

Assessing Cervical Spine Response to Head-First Impact Using Vertebral Segments, Head-Neck, and
Full-Body Computational Human Models

by

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Author's Declaration

This thesis consists of material all of which I authored or co-authored: see Statement of Contributions included in the thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Statement of Contributions

Maxwell Morgan was the sole author for Studies 1 and 3, which were written under the supervision of Dr. Duane Cronin and have not been published.

This thesis consists of part of a manuscript written for publication. Exceptions to sole authorship of material are as follows:

Research presented in Study 2:

The research in Study 2 was conducted at the University of Waterloo by Maxwell Morgan under the supervision of Dr. Duane Cronin. A manuscript has been submitted to Annals of Biomedical Engineering and is under review.

Maxwell Morgan was responsible for developing, running and often troubleshooting the finite element models, processing the data, writing, and editing of the manuscript.

Dr. Duane Cronin was the principal investigator and supervisor, providing guidance on the research scope, data analysis and editing of the manuscript. Miguel Corrales was responsible for the repositioning in Study 2 and assisted with editing of the manuscript, Harsharan Kaur helped with the development of the model of the boundary conditions, and Dr. Peter Crompton provided an advisory role throughout the project.

Abstract

Head-first impact (HFI), which can occur in automotive rollovers and sports collisions, is associated with a high risk of cervical spine injury. Cervical spine injuries from HFI such as fracture-dislocations frequently lead to severe spinal cord injuries and in some cases death, as reported in field data and epidemiology. Experimentally, isolated motion segments have been tested in compression and bending to mimic loads incurred in HFI, while cadaveric head and necks with torso surrogate masses (TSMs) and full body (FB) cadavers have been inverted and dropped to investigate HFI. However, isolated segment tests are limited in producing the complex kinematics of HFI, and TSM response has not been quantified with respect to FB testing. Recently, computational human body models (HBMs) have been developed to simulate humans in injurious loading conditions, but have only seen limited application in HFI.

In this study, computational models were applied to investigate HFI, using an isolated vertebral segment model, an isolated head and neck with a TSM, and a contemporary FB human model. First, an existing and validated cervical motion segment model was loaded in combined compression and flexion relevant to HFI, to investigate the loads and moments associated with fracture-dislocation failures. Next, TSM and FB head-first impacts were modelled using a contemporary HBM in three postures (flexed, neutral, extended) at three impact velocities. Finally, the FB model was compared with a unique set of experimental full body cadaver HFI tests.

In isolated segment loading, combined compression and flexion produced hard tissue failure patterns reported in fracture-dislocations. Fracture-dislocation was achieved by simultaneously rotating and translating the superior vertebra anteriorly. Comparing the isolated head and neck TSM and FB models, the TSM condition demonstrated higher neck forces, internal energy, and a larger volume of hard tissue failure compared to the FB models under the same impact conditions. Despite similar head contact forces between TSM and FB, the compliant thorax of the FB model reduced the neck forces by half,

which significantly reduced corresponding energy stored in the neck tissues. The neutral and extended neck postures predicted higher neck forces due to facet joints engaging, while neck flexion in the flexed posture reduced neck forces by misaligning the spine from the impact. Finally, it was found that the FB model had similar head impact forces and comparable T1 rotation to cadaveric HFI experiments, but a high sensitivity to initial posture was identified.

This study identified forces and moments that can create a fracture-dislocation in a motion segment using prescribed boundary conditions. The TSM and FB simulations demonstrated compression loads and moments of a similar magnitude to the motion segment, but differed in timing, generating higher axial loads leading to the onset of fracture in the spine. The neck loads were higher using the TSM boundary condition compared to the FB condition. Both TSM and FB models identified the importance of neck posture on response, showing that an initially extended neck posture leads to higher neck forces compared to a flexed posture. This study identified the importance of full body boundary conditions for the simulation of HFI, the complex dependency of kinematic and kinetic response on neck posture, providing model results in agreement with the small number of full body experiments. Further experimentation was recommended to provide detailed measurements necessary for model assessment and validation. Future computational studies will integrate the motion segment and FB results to improve the understanding of fracture-dislocation.

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List of Abbreviations

DOF: degrees-of-freedom

FB: full body

FB_{EXT}: a simulation with a FB boundary condition and an extended neck

FB_{FLX}: a simulation with a FB boundary condition and a flexed neck

FB_{NEUT}: a simulation with either a FB boundary condition and a neutral posture

FE: finite element

GHBMC: the Global Human Body Models Consortium

GHBMC M50: 50th percentile male Global Human Body Models Consortium model

HBM: human body model

HFI: head-first impact

PMHS: postmortem human subject

PVC: Polyvinyl Chloride

TSM: torso surrogate mass

TSM_{EXT}: a simulation with a TSM boundary condition and an extended neck

TSM_{FLX}: a simulation with a TSM boundary condition and a flexed neck

TSM_{NEUT}: a simulation with a TSM boundary condition and a neutral neck

List of Symbols

ϵ	strain
S	displacement
l_0	foam initial thickness
F	force
A_{approx}	approximate area
a	acceleration

Chapter 1

Introduction

1.1 Research Motivation

Head-first impact (HFI) has been linked to severe injuries associated with automotive rollovers (Viano, Parenteau, Gopal, & James, 2009; Al-Salehi, et al., 2024) and sports injuries (McElhaney, Snyder, States, & Gabrielsen, 1979; Torg, Vegso, O'Neill, & Sennett, 1990), with rollovers being the leading cause of neck injury in automotive crashes (Yoganandan, et al., 1989; Fakharian, et al., 2015). In rollovers where the occupant head interacts with the vehicle roof, the lower cervical spine (C5-C6-C7) is injured most frequently (Bambach, Grzebieta, McIntosh, & Mattos, 2013), with fractures in the C5 and C6 pedicles and facets, as well as C7 transverse process being the most common. Such injuries can be severe, leading to severe spinal cord injuries or fatalities (Yoganandan, et al., 1989). Moreover, fracture-dislocation (the anterior dislocation of a superior vertebrae relative to an inferior vertebrae and a concurrent vertebral fracture) is the most common cause of spinal cord injury (Sekhon & Fehlings, 2001), and is caused most frequently by automotive rollovers (Parenteau & Viano, 2014; Quarrington, et al., 2018) compared to other automotive collisions. Head-first impact is the suspected mechanism of neck injury in automotive rollovers, as peak neck loads in experimental rollovers with a crash test dummy occur when the head is in contact with the roof (Viano, Parenteau, Gopal, & James, 2009).

To understand injuries associated with head-first impact including fracture-dislocations, cervical segments with 2-3 vertebrae have been dissected and loaded in compression and bending (Camacho, et al., 1997; Nightingale, et al., 2002; Wheeldon, Pintar, Knowles, & Yoganandan, 2006). These studies typically do not load until failure, and ones that do (Nightingale, et al., 2007; Carter, Ku, Nuckley, & Ching, 2002) have limited combined loading trials despite HFI being expected to include combined loads to the cervical spine (Nightingale, Sganga, Cutcliffe, & Bass, 2016). Experimentalists (Ivancic, et al., 2008) reliably reproduced bilateral facet dislocations (without fracture) by applying acceleration to a motion segment of two vertebrae that had a mass and spring apparatus mounted to it.

Full body (FB) post-mortem human subjects have been subjected to experimental HFI to study associated injury mechanisms (Culver, Bender, & Melvin, 1978; Nusholtz, Huelke, Lux, Alem, & Montalvo, 1983; Yoganandan, et al., 1986), finding evidence that the initial posture and impact velocity affected the severity and location of cervical spine fractures. A more recent short communication of a

FB experimental HFI study (Kerrigan, et al., 2014) found that injury was less likely to occur at 3.14 m/s than was predicted in TSM studies, but the study was limited to four specimens. However, FB experiments are challenging, in part due to difficulties in achieving consistent initial conditions such as posture, as well as the cost of using a full body. To simplify the boundary conditions and reduce variability, many HFI experiments have used a torso surrogate mass (TSM) to replicate the constraining effect and loads produced by the torso (Nightingale, et al., 1997; Toomey, et al., 2009; Saari, Itshayek, & Cripton, 2011). Similar to FB studies, the TSM cadaveric head-first impact studies reproduced clinically relevant injuries, but despite the uniform T1 boundary conditions, fractures were highly variable between specimens (Nightingale, et al., 1997; Yoganandan, Banerjee, DeVogel, Pintar, & Baisden, 2018). The initial velocity used in many experiments (3.14 m/s) was based on onset of neck injury in diving injury reconstructions (McElhaney, Snyder, States, & Gabrielsen, 1979), and is on the low end of velocities from head-on rugby tackles (5.0 ± 2.0 m/s for ball-carriers, and 5.7 ± 2.1 m/s for tacklers) (Hendricks, Karpul, & Lambert, 2014).

Finite element (FE) models can measure the effect of changing parameters on the mechanical response computationally. In particular, FE Human Body Models (HBMs) have been used to investigate the mechanism of injury under HFI. A FB HBM (THUMS v4.01 occupant seated model) was used to model the effect of: impactor velocity, displacement, and direction in HFI (Mattos, McIntosh, Grzebieta, Yoganandan, & Pintar, 2015), reporting that the direction of the impact influenced the location of predicted injury. This study (Mattos, McIntosh, Grzebieta, Yoganandan, & Pintar, 2015) only evaluated a nominal seated position, while posture during HFI has been found to vary (Newell, Blouin, Street, Cripton, & Siegmund, 2013; Newell, Seigmund, Blouin, Street, & Cripton, 2014). Numerical HFI studies have also applied the TSM boundary condition, allowing for validation against TSM cadaveric HFI studies. A variety of neck models have been developed and validated in TSM HFI (Yang, Zhu, Luan, Zhao, & Begeman, 1998; Hu, Yang, Chou, & King, 2008; Nightingale, Sganga, Cutcliffe, & Bass, 2016; Nasim, Cernicchi, & Galvanetto, 2021; Darweesh M. , 2022), but none have compared the TSM boundary condition to a FB boundary condition in HFI.

The Global Human Body Models Consortium (GHBMC) 50th percentile male detailed human body model (M50) was developed using MRI and CT scans from a 26-year-old adult with mid-size male anthropometrics (Gayzik, et al., 2011). The neck region of the GHBMC M50 model includes detailed representations of hard and soft tissue in the human body, using material properties obtained experimentally from: bones, ligaments, and passive muscles, which have been validated in a

hierarchical manner against cadaveric and volunteer data (Barker & Cronin, 2021). The neck of GHBM M50, Version 5.0 was assessed against experimental head-first impact loading (Darweesh, Corrales, Cripton, & Cronin, 2020), demonstrating comparable axial forces, fracture pattern and onset of fracture to experimental results (Nightingale, et al., 1997), despite uncertainties around modelling the surrogate head used in the experiment. Version 6.0 of the GHBM M50 includes the assessed neck of Version 5.0 with enhancements to the model improving biofidelity, including remeshing problematic parts and enhancing the upper cervical spine ligament properties to more closely match experimental results (Hadagali & Cronin, 2023). Re-validation of Version 6.0 in HFI was not pursued in this study due to uncertainties around experimental boundary conditions and the similarity of the neck models.

To study postural effects on HBMs, reposturing is often used (Nightingale, Sganga, Cutcliffe, & Bass, 2016; Correia, Corrales, McLachlin, & Cronin, 2023; Corrales & Cronin, 2021). However, HBMs are challenging to reposition while maintaining the FE mesh quality (Janak, Lafon, Petit, & Beillas, 2018). An open-source repositioning software (PIPER), developed to reposition and morph detailed HBM (Beillas, et al., 2015) allows researchers to precisely reposition detailed FE models while retaining mesh quality (Janak, Lafon, Petit, & Beillas, 2018). However, the effect of posture has yet to be investigated using full body human body models under HFI. Reposturing a model requires quantification of vertebral positions in the neck, which has been done for varying postures in the form of a two-dimensional geometrical Cervical Spine Posture Predictor model (Reed & Jones, 2017). The Cervical Spine Posture Predictor predicts the vertebral positions based on head angle, sex, age, stature and seated stature. A total of 177 seated subjects from 18- to 74-year-old were radiographed in a neutral posture, maximum extension, and maximum flexion (Snyder, Chaffin, & Foust, 1975) and digitized (Klinich, et al., 2004) to serve as the database of the Cervical Spine Posture Predictor.

This study addresses limitations described in Section 1.1, which include: the limited data and therefore understanding of combined compression and flexion failure, the limitation that TSM and FB boundary conditions have not been directly compared in HFI, and the limitation that FB HBMs have not been compared with FB cadaveric HFI experiments.

1.2 Research Aim and Objectives

The aim of this research was to model head-first impact, quantify the forces and kinematics of the head and neck, and measure the effect of using different boundary conditions and postures. To achieve the aim, the work was divided into three studies, organized in terms of geometric complexity, using isolated

vertebral segments subjected to combined compression-flexion loading, an isolated head and neck model in HFI conditions, and a full body model in HFI conditions. The specific objectives of each study are listed below, and described in further detail.

