarding compressive impact to the cervical spine:

1. Active neck muscles do not affect cervical spine loads because they cannot react compression.
2. Constraints on torso motion do not affect cervical spine loads because injury happens before there is time to significantly alter the direction of torso momentum,
3. The initial orientation of the cervical spine affects the loads by altering the buckling stability.

These results should provide better insights into the strengths and limitations of previously published cadaver models. This study emphasizes the importance of cervical alignment in determining both the risk for injury and the injury type. It also demonstrates how most of the clinically important cervical spine fractures can be generated with fairly subtle changes in initial conditions.

Methods

The Duke University Head and Neck model was used for these simulations. The model is a 3-D LS-DYNA hybrid finite element/multibody dynamics model composed of rigid vertebrae connected with nonlinear springs and dashpots. It was originally developed and validated for compression by Camacho et al. (1997) and subsequently improved upon (Camacho et al., 1997; Chancey et al., 2003; Dibb, 2011; Van Ee et al., 2000). The latest version is scalable for age and gender and includes 23 pairs of active neck muscles that follow anatomically appropriate paths (Dibb, 2011; Dibb, et al., 2013, 2014). The model consists of eight rigid body vertebrae, from the first cervical (C1) to the first thoracic (T1), with a rigid torso coupled to T1. The skull consists of viscoelastic shell elements based on the NIH VHP (https://www.nlm.nih.gov/research/visible/visible_human.html) with a tuned dynamic absorber within to represent the brain mass (Camacho et al., 2001). A mesh of null elements derived from the LSTC-NCAC Hybrid III male latex skin part forms the exterior surface of the head to represent the 50th percentile male anthropometry. A viscoelastic head is necessary to dissipate kinetic energy and to ensure the appropriate loads are applied to the superior end of the cervical spine. The vertebrae and head are connected by seven joints each consisting of massless, nonlinear 6-DOF springs (LS-DYNA *MAT_GENERAL_NONLINEAR_6DOF_DISCRETE_BEAM) in parallel with linear dampers. The joint centers are located in accordance with the literature (Chancey et al., 2007; Dvorak et al., 1991; Van Mameren et al., 1992)

The baseline simulations used the same impact energy and head and neck orientations as a prior cadaver compressive impact study (Nightingale et al., 1997, 1996a, 1996b). The T1 vertebra of the model was attached to a rigid body model of the Hybrid III upper torso, which was developed by NCAC (LSTC, Livermore, CA). In order to deliver the same energy as in the cadaver experiments, the mass of the torso was set to 16 kg and the velocity was 3.2 m/s. Estimates for the torso center of gravity (CG) location and mass moments of inertia (Cheng et al., 1994) were transformed to the model's coordinate system. The torso CG was 2.95 cm posterior and 33.46 cm inferior to the OC joint. The mass moments of inertia (kg-m^2) were: $I_{xx}=0.4817$, $I_{xy}=0$, $I_{xz}=0.0434$, $I_{yy}=0.3628$, $I_{yz}=0$, and $I_{zz}=0.3184$. The initial position placed the top of the head 1 cm above the plate, with gravity acting downwards. Except where noted, none of the model components were constrained (Fig. 1). The simulations were run for 20 ms because, in the experiments, injury occurred in less than 10 ms and peak axial force in less than 15 ms (Nightingale et al., 1996a, 1996b). Since these simulations do not model failure, any results beyond 20 ms are unlikely to be realistic. All of the simulation data was filtered in accordance with SAE J211 Class 1000.

The baseline simulation was unconstrained and had active muscles. The optimized relaxed muscle activations from Chancey et al. were used (Chancey et al., 2003). This activation scheme is based on the objective of minimizing the energy to hold the head upright under the action of gravity, which reasonably represents an individual with no pre-impact awareness (Chancey et al., 2003; Dibb et al., 2006). In this scheme, no muscles were more than 20% active and many were less than 5%. The muscle optimizations were run in an upright posture, but the simulations were run in an inverted posture. Therefore, the models were allowed to equilibrate for 20 ms prior to the application of the velocity initial condition.