- Study 1 objective: to determine the forces and moments that lead to vertebral fracture-dislocation injuries in the cervical spine using vertebral motion segment models.
- Study 2 objective: to compare TSM and FB responses in HFI for 3 neck postures using a human body model.
- Study 3 objective: to compare the response of the FB human model to experimental full body postmortem human subject (PMHS) HFI results.

The objective of Study 1 was to determine the forces and moments that lead to vertebral fracture-dislocation injuries in the cervical spine using vertebral motion segment models. Five motion segment models were isolated from Version 6.0 of the GHBMCM50 full body model, subjected to combined compression and flexion loading, and the failure loads were measured. To stabilize the simulations after the failure of ligaments and bone from the combined compression and flexion, motion segments were then subjected to displacement-based control to obtain fracture-dislocation (Fig. 1). The outcome of Study 1 was fracture-dislocations at the motion segment level with associated forces and moments quantified.

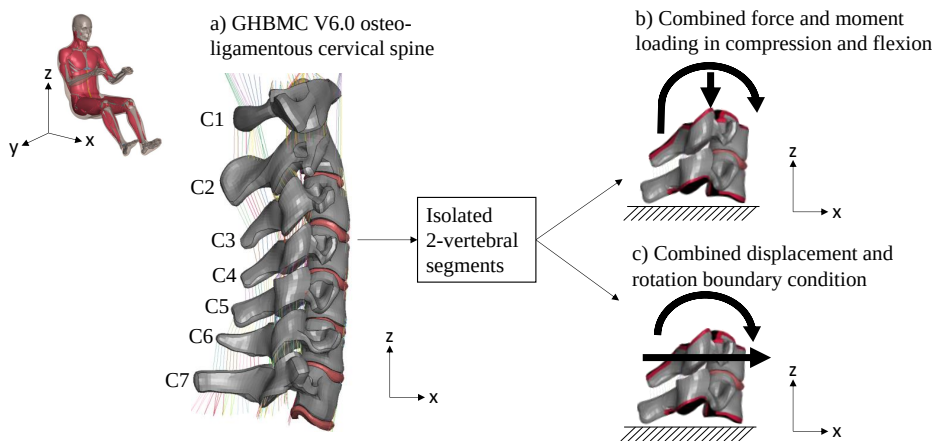


Figure 1: Overview of Study 1 including a) the GHBMCM50 version 6.0 osteo-ligamentous spine, wherein five motion segments composed of 2 vertebrae were isolated, b) combined force and moment loading in compression and flexion, and c) combined displacement and rotation boundary condition.

The Study 2 objective was to compare TSM and FB responses in HFI for 3 neck postures using a human body model. To obtain a TSM model, the head and neck of Version 6.0 of the GHBMCM50 human body model was positioned in flexed (FLX), neutral (NEUT), and extended (EXT) neck postures with a constrained 16 kg mass attached to T1, to match the setup used by (Nightingale, et al., 1997). The FB model was an unconstrained, full GHBMCM50 human body model (also Version 6.0) in FLX, NEUT, and EXT postures. Both FB and TSM models in all three postures were inverted and dropped onto a rigid plate (Fig. 2). The models were assessed using head contact force, intervertebral kinematics and kinetics in the lower cervical spine, and the effect of enabling hard tissue failure in the models was quantified. The outcome of this study was demonstrating the difference in the response of TSM and FB boundary conditions, and quantifying the effect of initial posture on HFI.

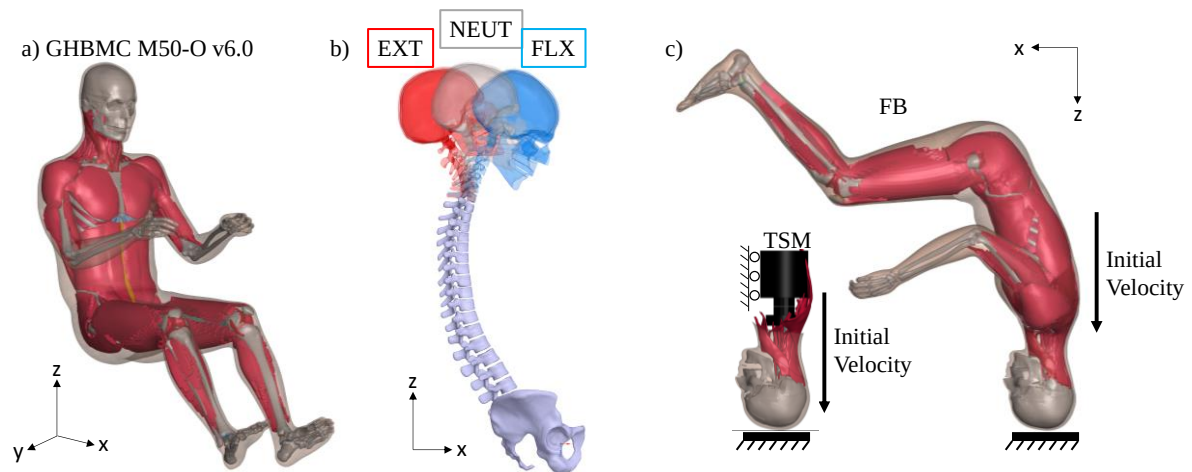


Figure 2: Overview of Study 2 including: a) the GHBMCM50 Version 6.0 model, b) the three cervical spine postures: EXT (extended neck), NEUT (neutral neck), and FLX (flexed neck), and c) the TSM (torso surrogate mass) and the FB (full body) head-first impact boundary conditions.

The objective of Study 3 was to compare the response of the FB human model to experimental full body postmortem human subject (PMHS) HFI results obtained by (Kerrigan, et al., 2014). Study 3 involved modelling the foam impact surface used in the experiment, reposturing the FB model to be similar to the PMHS, and comparing the results with the experimental results (Fig. 3). The model and

experiments were compared using head contact force and the first thoracic vertebral rotation. The outcome of this study was demonstrating that first thoracic vertebra motion, which affected the response of the cervical spine was comparable in the HBM and PMHS tests.

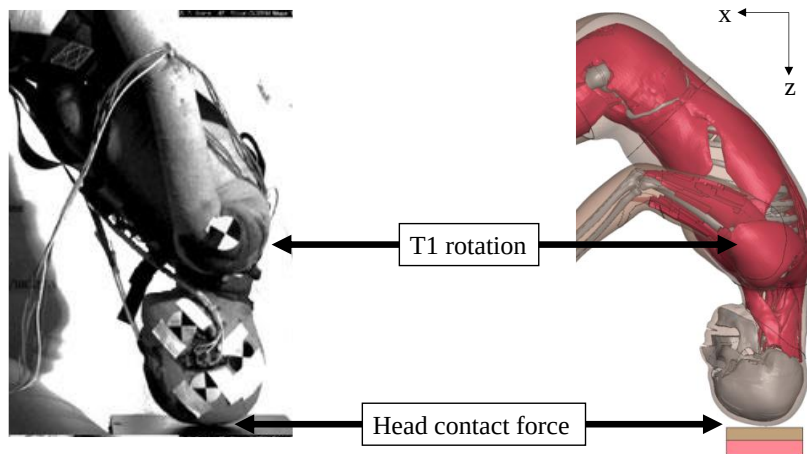


Figure 3: Overview of Study 3 including the experimental results from (Kerrigan, et al., 2014) (left), and the repostured GHBM M50 (right), with measurements used for comparison labelled.

1.3 Thesis Organization

The current (first) chapter contains an introduction to the thesis, including the motivation for the research, the aims of the research, and the organization of the document. The second chapter provides the background information necessary to understand the work undertaken in this thesis, including relevant anatomical information, injuries which occur from head-first impact, and previous head-first impact studies (experimental and computational).

All subsequent chapters are divided between the three studies in this thesis, listed in Section 1.2. The third chapter describes the methods undertaken in the three studies. The fourth chapter describes results obtained from these studies, and the fifth chapter discusses and provides context to the results. The discussion section includes relating the numerical results to experimental results and explaining differences between them. Additionally in the discussion, the trends in the results are identified explained. The sixth chapter provides conclusions and recommendations from the three studies.

Chapter 2

Background

2.1 Cervical and Upper Thoracic Spine Anatomy

2.1.1 Planes and Directions in Human Anatomy

To denote directions and locations in the human body, standard terms are typically used. For a human standing upright, the upward direction (towards the head) is called superior, and the downward direction (towards the feet) is called inferior. The forward direction (the direction of the eyes) is denoted as anterior, and the rearward direction posterior/rear. In this document, the left and right directions will be denoted as left and right. The plane perpendicular to the superior-inferior direction is the transverse/horizontal plane, the plane perpendicular to the anterior-posterior direction is the frontal/coronal plane, and the plane perpendicular to the left-right direction is the sagittal plane. Axes (transverse, frontal, and sagittal) are perpendicular to their respective planes. Additionally, the lateral direction is perpendicular to the sagittal plane, away from the mid-sagittal plane of the body, and the medial direction is perpendicular to the sagittal plane, towards the mid-sagittal plane. The axes of the human body are shown (Fig. 4)

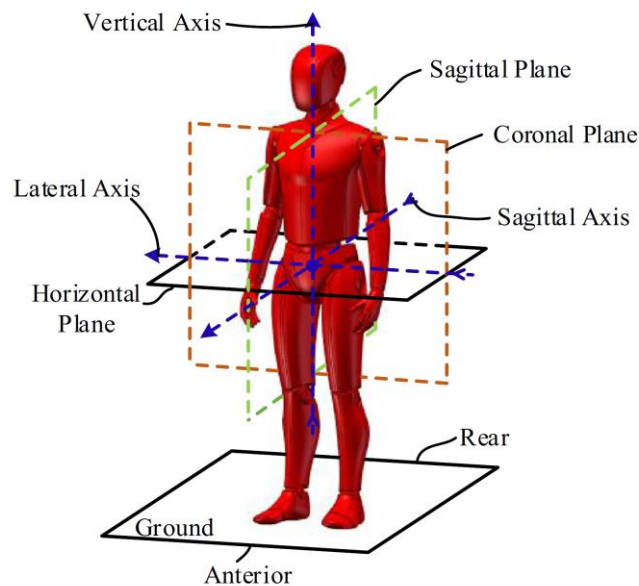


Figure 4: Axes and directions of the human body obtained from (Song, Guo, Wang, & Qu, 2020, open access).

2.1.2 Cervical Spine Overview

The human spine is a structure that can be divided into the cervical, thoracic, lumbar, sacral and coccygeal spine and includes 33 vertebrae. The cervical spine connects the head to the thoracic spine, and is composed of: seven vertebrae (Fig. 1a), connective tissues between the vertebrae, a spinal cord, muscles, and tendons. The cervical spine provides structural stability, protecting the spinal cord and (working with the neck muscles) maintaining an upright posture for the head and neck, while being flexible to enough to move the head in three rotational degrees of freedom (Sénégas, 2020).

Cervical vertebrae are typically listed in descending order moving from the most superior (C1) to the most inferior (C7), and can be divided into atlanto-axial (C1-C2) and subaxial (C3-C7). Additionally, the thoracic vertebrae are denoted T1-T12, with T1 being directly inferior to C7. The 7 cervical vertebrae and their connective tissues each provide six degrees of freedom (three rotational and three translational degrees of freedom) (Vital, Sénégas, Garnier, & Bouloussa, 2020) making the spinal column kinematically over-defined. The cervical spine can therefore rotate the head in 3-degrees of freedom: a) about the transverse plane (15° left or right) b) about the frontal plane (15° - 30° left or right), and c) about the sagittal plane (70° anteriorly, 80° posteriorly) (Vital, Sénégas, Garnier, & Bouloussa, 2020).

All vertebrae are composed of bone tissue, which contains porous (trabecular) bone, surrounded by lower porosity (cortical) bone (Louis, 1983) (Fig. 5). The trabecular bone reduces the weight of the bone, while the cortical bone provides structural strength.

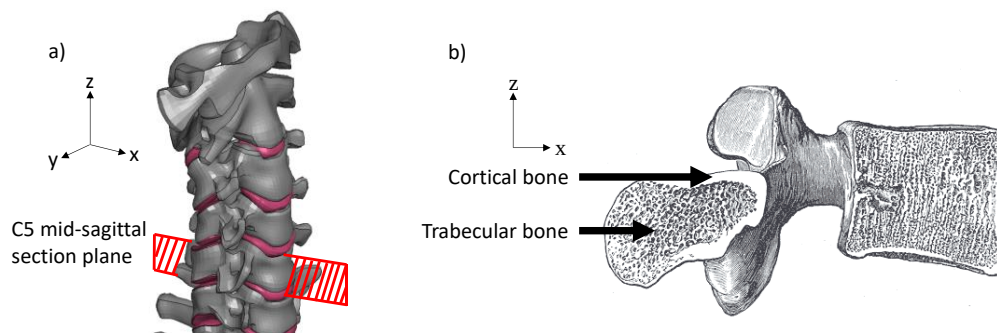


Figure 5: Vertebral bone including a) a diagram of a C5 mid-sagittal section and b) a mid-sagittal section of a vertebra showing trabecular and cortical bone obtained from (Gray, 1918, public domain), denoting porous trabecular bone and less porous cortical bone.

2.1.3 Subaxial (Lower) Cervical Spine Anatomy

Subaxial vertebrae (C3-C7) are responsible for lateral flexion (C3-C5), and sagittal plane flexion-extension (C4-C6) (Vital, Sénégas, Garnier, & Bouloussa, 2020) of the head. The vertebrae contain a vertebral body, located on the anterior side of the vertebral foramen. Posterior and lateral to the vertebral body are the transverse processes, which are composed of two tubercles. Posterior to the transverse foramen are the lateral masses, which on the superior and inferior sides connect to facet joints of adjacent vertebral bodies. The lateral masses are connected to the vertebral body by the transverse processes and pedicles, creating an opening called the transverse foramen, which the vertebral vein and artery pass through. Laminae connect the lateral masses to the spinous process, which is the posterior most part of the vertebrae, and are divided into two lobes by a median cleft. Between the lamina, pedicles, and vertebral body is the spinal canal, which holds the spinal cord (Standring & Gray, 2008) (Fig. 6).

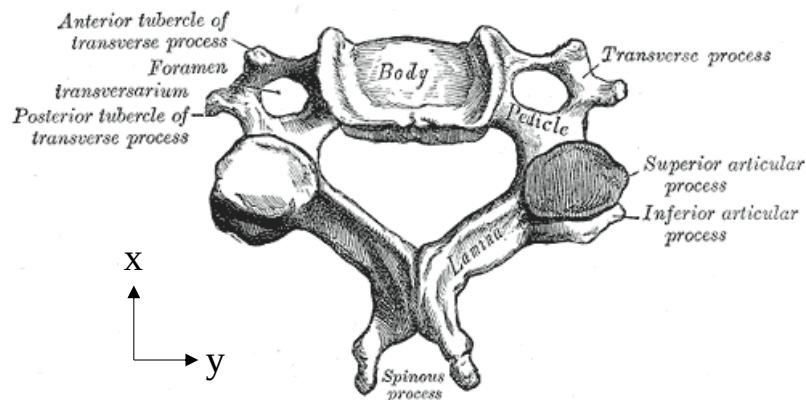


Figure 6: Superior view of a subaxial cervical vertebra, obtained from (Gray, 1918, public domain), with the addition of the coordinate system used in this paper.

2.1.4 Atlanto-Axial Vertebrae

The atlanto-axial joint (C1-C2) is responsible for head rotation about the transverse axis, whereby the atlas (C1) rotates about the axis (C2) (Vital, Sénégas, Garnier, & Bouloussa, 2020). The atlas (C1) is a ring-shaped vertebra containing two lateral masses but no vertebral body. The lateral masses have facet joints connecting them to the skull occipital condyles, and the facet bones of the axis (C2). Lateral of

the lateral masses are transverse foramen and transverse processes. The axis contains the same features as the subaxial vertebrae, with an additional conic process called the odontoid or dens. The odontoid, superior to the axis vertebral body, sits between the atlas lateral masses, and is connected to a facet joint on the inside of the atlas anterior arch (Fig. 7) (Standring & Gray, 2008).

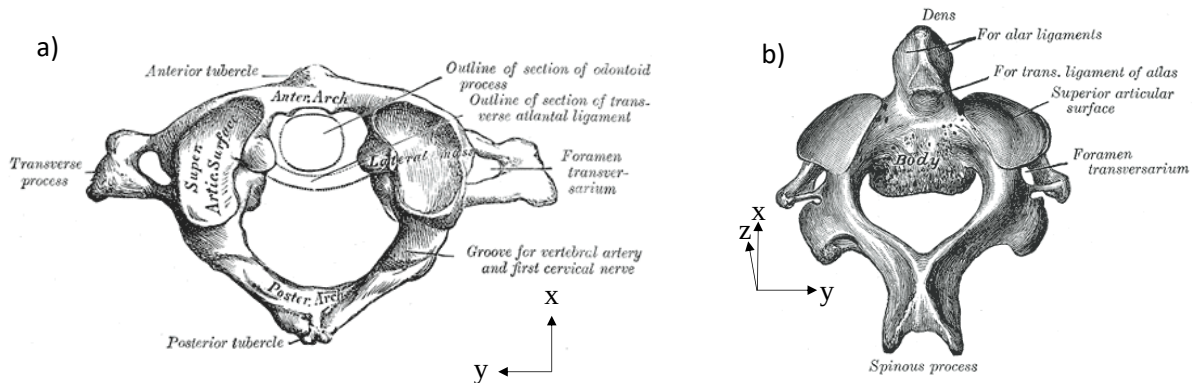


Figure 7: Axis (C2) a) inferior view and b) superior-posterior view, obtained from (Gray, 1918, public domain), with the addition of the coordinate system used in this paper.

2.1.5 Cervical Spine Joints and Ligaments

Intervertebral discs are the principle load bearing and articulating joints between vertebrae from C2 to T1, located between vertebral bodies, containing a nucleus, annulus fibrosus, and cartilaginous endplates (Lavignolle, 2020) (Fig. 8). The nucleus is a gel-like structure in the middle of the disc made up of mostly water. The annulus fibrosus consists of layers of collagen fibers surrounding the nucleus. The cartilaginous endplates are plates of cartilage that connect the superior and inferior vertebral bodies to the annulus fibrosus and nucleus (Louis, 1983). Movement in the spine (tilting and shearing) is limited by the intervertebral discs (Standring & Gray, 2008).

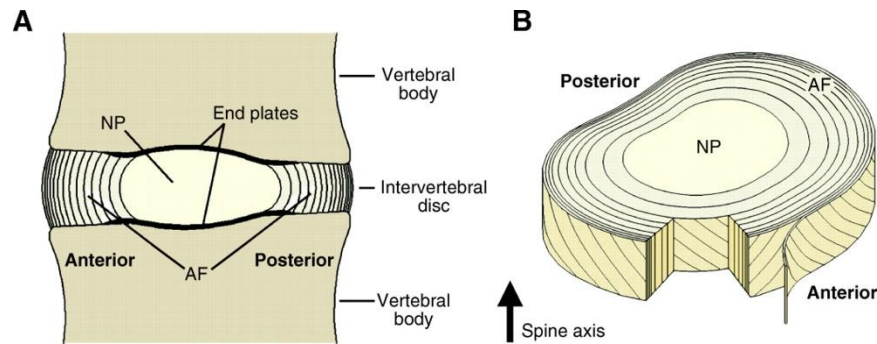


Figure 8: Intervertebral disc a) frontal section and b) isolated superior view with the nucleus pulposus (NP), annulus fibrosus (AF), and end plates indicated, obtained from (Smith, Nerurkar, Choi, Harfe, & Elliott, 2011, open source).

Another load bearing structure responsible for vertebral motion is the facet joint, which is located between superior and inferior articular surfaces for the lateral masses. Facet joints comprise of a capsular ligament, which envelops a synovial fluid, acting as a lubricant between cartilaginous endplates, which lie on superior and inferior articular surfaces (Fig. 9) (Standring & Gray, 2008).

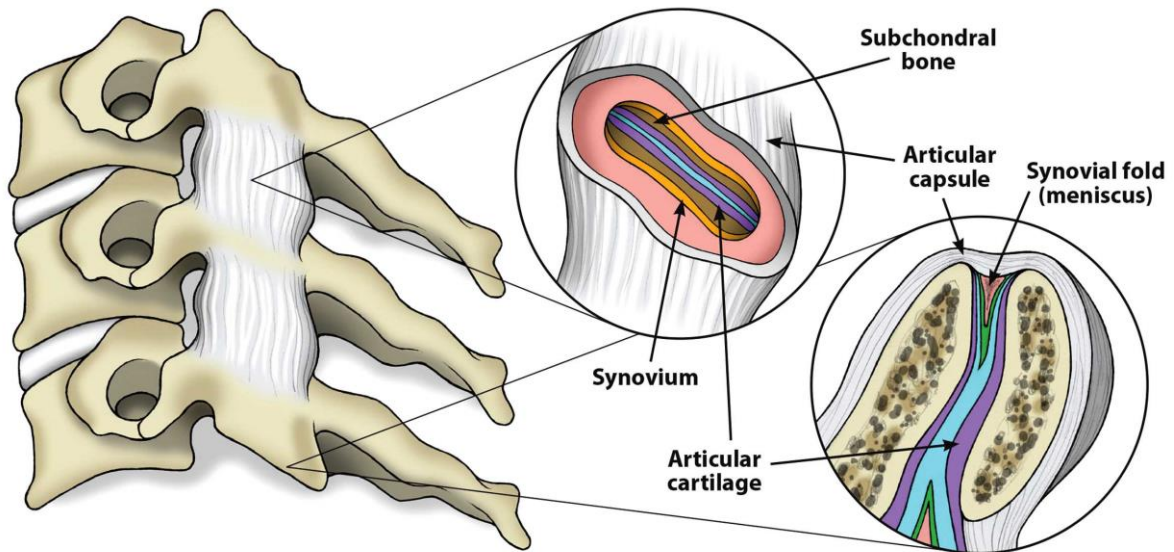


Figure 9: Facet joint, showing a capsular ligament (articular capsule) enveloping cartilaginous endplates as well as synovial fluid, from (Steilen, Hauser, Woldin, & Sawyer, 2014, open source).

The anterior and posterior longitudinal ligament are bands which run through the vertebral column, covering the anterior and posterior of the vertebral bodies and intervertebral discs. Running down the posterior side of the spinal canal is the ligamentum flavum. Connecting adjacent spinous processes are interspinous ligaments (Fig. 10) (Standring & Gray, 2008).

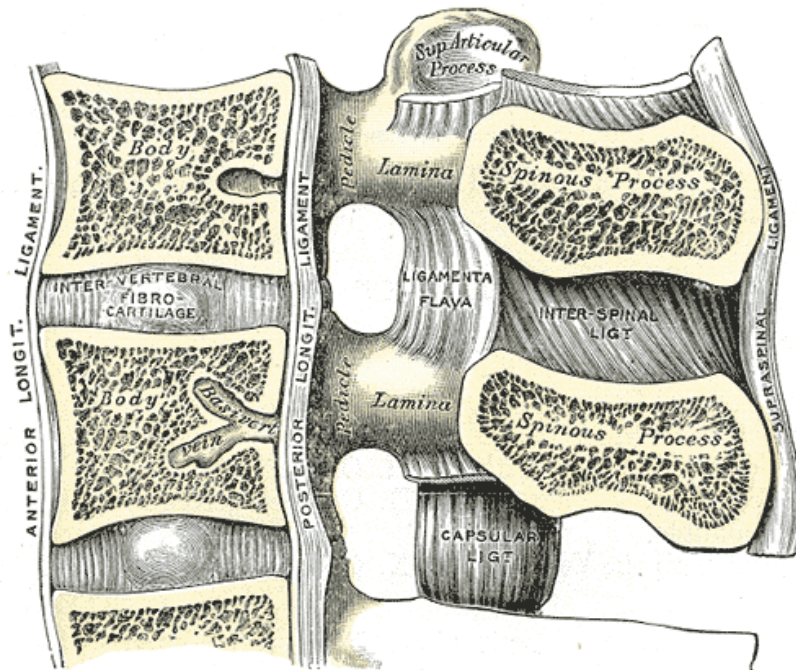


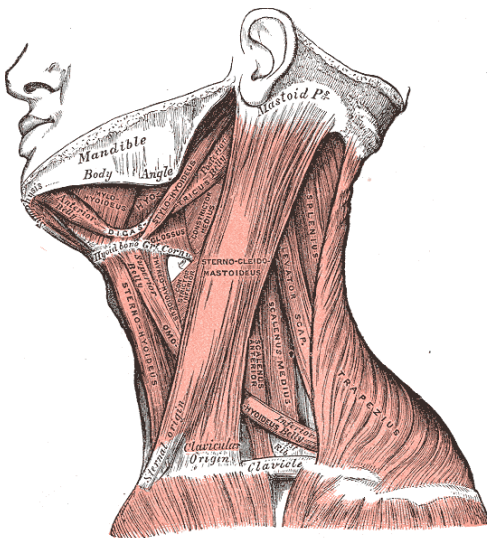
Figure 10: Mid-sagittal section of sub-axial vertebrae, with connecting ligaments indicated, including: anterior and posterior longitudinal ligaments, ligamentum flavum, interspinous ligaments, supraspinal ligament, capsular ligament, and intervertebral discs, obtained from (Gray, 1918, public domain).