The effects of pre-flexion, torso constraint, and muscles were all evaluated by analyzing changes in the magnitudes of compressive loading, shear loading, and moment on the O-C2 C4-C5, and C7-T1 joints. To study the effect of neck pre-flexion, simulations were run with 32 neck flexion angles ranging from 0° to 64°. To establish the initial orientations, a 2 s flexion simulation was run applying equal and opposite moments to the torso and the head. This approach keeps the neck alignment vertical with respect to the impact surface and velocity vector. Using the results of this simulation, new models with pre-flexed nodal positions were generated at 2° intervals (Fig. 2). To examine the effect of muscle activation, the baseline simulations were compared to simulations without muscles. The effects of the torso constraints were tested in the same manner as muscle activation. The baseline simulations were compared with a series where the torso was constrained to linear motion along the *Z*-axis in same manner as it was in the cadaver experiments (Nightingale et al., 1996a, 1996b). A total of 96 simulations were run.

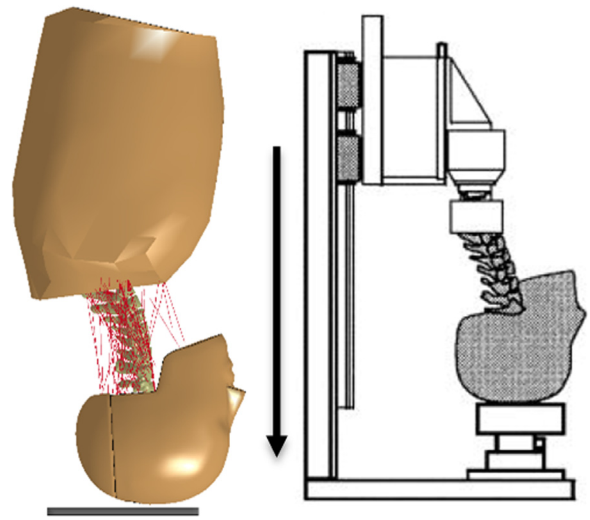

Fig. 1. Model used in simulations, shown with muscles present and an arrow pointing in the direction of gravity and initial velocity.

Results

As would be expected in compressive column impact, the cervical spine kinetics were complex. In the baseline model, the joint loads were dependent on the vertebral level, the constraint condition, and muscle condition; but they were most strongly affected by the pre-flexion angle. The compressive forces were of similar magnitude at each vertebral level (Fig. 3, Row 1), but the moments and shear forces were quite different (Fig. 3, Rows 2 and 3).

Constraining the torso had a relatively minor effect on the loads experienced by the cervical spine (Fig. 4 Columns 2 and 3, Video 1). The constrained and unconstrained cases showed similar trends and magnitudes in the compressive force, the shear force, and the moment experienced at C7-T1 (Fig. 4). This was also seen in the O-C2 and C4-C5 joints. In general, there was a slight increase in the magnitude of most loads at most levels. For compression across all vertebral levels and pre-flexion angles, there was a mean increase of 234 ± 190 N in the peak compressive force (Fig. 5, Column 2) with the constrained torso. However, constraint caused a larger increase (near 700 N) in peak compressive load at impact angles near 30°. For shear, the mean increase was 115 ± 175 N (Fig. 6, Column 2); for moment, it was 6 ± 20 N-m (Fig. 7, Column 1). The small effect and large standard deviations for shear and moment are due to the fact that the effect was bi-polar.

Supplementary material related to this article can be found online at http://dx.doi.org/10.1016/j.jbiomech.2016.01.006.

Muscles had an effect on both the magnitude and timing of peak loads in the cervical spine (Fig. 4 Columns 1 and 2, Video 2). The simulations with muscles reached higher peaks in compression and shear and maintained high levels of force longer than the model without muscles. On average, the simulations with muscle had 32% larger peak compressive forces (771 ± 738 N) across all flexion angles; however, this difference was less pronounced for pre-flexion angles greater than 25° (Fig. 5, Row 1). For example,