2.1.6 Cervical Spine Skeletal Muscles

Skeletal muscles in the cervical are tissues that stabilize and control the articulation of the head and cervical spine. The skeletal muscles in the cervical spine are connected to skeletal bones with tendons, through which they apply a contractile force. There are 26 pairs of cervical skeletal muscles (Fig. 11), which can be divided into flexors and extensors, for their role in moving the head in flexion or extension (Dimeglio, 2020). For example, the sterno-hyoid muscle connects the sternum to the hyoid bone and

lies anterior to the cervical spine and is denoted a flexor, whereas the splenius capitis connects the posterior-lateral C4-C6 to the posterior-inferior occipital bone (posterior to the cervical spine) and is an extensor.

a) Lateral View



b) Anterior View

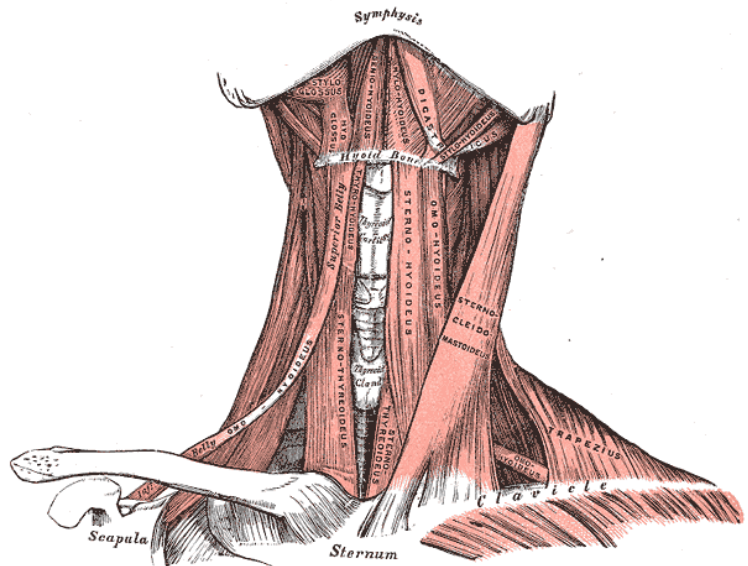


Figure 11: Neck musculature from a a) lateral view, and b) anterior view, obtained from (Gray, 1918, public domain)

Skeletal muscles have active and passive mechanical properties. Passive skeletal neck muscles have been observed to be viscoelastic (Fung, 2013) when mechanically characterized; meaning the muscles exhibit an elastic stress and strain relationship, as well as a viscous strain rate and stress relationship. To perform their functions of stabilization and locomotion, muscles change the amount of tensile force they provide through electrical signals, sent by the nervous system. The changing of tensile force allows for an imbalance of opposing muscles forces, resulting in motion and/or stabilization of skeletal tissues (Dimeglio, 2020).

2.1.7 Spinal Cord Anatomy and Common Spinal Cord Injuries

The spinal cord is a structure that runs from the medulla oblongata of the brain, through the vertebral canals of the cervical spine, to the upper surface of the coccyx at the base of the spinal column. The spinal cord sends signals (sensory and motor) between receptors in the brain and muscles, and is

responsible for regulating a variety of involuntary processes and reflexes in the body (Standring & Gray, 2008).

Spinal cord injury is often fatal, can cause paraplegia and quadriplegia, and has significant socioeconomic costs (Mahabaleshwarkar & Khanna, 2014). Injury to the spinal cord occurs most commonly in the cervical spine, and is most frequently a fracture-dislocation or a burst fracture (40% and 30% of spinal cord injuries, respectively) (Sekhon & Fehlings, 2001).

Cervical fracture-dislocations involve the anterior translation of a vertebra (most commonly C5-C7) relative to the neighboring inferior vertebra, where one (unilateral) or both (bilateral) facets become dislocated and sometimes lock in place. Usually there is a fracture associated with the dislocation, either in the vertebral body or the facets. The relative motion of the vertebrae often results in damage or disruption to the spinal cord, leading to injury (Nightingale, Myers, & Yoganandan, Neck Injury Biomechanics, 2015) (Fig. 12).

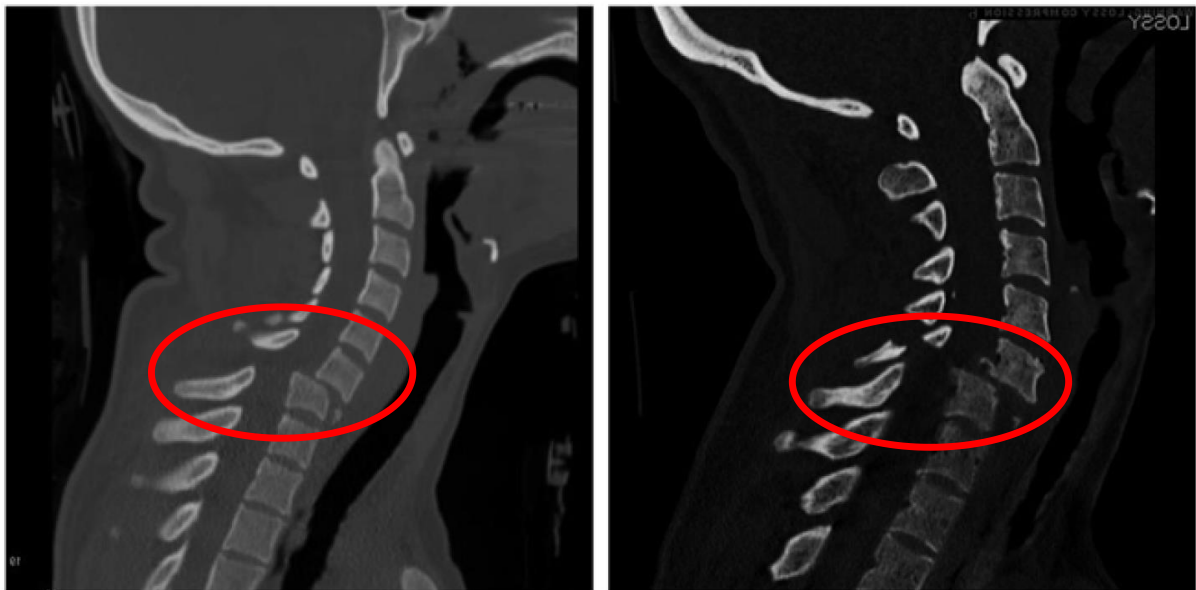


Figure 12: Mid-sagittal view of bilateral facet fracture-dislocations from (Nightingale, Myers, & Yoganandan, Neck Injury Biomechanics, 2015, used with permission from Springer). Red circles highlighting the dislocated facets have been added, showing a dislocation at C6-C7 in both images.

Burst fractures result from the crushing and fracturing of the vertebral body of a vertebra, and occur at all levels from C3-T1, but most commonly at C4, C5, and C6 (McElhaney, Snyder, States, & Gabrielsen, 1979). Burst fractures present a high risk of spinal cord injury (Kiwinski, 1991), due to fragments of the crushed vertebral body moving into the spinal canal.

Researchers (Allen, Ferguson, Lehmann, & O'Brien, 1982) created a mechanistic classification system for lower-cervical spine fractures and dislocations by linking the injury pattern to the reported circumstances of injury. The patterns of injuries were sub-divided into numbered stages, with higher numbers indicating increased severity and progression of the injury.

The most commonly occurring injury pattern (37% of injuries) was distractive flexion, which involved an automotive collision 57% of the time. Distractive flexion injuries include tearing of the posterior ligamentous complex (stage 1), and the dislocation of one (stage 2) or both (stage 3-4) superior facets, anterior to the inferior facet, and locking in place (Fig. 13). Additionally, Allen et al. (Allen, Ferguson, Lehmann, & O'Brien, 1982) reported a “blunting of the anterior superior vertebral margin to a rounded contour” often occurring, which was a fracture of the anterior-superior vertebral body of the inferior vertebra. The other common injury patterns were compressive extension (24% of injuries) and compressive flexion (22% of injuries). Compressive extension involves fracture of the vertebral arch, which may include a fracture in the: articular process, pedicle and/or lamina. Highly injurious cases of compressive extension (Stage 5 compressive extension) involve the superior vertebra translating a full vertebral body width anteriorly. Compressive flexion includes the same “blunting of the anterior superior vertebral margin” as distractive flexion, without the posterior ligament tears (Stage 1).

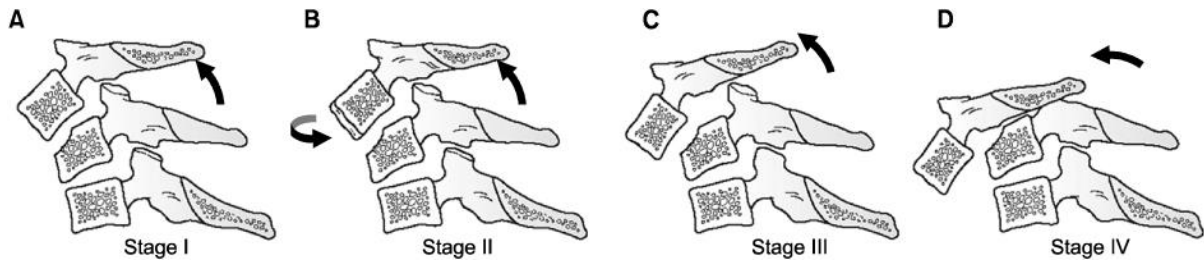


Figure 13: Mid-sagittal view of the four stages of distractive flexion described by (Allen, Ferguson, Lehmann, & O'Brien, 1982), figure from (Ahn, 2011, open access) including: a) stage 1 showing separation of posterior vertebrae and rounding of the vertebral margin, b) stage 2 showing one perched facet, c) stage 3 showing bilateral facet subluxation, and d) stage 4 showing bilateral facet dislocation.

2.2 Cervical Spinal Injuries Resulting from Head-first Impact

Automotive rollovers are chaotic collisions where the driver loses control of the vehicle, the vehicle inverts, causing occupants to hit their heads on the roof. Compression of the neck between the head and the thorax is thought to cause spinal cord injury (Viano, Parenteau, Gopal, & James, 2009). While comprising 2.6% of crashes, rollovers make up 29% of fatalities (National Highway Traffic Safety Administration (NHTSA), 2024), and cause the most AIS 2+ neck injuries compared to all other automotive collisions (Yoganandan, et al., 1989). AIS 2+ neck injuries are moderate to untreatable as defined by (Gennarelli & Woodzin, 2016). Most fracture-dislocations, which are the most common cause of spinal cord injury (Sekhon & Fehlings, 2001), are caused by rollovers (Parenteau & Viano, 2014; Quarrington, et al., 2018) compared to other automotive collisions. In rollovers with interactions with vehicle roofs, the lower cervical spine (C5-C6-C7) are injured most frequently, with the axis (C2) being injured fourth most frequently (Bambach, Grzebieta, McIntosh, & Mattos, 2013) (Fig. 14).

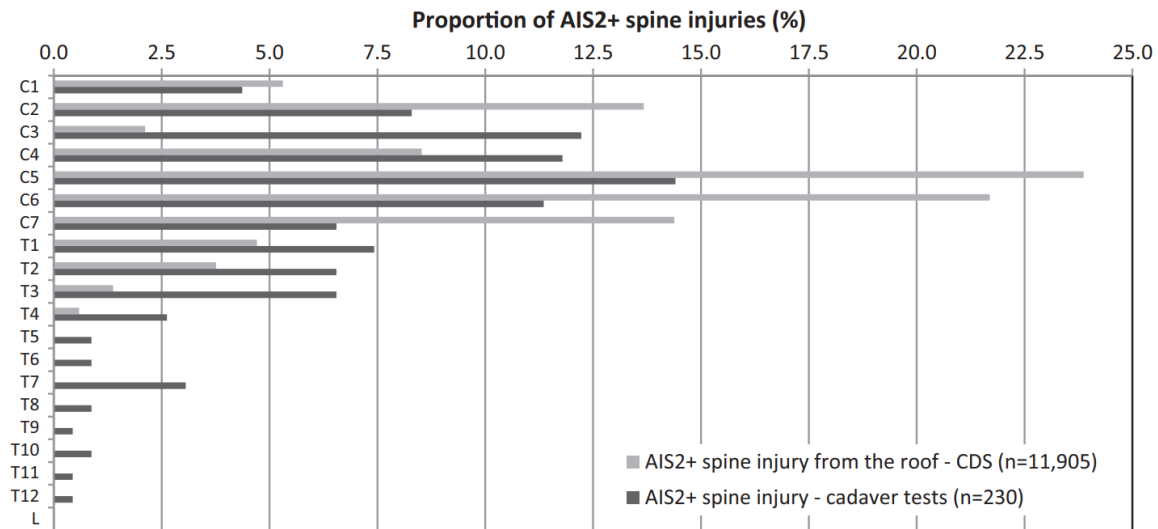


Figure 14: Cervical spine injury levels during rollovers with roof interaction versus cadaver tests, obtained from (Bambach, Grzebieta, McIntosh, & Mattos, 2013, used with permission from Elsevier). For Section 2.2, the relevant information is that the C5-C6-C7 is most commonly injured in rollovers.

Head-first impact is the suspected mechanism of neck injury in automotive rollovers. Peak neck loads in experimental rollovers with a crash test dummy occur when the head is in contact with the roof (Viano, Parenteau, Gopal, & James, 2009), and in an epidemiological study on rollovers, the roof assembly was identified as the source in over 70% of the neck injuries (Hu, Lee, Yang, & King, 2005). Therefore, studies attempting to understand cervical spine injuries from automotive rollovers have investigated HFI (Saari, et al., 2013; Darweesh M. , 2022; Nightingale, et al., 1997).

Head-first impact from diving into shallow water is another cause of cervical spine injury. Similar to automotive rollovers, diving cervical spine injuries are most commonly found in the lower cervical spine (McElhaney, Snyder, States, & Gabrielsen, 1979; Borijs, Gouader, Bousquet, Draper, & Roux, 2010; Yilmaz, et al., 2021), with the most common injuries to the spinal cord being: spinal cord compression, edema, and dislocation (Yilmaz, et al., 2021). In a biomechanical reconstruction of swimming pool neck injuries (McElhaney, Snyder, States, & Gabrielsen, 1979), it was also found that neck injuries occurred at impact velocities of 3.11 m/s or higher.

A study on cervical spine trauma in American football (Torg, Vegso, O'Neill, & Sennett, 1990) found that axial compression (associated with head-first tackling) was causing the highest percentage of fractures, dislocations, and subluxations in the sport. A similar study was conducted on ice hockey in Finland and Sweden (Mölsä, Tegner, Alaranta, Myllynen, & Kujala, 1999), finding that head-first impact (from being hit from behind into the boards) was the biggest cause of traumatic spinal cord injury, most commonly comprising compression fractures occurring in C5-C7.

2.3 Isolated Motion Segment Tests

Isolated motion segments have been used to understand the kinematic and kinetic limits of vertebrae. This includes dissecting the spine to isolate 2 or 3 vertebrae and their connective ligaments, and loading them in various directions.

Experimentalists (Nuckley, Nausdle, Eck, & Ching, 2007) loaded cadaveric specimens composed of 3 baboon vertebrae in compression until failure, determining that age correlated with compressive failure load, ranging from on average 0.55 kN (0 years old) to 4.68 kN (30 years old). To measure the effect of eccentric loading, experimentalists (Carter, Ku, Nuckley, & Ching, 2002) compressed cadaveric human specimens comprising 3 vertebrae (aged 30-94 years old), with both the superior and inferior vertebrae constrained to only move in the vertical direction to ensure compressive failure. Compressive failure loads were on average 0.77 kN for specimens initially rotated anteriorly (denoted as compression-flexion), 3.26 kN for specimens not initially rotated (denoted compression), and 3.47 kN for specimens initially rotated posteriorly (denoted compression-extension).

A variety of studies have loaded cervical segments in extension/flexion to a cutoff moment of between 1-10 Nm (Camacho, et al., 1997; Nightingale, et al., 2002; Wheeldon, Pintar, Knowles, & Yoganandan, 2006). These studies most often test within or just beyond a linear force-displacement range. Experimentalists (Nightingale, et al., 2007) rotated segments in pure flexion and extension until failure, reporting the flexion range of motion for failure for C4-C5 (13.1°) and C6-C7 (14.9°) and the failure bending moments for C4-C5 (19.2 Nm) and C6-C7 (21.7 Nm). Injuries from flexion loading all included complete disruption of the ligamentous structures between the vertebrae, and some specimens had associated fractures.

Ivancic et al. (Ivancic, et al., 2008) applied accelerations to C3-C4, C5-C6, and C7-T1 functional spinal units, which had a mass mounted to the superior vertebral potting, resulting in repeatable bilateral facet dislocations. During testing, forces and displacements were tracked, allowing for a hypothesis of a

possible injury mechanism for fracture-dislocation to be quantified. Generally, facets separated and rotated first, then had peak shears, and finally had negative peak vertical separations. The motion applied in this experiment resulted in the superior facet being locked in a position anterior to the inferior facet.

2.4 Experimental Head-First Impact Studies

2.4.1 TSM HFI and Compression Loading Experiments

To simplify the boundary conditions and reduce variability, most recent HFI and neck compression experiments have used a torso surrogate mass to replicate the constraining effect and loads produced by the torso (Nightingale, et al., 1997; Toomey, et al., 2009; Saari, Itshayek, & Cripton, 2011) (Fig. 15). The TSM boundary condition involves potting the 1-2 most superior thoracic vertebra(e) T1-T2, attaching a 16 kg mass to the potting, and putting the mass on rails to only translate in the axial direction of loading. A 16 kg mass was used (Nightingale, et al., 1997) to approximate the effective mass of the thorax that acted on the cervical spine during a head-first impact. The approximation was of the thorax of a 50th percentile male. In these experiments, (Nightingale, et al., 1997; Toomey, et al., 2009; Saari, Itshayek, & Cripton, 2011), the skin and muscles are removed, keeping the vertebrae and their connective tissues.

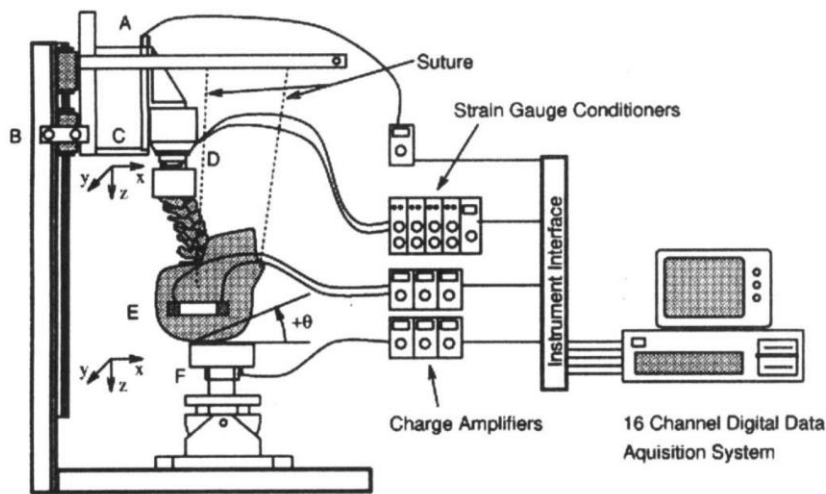


Figure 15: Cadaveric head and neck head-first impact test setup, including a Torso Surrogate Mass (Nightingale, et al., 1997, used with permission from SAE International).

In general, TSM HFI experimental studies reproduced clinically relevant injuries, but despite the uniform T1 boundary conditions, fractures were highly variable between specimens. The initial velocity (3.14 m/s) was based on the onset of neck injury in diving injury reconstructions (McElhaney, Snyder, States, & Gabrielsen, 1979).

Using a constrained torso boundary condition similar to a TSM, experimentalists (Myers, et al., 1991) quasi-statically loaded cadaveric osteo-ligamentous cervical spines axially, modifying the boundary condition of the superior end of the cervical spine (a stand-in for the head boundary condition) (Fig. 16). When there were no constraints on the head, except in the axial the direction of compression (the head was free to rotate and move laterally), there were no injuries in any of the lower cervical spines of the 6 specimens. When a rotational constraint was added to the head, but not a lateral constraint, there were consistently bilateral facet dislocations in the lower cervical spine. When the head was fully constrained, the forces and moments were highest, and there were compressive fractures in all specimens. It was concluded that the head boundary condition in this study changed the lower cervical spine injury outcome. Analysis of this study suggests that the head boundary condition acts as a stand-in for the impact surface, wherein a rigid slippery surface may allow the neck to avoid injury, but a compressing foam, which traps the head once compressed, may result in more neck injury. Similarly, in cadaveric HFI with a TSM boundary condition (Nightingale, et al., 1997) to a rigid surface and a padded surface, there was a higher frequency of injury in padded impacts compared to rigid impacts. A rigid impact surface allowed the head to escape injury, while the padded surface constrained the head, allowing for load to be concentrated between the head and torso.

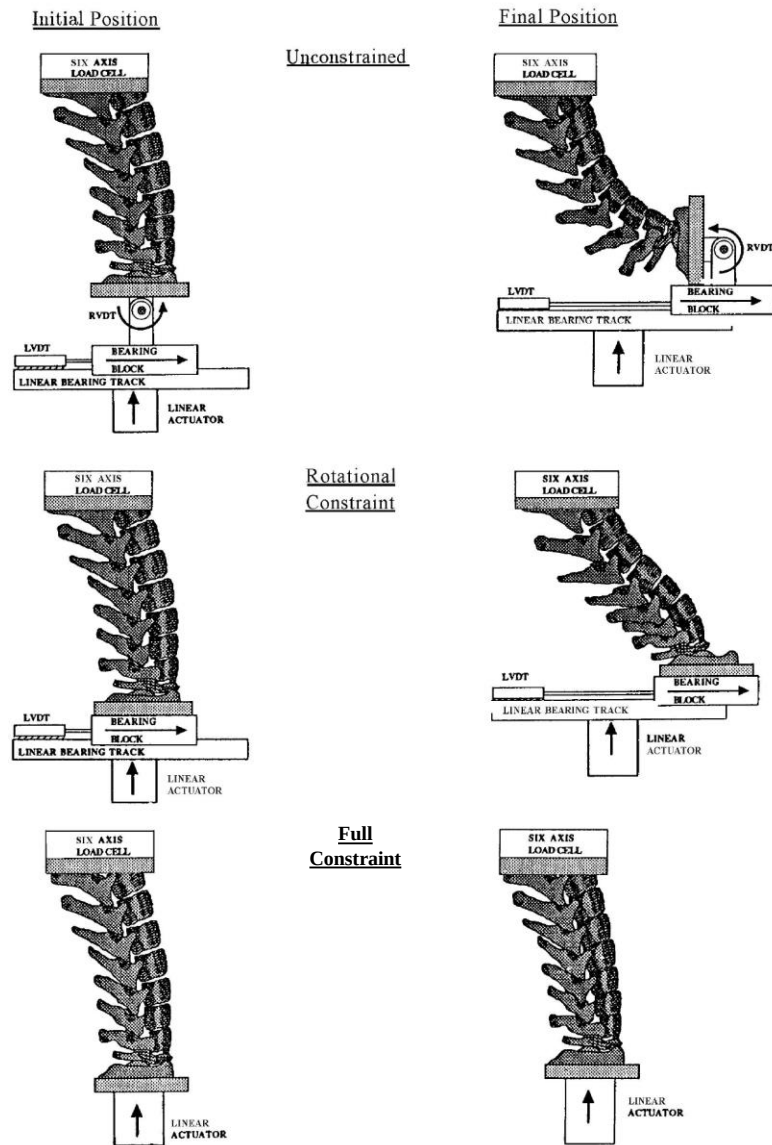


Figure 16: Quasi-static cadaveric neck compressive loading from (Myers, et al., 1991, used with permission from SAE International). In this experiment, the lower cervical spine was constrained in a similar method to the TSM, and the base of the skull had three different boundary conditions: unconstrained, rotationally constrained, and fully constrained.

Vertebrae each have three rotational degrees of freedom, therefore axially loading 6 cervical and 12 thoracic vertebrae can result in a variety of kinematic bending patterns. The angular deformation of the

cervical spine under compressive loading is referred to as buckling, and has been hypothesized as a factor in the injury outcome in compressive loading (Torg, Vegso, O'Neill, & Sennett, 1990). In experimental studies, buckling is often said to have occurred when there is a decrease in neck axial force without an injury occurring (Nightingale, et al., 1997; Myers, et al., 1991).

While the head boundary condition was concluded to affect the risk of injury (Myers, et al., 1991), the boundary condition also may have affected the buckling of the spine, which may be the reason for differing injuries. Specimens with unconstrained head boundary conditions showed first order buckling, wherein there was a rapid increase in lordotic curvature in the cervical spine. The unconstrained boundary condition had no injuries reported. Specimens with a rotational constraint all had bilateral facet dislocations in C5-C6, C6-C7, or C7-T1, wherein the eccentricity of T1 and the head constraint increased in the simulations. The increased eccentricity in the rotational constraint specimens resulted in upper cervical spine extension and lower cervical spine flexion, which can be classified as second order bending.

Buckling has been shown to occur dynamically by Nightingale et al. (Nightingale, et al., 1997). Unlike quasi-static compression, in dynamic loading, buckling was observed for all specimens regardless of head boundary condition (i.e. impact surface). After initial head impact and buckling (reduction in neck forces due to a “snap-through”), there was extension in the upper cervical spine and flexion in the lower cervical spine, which was denoted in this study as a first order mode of buckling. It was reported that specimens with first order buckling had flexion failures in the lower cervical spine due to the flexion in the lower cervical spine, and extension failures in the upper cervical spine due to extension in the upper cervical spine. Additionally, some specimens had transient higher-order bending, including three or more areas of alternating flexion and extension through the spine. The higher order buckling was unstable, and led to the spine moving away from the buckling. As a result, specimens either escaped injury or were injured later in the simulation. Therefore, the occurrence and type of buckling likely affects the severity and type of fractures in the cervical spine during HFI, and the mode of buckling is likely caused by factors such as posture and impact surface.

The direction of loading to the head has been shown to affect the injury severity of the cervical spine during HFI. Nightingale et al. (Nightingale, et al., 1997) rotated the impact plate in their TSM HFI simulations $\pm 30^\circ$, $\pm 15^\circ$, and 0° anterior-posteriorly. Impacts to posteriorly rotated plates were reported to be less severe than vertex and anterior impacts. The rotation of the impact plate caused the head to

move during the impact, but the head motions were reported to occur much later than injuries occurred. Similarly, injury patterns were not linked to the angle of the plate in this study.

One major difference between cadaveric HFI and in-vivo/real-life HFI is the lack of active musculature in cadaveric studies. During a head-first impact such as an automotive rollover or a diving accident, the neck muscles are typically co-contracted either as a reflex, or a slower reaction to the impending collision. Researchers (Al-Salehi, Cripton, & Siegmund, 2023) demonstrated that experiencing an inverted HFI leads to co-contraction in the neck, and subsequent postural changes. Similarly, Saari et al. (Saari, et al., 2013) performed cadaveric HFI experiments using a TSM boundary condition, and spring-loaded cables attached to the cervical spine to represent muscles. While they (Saari, et al., 2013) did not report significant changes in buckling, forces, or injuries when comparing specimens with and without spring-loaded cables as muscles, there was a decrease in time between head and neck loading when muscles were used. The decrease in time between head and neck loading was referred to as axial coupling, wherein specimens with spring-loaded muscles had more coupling between the torso and the head during HFI.

2.4.2 Full Body HFI Experiments

Full body post-mortem human subjects have undergone experimental HFI loading to study injury mechanisms (Culver, Bender, & Melvin, 1978; Nusholtz, Huelke, Lux, Alem, & Montalvo, 1983; Yoganandan, et al., 1986; Nusholtz, Melvin, Huelke, Alem, & Blank, 1981), finding evidence that the initial posture, impact location, and alignment of the skull effected the type and location of cervical spine fractures. Nusholtz et al. (Nusholtz, Melvin, Huelke, Alem, & Blank, 1981) impacted the crown of the skulls of cadavers using a pendulum impactor, with varying initial neck and torso angles relative to the impactor, and found different damage was produced by different neck/torso angle combinations, and that head motion did not predict the type of damage. When the torso angle was 0° and the neck angle was above 20° (the thoracic spine was parallel to the impact, the cervical spine was more than 20° rotated so that the head moved in the flexed direction), there was head flexion and C5-C7 extension/compression damage. The authors suggest extension/compression damage occurred in with the flexed head during the loading phase due to differential motion of the vertebrae. When the torso angle was 0° and the neck angle below 20°, there was a “serpentine” head trajectory, meaning the head moved in the direction of the impactor during contact, and then after the impact travelled in an anterior and then posterior direction, creating an “S.” During the serpentine head motion, the damage was also identified to be compression-extension, occurring in C2-T1. In tests with a torso angle above 15° and

a neck angle below 20° , there was also a “serpentine” head trajectory, and there was flexion/compression damage in the upper thoracic spine (T1-T4), and extension/compression damage in the upper cervical spine. Finally, tests with a torso angle above 15° and a neck angle above 20° had flexion/compression damage in the upper thoracic spine, and one case had extension/compression in C5 and C7. Later, Nusholtz et al. (Nusholtz, Huelke, Lux, Alem, & Montalvo, 1983) dropped cadavers on their heads (rather than pendulum impacts), verifying their findings from (Nusholtz, Melvin, Huelke, Alem, & Blank, 1981), and added that flexion damage could be produced if an initial posture that allowed for non-sagittal plane motion was used (Fig. 17). Additionally, they reported that the thoracic spinal motion was important in the kinematics and damage during impacts.

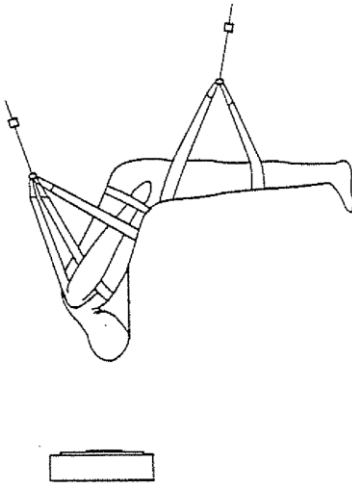


Figure 17: Setup of full body head-first impact from (Nusholtz, Huelke, Lux, Alem, & Montalvo, 1983, used with permission from SAE International).

In a more recent short communication, researchers (Kerrigan, et al., 2014) performed full body HFI experiments on four specimens. Compared to TSM experiments with similar loading conditions from (Nightingale, et al., 1997), observed injuries in TSM experiments were more severe than when using a FB. The full body HFI only observed injury during one of the four HFI experiments, whereas HFI with similar impact velocity (Nightingale, et al., 1997) had neck injuries in 18 of 22 specimens.

2.5 Numerical Head-First Impact Studies

2.5.1 TSM HFI and Compression Loading Simulations

FE models allow for quantifying the effect of changing parameters in a computational environment on the mechanical response. In particular, FE HBMs have been used to understand the mechanism of injury under HFI. Numerical HFI studies have also applied the TSM boundary condition, allowing for validation against TSM cadaveric HFI studies. A variety of neck models have been developed and validated in TSM HFI (Yang, Zhu, Luan, Zhao, & Begeman, 1998; Hu, Yang, Chou, & King, 2008; Nightingale, Sganga, Cutcliffe, & Bass, 2016; Nasim, Cernicchi, & Galvanetto, 2021; Darweesh M. , 2022), in general demonstrating comparable forces and kinematics with the experiments (Fig. 18a) but none have compared the TSM boundary condition to an FB boundary condition in HFI.

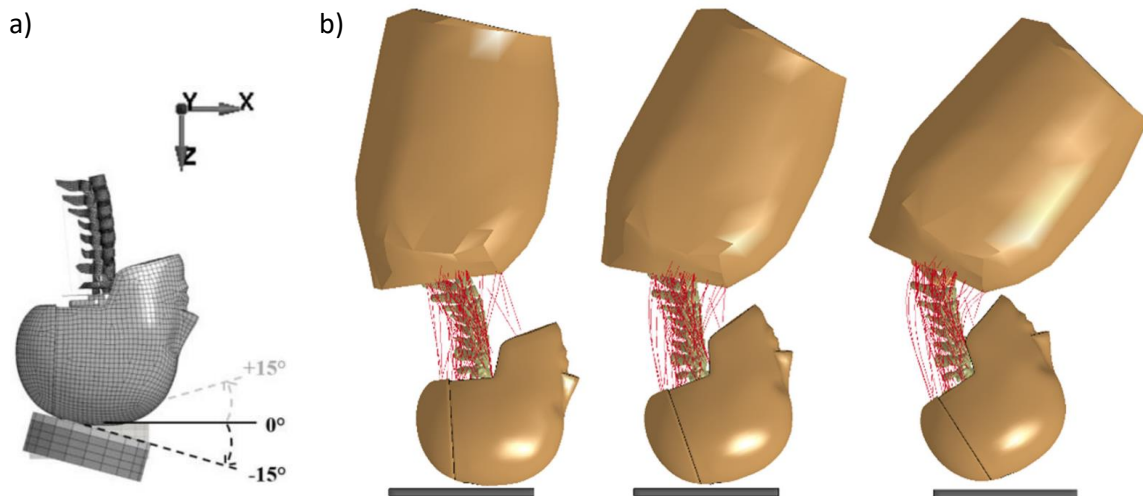


Figure 18: a) Computational head-first impact of a computational head and neck with a Torso Surrogate Mass boundary condition from (Nasim, Cernicchi, & Galvanetto, 2021, used with permission from Sage Publications), b) setup of head-first impact to a head, neck, and rigid torso in three different postures: neutral (0° angle between the head and torso), an intermediate flexed position (32° between the head and torso), and a flexed position (64° between the head and torso) from (Nightingale, Sganga, Cutcliffe, & Bass, 2016, used with permission from Elsevier).

Researchers (Nightingale, Sganga, Cutcliffe, & Bass, 2016) performed HFI to a computational human body model including a head and neck attached to a rigid thorax (Fig. 18b). In this study the initial

posture was varied from neutral to increasingly flexed. It was found that increasing the flexion up to about 40° of flexion straightened the spine. The straightening of the spine increased the compressive force in the spine during impacts. The higher compressive force was hypothesized to be caused by changing in the buckling behavior, where impacts with increasingly straight spines had higher order buckling.

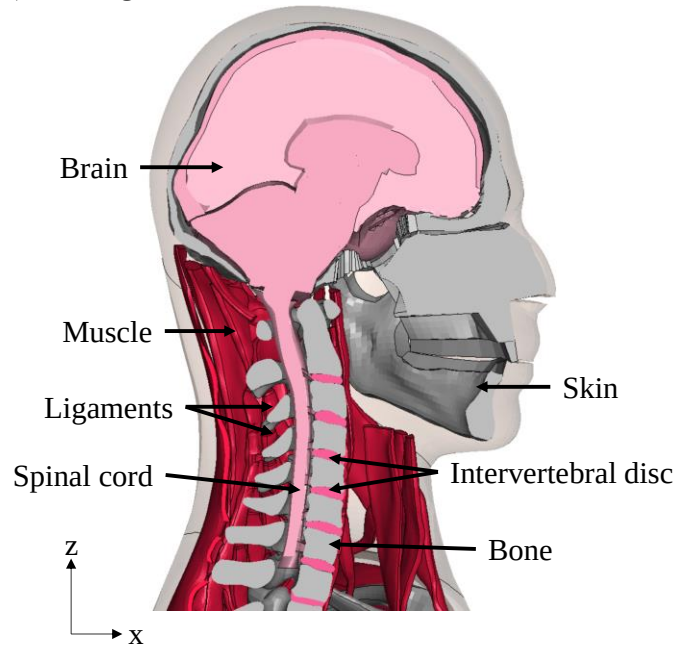
While muscle co-contraction is expected to change the initial posture in HFI (Al-Salehi, Crompton, & Siegmund, Pre-Impact Cervical Spine Posture of Human Subjects in Headfirst Free Falls, 2023) there are additionally thought to be non-postural effects of muscular co-contraction. Researchers (Nightingale, Sganga, Cutcliffe, & Bass, 2016) additionally modelled active muscles as beam elements, and found an increase in compressive and shear forces; larger than was expected from muscular force alone. Similarly, Hu et al. (Hu, Yang, Chou, & King, 2008) used a computational HBM to simulate impact to the head with no muscles, with passive muscles, and with active muscles. Here, 23 pairs of cervical spine muscles were modelled with hill-type beam elements, with both a passive and active force component. They reported that injury risk was significantly higher in simulations with active musculature compared to passive and no muscles.

The Global Human Body Models Consortium 50th percentile male detailed human body model version 6.0 (Fig. 19) was developed using MRI and CT scans from a 26-year-old adult with mid-size male anthropometrics (Gayzik, et al., 2011). The neck region of the M50 model includes detailed representations of hard and soft tissue in the human body, using material properties obtained experimentally from: bones, ligaments, and passive muscles, which have been validated in a hierarchical manner against cadaveric and volunteer data (Barker & Cronin, 2021). The neck of GHBMCM50-O, Version 5.0 has been assessed against experimental head-first impact loading (Darweesh, Corrales, Crompton, & Cronin, 2020), demonstrating comparable axial forces and fracture pattern and onset to experimental results (Nightingale, et al., 1997).

a) GHBMCM V6 50th Percentile Male

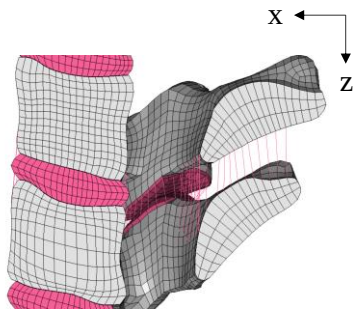


b) Mid-sagittal section of head and neck labelled



c) Mid-sagittal section of C4-C5 showing intervertebral ligaments and mesh.

i) GHBMCM M50



ii) Human body depiction (Gray, 1918)

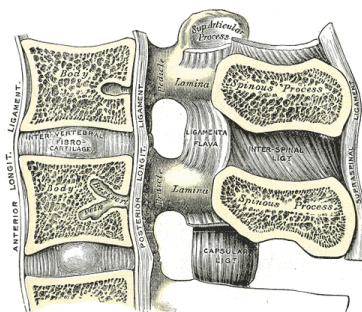


Figure 19: GHBMCM V6 50th percentile male human body model a) full body depiction in a neutral seated posture, b) a mid-sagittal section of the head and neck with parts labelled, and c) a mid-section of C4-C5 demonstrating intervertebral ligaments and mesh (i) contrasted with a human body depiction (Gray, 1918, public domain) showing intervertebral ligaments and bone porosity (ii).

2.5.2 Full Body HFI Simulations

A FB multibody model composed of rigid bones and skeletal neck muscles was simulated in head-first impacts, with postures similar to those used in injurious rugby tackling. Forces, moments, and rotations were compared for the different postures, and the extended posture was found to have injurious forces and moments (Silvestros, et al., 2024) (Fig. 20). The multibody model did not consider vertebral deformation, which may have affected forces generated in the simulation, but this study supports buckling as a cause of lower cervical spine fracture-dislocations, rather than hyperflexion.

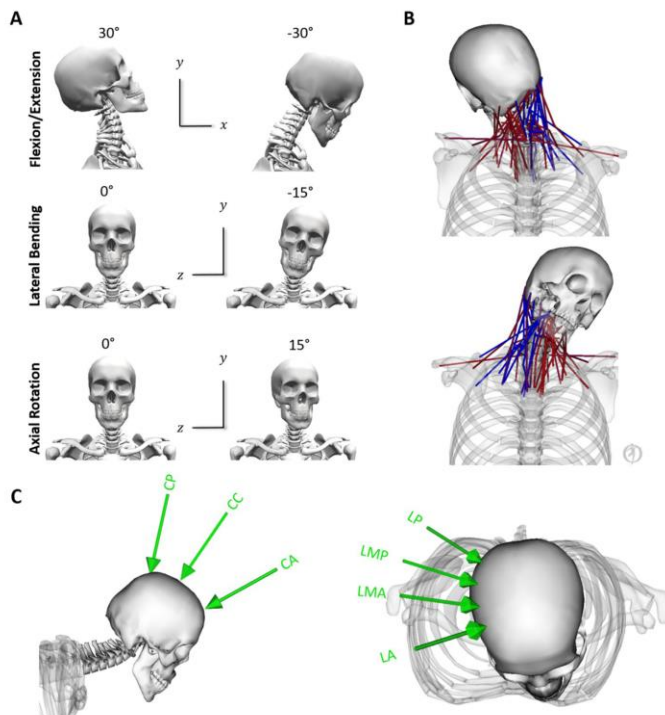


Figure 20: Setup of simulated head-first impact to a full body multibody human body model repositioned and pre-tensioned to match initial conditions of injurious rugby tackles, from (Silvestros, et al., 2024, open source)

A full body HBM (THUMS v4.01 occupant seated model) was used to model the effect of: impactor velocity, displacement, and direction in HFI (Mattos, McIntosh, Grzebieta, Yoganandan, & Pintar, 2015), reporting that serious brain injury from HFI was more likely at impacts angled further away from vertical, whereas injury to the cervical spine was more likely in impacts closer to vertical (Fig.

21). The simulated full body impacts (Mattos, McIntosh, Grzebieta, Yoganandan, & Pintar, 2015) only evaluated a nominal seated position, while posture during HFI has been found to vary (Newell, Blouin, Street, Crompton, & Siegmund, 2013).

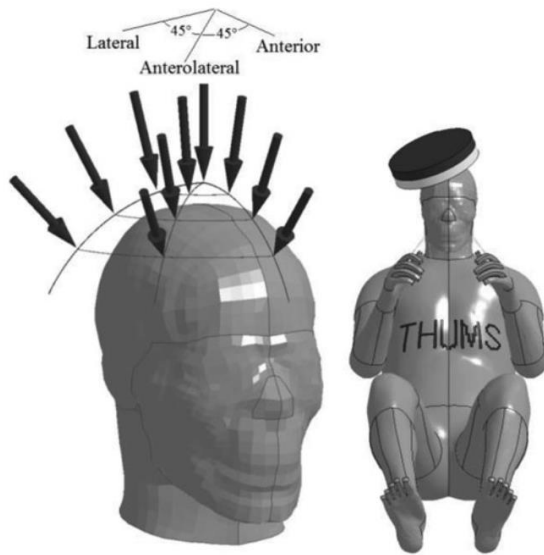


Figure 21: Setup of head-first impact simulated to a seated full body model (THUMS) with impacts from varying directions, obtained from (Mattos, McIntosh, Grzebieta, Yoganandan, & Pintar, 2015, open source).

2.6 Summary

To summarize the background, in head-first impact, injuries typically occur in the lower cervical spine, fracture-dislocation being common. To prevent injuries from head-first impact, they must first be replicated in a lab. Biomechanical studies have attempted to replicate lower cervical spine fracture-dislocations by experimentally and computationally modelling HFI loading at the segment level, using a TSM, and using a FB (Fig 22). While providing insight into HFI, none of these studies have reliably produced lower cervical spine fracture-dislocation in dynamic loading. The three objectives of this thesis are linked to the background, with Study 1 attempting to obtain a lower cervical spine fracture-dislocation using a motion segment model, Study 2 comparing the TSM and FB, and Study 3 comparing experimental and numerical FB models.

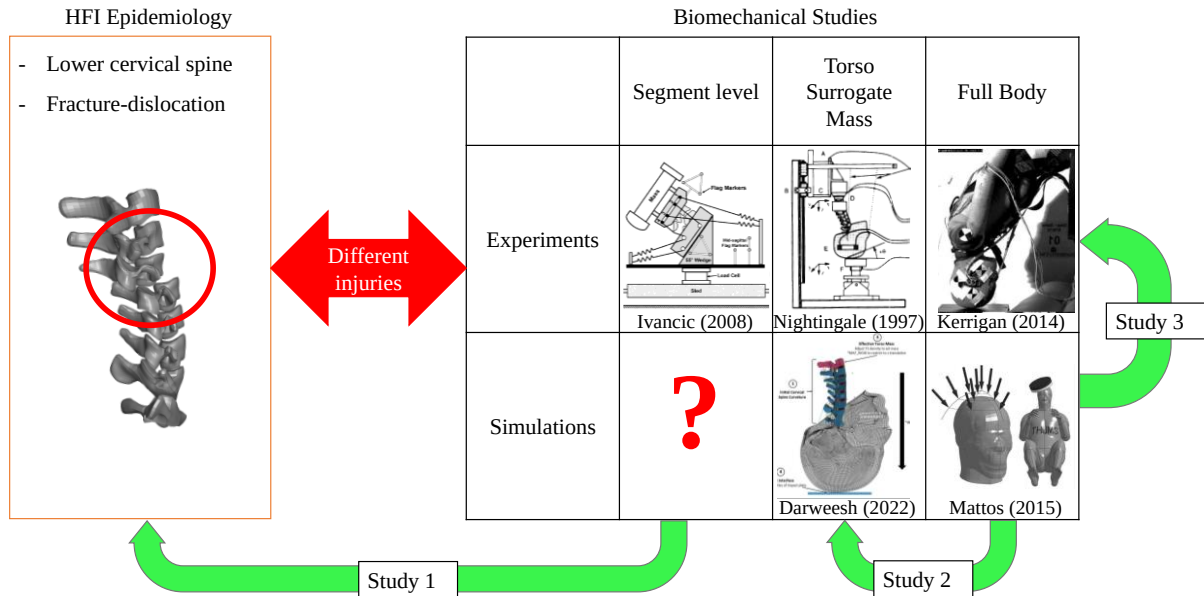


Figure 22: Summary of the background showing that the injuries from head-first impact, which are lower cervical spine fracture-dislocations, have not been produced by biomechanical studies. The biomechanical studies shown include experimental (Ivancic, et al., 2008; Nightingale, et al., 1997; Kerrigan, et al., 2014) and simulated (Darweesh M. , 2022; Mattos, McIntosh, Grzebieta, Yoganandan, & Pintar, 2015) HFI studies. Additionally, green arrows connect the objectives of this thesis to the background, with study 1 attempting to obtain a lower cervical spine fracture-dislocation using a motion segment model, Study 2 comparing the TSM and FB, and Study 3 comparing experimental and numerical FB models in HFI.

Chapter 3

Methods

Methods in this document have been divided hierarchically into three studies, with Section 3.1 corresponding to Study 1, Section 3.2 corresponding to Study 2, and 3.3 corresponding to the Study 3. Study 1 uses isolated vertebral segment models to understand forces associated with fracture-dislocation. Study 2 aims to understand the effect of a TSM boundary condition on HFI by comparing it with a FB model in multiple postures. Study 3 compares the FB M50 response to a cadaveric HFI experiment.

3.1 Study 1: Isolated Motion Segment Model to Investigate Response Under Combined Compression and Anterior-Posterior Bending and Relationship to Fracture-Dislocation

Given the prevalence and risk of fracture-dislocations in head-first impacts and the lack of reliability in reproducing them experimentally or computationally, the aim of Study 1 was to determine kinetics and kinematics associated with fracture-dislocations, which involved two sub-studies: obtaining the fracture of interest, and subsequently obtaining the dislocation of interest.

3.1.1 Obtaining the Fractures Associated with Distractive Flexion Fracture-Dislocations Using Different Ratios of Linearly Increasing Compressive Force and Bending Moment

C2-C7 motion segments composed of 2 vertebrae were loaded in combined flexion bending and compressive loading to determine forces associated with failure in the numerical model. Different combinations of compression and flexion are hypothesized to result in the three most common patterns of fracture and dislocation in the cervical spine (Allen, Ferguson, Lehmann, & O'Brien, 1982): distractive flexion, compressive extension, and compressive flexion. However the true combinations of compression and flexion that cause these patterns of fracture and dislocation is unknown. The segment models for C2-C3, C3-C4, C4-C5, C5-C6, and C6-C7 were isolated and set up using a previously published methodology (Barker & Cronin, 2021) (Fig. 23a). Isolating the motion segment models involved removing all parts from the GHBMCM50 except the two vertebrae (cortical and trabecular bones) of interest and their connective tissues. The included connective tissues were the intervertebral

discs, anterior and posterior longitudinal ligaments, ligamentus flavum, and interspinous ligaments. The models included a rigid top endplate of the superior vertebra and bottom endplate of the inferior vertebra. Loads were applied to the superior endplate, and the inferior endplate was fixed in place.

Separate simulations with pure compressive loading and pure bending moments were first analyzed. Subsequently, combined loading was simulated, where both compressive load and bending moment were ramped up simultaneously (Fig. 23b) until failure occurred in the bone elements. Bone failure was modelled as element erosion, which used plastic strain thresholds for both cortical (2D quadrilateral elements) and trabecular (3D hexahedral elements) bone representation, which were 1.78% and 9.5%, respectively. The ratio of compression to bending moment in the combined loading simulations were modified so that each failure mode had at least two different combinations of compression and flexion.

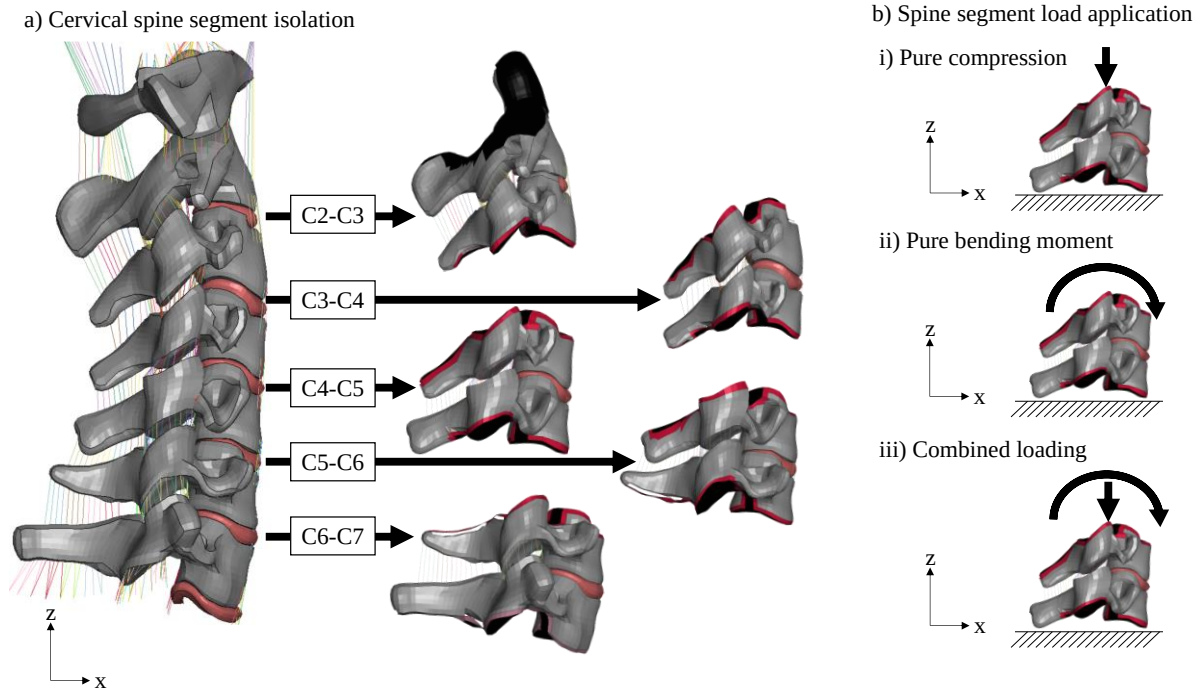


Figure 23: a) Isolation of C2-C7 motion segments composed of 2 vertebrae using methods from (Barker & Cronin, 2021), and b) the load applied to each segment, including: i) pure compression to the superior vertebra, ii) pure bending moment to the superior vertebra, and iii) combined compression and bending moment to the superior vertebra. For all cases, the inferior vertebra was held rigidly.

In the segment level compression and bending moment simulations, the direction of rotation of the superior vertebra was measured against the fixed inferior endplate. As well, the applied compressive force and bending moment were tracked throughout the simulation, in addition to the location and time of initial element erosion.

3.1.2 Dislocating a Segment Model of 2 Vertebrae using Pre-Defined Motion, Based on Experiments from (Ivancic, et al., 2008)

To investigate lower cervical spine fracture-dislocation at the segment level, C4-C5, C5-C6, C6-C7 segments were modeled. Only the lower cervical spine was assessed for fracture-dislocation because in HFI, the lower cervical spine is site of fracture-dislocations. As in Section 3.1.1, segments were isolated using the methodology used in (Barker & Cronin, 2021). To obtain the desired dislocation, displacement-based control was applied to the rigid superior endplates of the superior vertebrae. The rigid inferior endplates of the inferior vertebrae were held in place with a rigid 6-degree-of-freedom constraint. Displacement-based control was used rather than force-based to improve simulation stability, especially after element failure, and additionally to more precisely control the motion of the vertebrae.

In these simulations, an anterior sagittal rotation (flexion), followed by an anterior shear, followed by a posterior sagittal rotation were applied. Rotations and translations were applied relative to the rigid inferior boundary condition. The order of displacements was followed to match the order of the peak rotations and translations in the experiments of (Ivancic, et al., 2008), in which bilateral facet dislocations were reliably produced, which were measured relative to markers on the inferior facet.

While in the experiments of Ivancic et al. (Ivancic, et al., 2008) peak rotations were on average 63°, those experiments produced facet dislocations without fracture, while the intent of the present study was to obtain fracture-dislocations. Moreover, the “representative example” provided in the paper had a peak rotation of 37°. Given that distractive flexion failures representative of those observed by Allen et al. (Allen, Ferguson, Lehmann, & O'Brien, 1982) by applying moment and compression (no shear force) were obtained in 3.1.1, a simulation was run with a rotation of 10°, 20° and 30° in the global x-z plane. The amount of shear displacement (in the global x-direction) applied to the endplate was scaled so that the displacement of the facet matched the experimental displacement. A diagram of the displacement control methodology, including an anterior sagittal rotation, followed by an anterior translation, and a sagittal posterior rotation, is shown (Fig. 24).

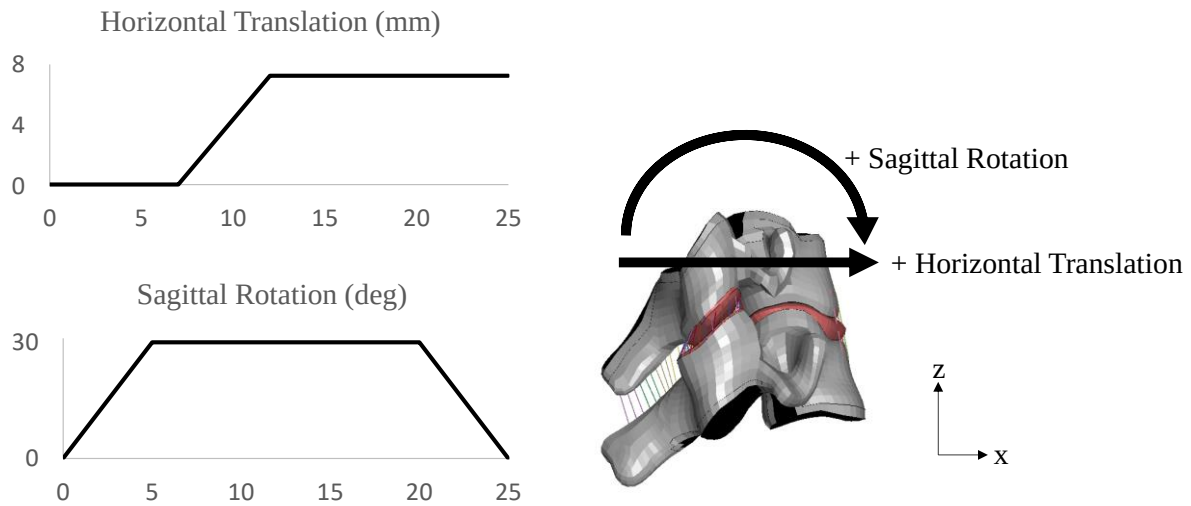


Figure 24: Applied translational constraints in the modelling of lower cervical spine (C4-C7) motion segment of 2 vertebrae fracture-dislocation including the horizontal translation and sagittal rotation input curves for a 30° simulation (left) and a diagram of the direction of loading (right). The applied loading includes: positive sagittal rotation (either 10°, 20°, or 30°) occurring in 5 ms, a horizontal translation of 7.15 mm occurring in 5 ms, an 8ms pause to allow the model to stabilize, and a negative sagittal rotation in 5 ms, returning the superior vertebra to 0°.

In the modelling of displacement-controlled fracture-dislocations, the forces (normal and transverse) as well as the bending moment applied to the rigid superior edge of the vertebrae were assessed with the model. Moreover, the location and time of element erosion in the hard tissues were monitored to explain changes in force and moment.

3.2 Study 2: Importance of Neck Boundary Condition and Posture on the Cervical Spine Response During a Head-First Impact

Study 2 compared the response of TSM and FB in three different postures and three different initial velocities to head-first impact.

3.2.1 Reposturing the Head and Neck of the GHBMC Full Body Model from a Neutral Posture to Flexed and Extended Postures

The head and neck of GHBMC M50 full body model was in a neutral posture and seated position (FB_{NEUT}), -5° head angle (Fig. 25b), 1846 mm standing stature, 26-year-old and 0.53 standing/seated stature ratio). The full body model was repostured to flexed (FB_{FLX}) and extended (FB_{EXT}) (Fig. 25 b) postures, using targets from the cervical spine posture predictor (Reed & Jones, 2017), and following a previously developed methodology (Corrales & Cronin, 2021). The Cervical Spine Posture Predictor head angle parameter was increased and decreased by 32° to develop the target postures for the FLX and EXT models respectively. A value of 32° was applied to have substantially different postures while being within the range of motion for flexion (Ferrario, Sforza, Serrao, Grassi, & Mossi, 2002), and extension for an upside-down relaxed subject (Newell, Blouin, Street, Crompton, & Siegmund, 2013). The head angle in the neutral posture was representative of an upside-down *in vivo* human subject looking forwards (Newell, Blouin, Street, Crompton, & Siegmund, 2013). The skull and vertebral positions were extracted from the Cervical Spine Posture Predictor for FLX and EXT postures and used as input for the reposturing process using PIPER (Corrales & Cronin, 2021). Following the repositioning, the soft tissue meshes were smoothed using the transformation smoothing option in PIPER (Janak, Lafon, Petit, & Beillas, 2018). The mesh quality of the repostured models was evaluated using previously reported metrics for human body models (warpage $<50^\circ$, aspect ratio <8 , skew $<70^\circ$, and Jacobian >0.4 (Corrales, Bolte, Malcolm, Pipkorn, & Cronin, 2022)).

3.2.2 Isolated Head and Neck Torso Surrogate Mass Model Extraction, Setup, and Boundary Conditions Used for Head-first Impact (HFI) in Three Postures

The neutral, extended, and flexed isolated head and neck models with a TSM (TSM_{NEUT} , TSM_{EXT} , and TSM_{FLX} , respectively) were extracted from the FB_{NEUT} , FB_{EXT} , and FB_{FLX} models following the methodology presented by (Barker & Cronin, 2021) (Fig. 2b). The effective mass of the torso was approximated by a concentrated load of 16 kg of a cylindrical steel mass constrained to the first thoracic vertebra (T1). The 16 kg mass was recommended to approximate the effective mass of a 50th percentile male upper torso from literature (Nightingale, et al., 1997). The mass was allowed to translate along the 'Z' axis (Fig. 25a) and constrained in the other degrees of freedom (constrained in 'X' and 'Y' translational, and all rotational degrees of freedom). The T1 constraint and mass represent the potting and constraining rails used in HFI TSM experiments (Nightingale, et al., 1997). Then, the TSM_{NEUT} ,

TSM_{EXT} and TSM_{FLX} models were simulated in head-first impact conditions by applying an initial velocity of 2.00 m/s, 3.14 m/s and 3.60 m/s towards the rigid plate (Table 1). The models were solved using a commercial explicit FE solver (LS-DYNA, R9.2, Ansys, double precision, Massive Parallel Processing). Version R9.2 was used for consistency with the GHBMCM validation (Barker & Cronin, 2021). Neck muscles were relaxed and exhibited passive mechanical responses for both FB models and TSM models. In TSM, the ends of the muscles were constrained to move with T1.

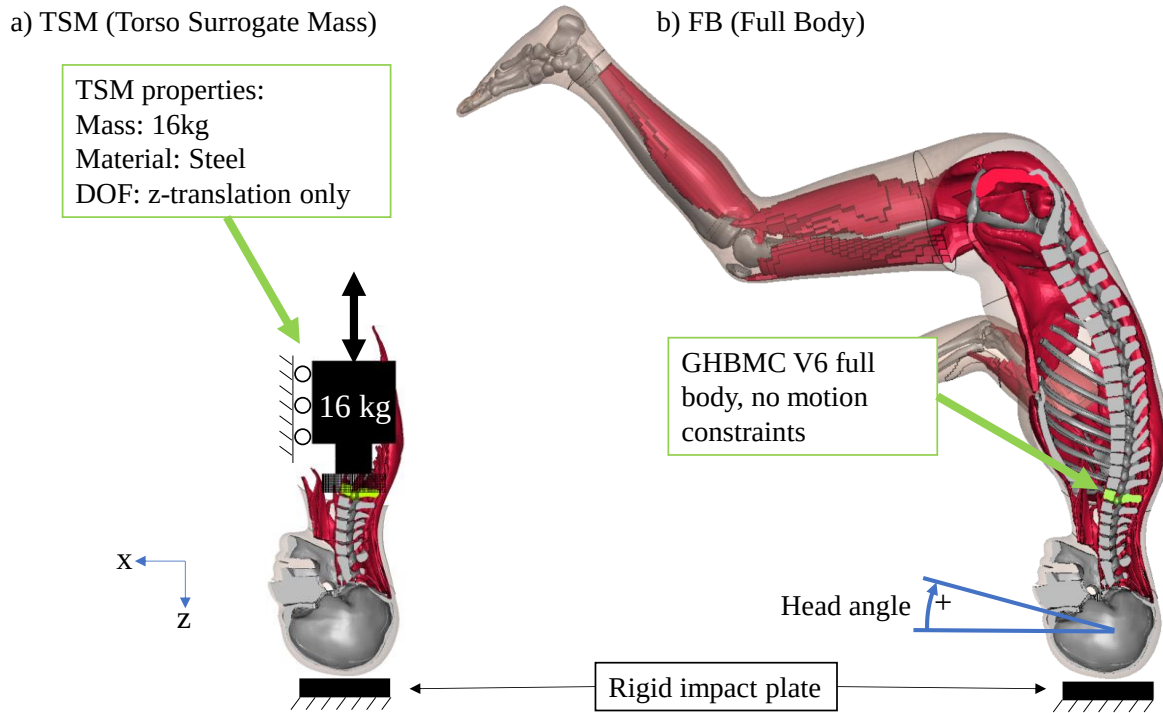
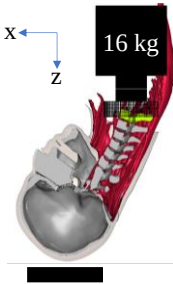
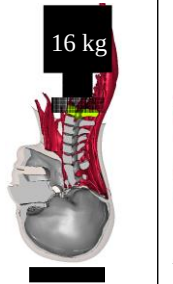
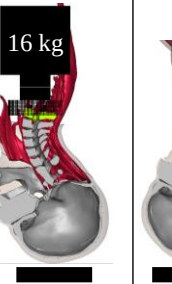
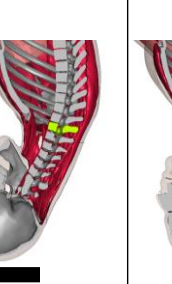
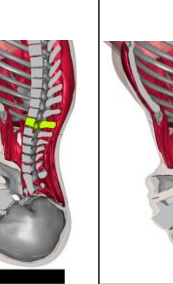
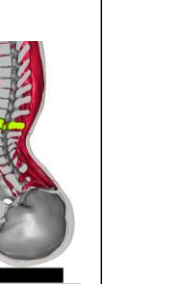


Figure 25: Boundary conditions: a) Torso Surrogate Mass including an isolated head and neck, with a 16 kg mass applied to the T1 and constrained to move only in the z direction, and b) Full Body with no external constraints and indicating the head angle with the positive direction shown, measured between the infraorbital and the tragon of the head. T1 vertebra shown in green.

Table 1: Torso Surrogate Mass and Full Body models with three pre-impact postures: FLX (flexed neck), NEUT (neutral neck), and EXT (extended neck) in three impact velocities. In TSM, the T1 (green) was constrained to only translate in the z direction (no rotations).

Torso model	Torso Surrogate Mass			Full Body		
Posture	Flexed	Neutral	Extended	Flexed	Neutral	Extended
Label	TSM _{FLX}	TSM _{NEUT}	TSM _{EXT}	FB _{FLX}	FB _{NEUT}	FB _{EXT}
Image						
Initial velocity	-2 m/s (z axis)					
	-3.1 m/s (z axis)					
	-3.6 m/s (z axis)					

3.2.3 Full Body Model Setup and Boundary Conditions Used for Head-first Impact (HFI) in Three Postures

After repositioning, the FB models (FB_{NEUT}, FB_{EXT} and FB_{FLX}) were rotated 180° into an upside-down position relative to the applied gravity vector (Fig. 25b). No constraints were placed on the full body models, and an initial velocity was applied to the whole body. A constrained rigid plate was located below the head and interacted with a frictionless contact defined between the head and the plate. To simulate the effect of being dropped, the FB_{NEUT}, FB_{EXT} and FB_{FLX} models were prescribed initial downward velocities of 2.00 m/s, 3.14 m/s and 3.60 m/s. The 3.14 m/s was the nominal velocity reported in (Nightingale, et al., 1997) as the threshold for injurious impact in a diving injury study (McElhaney, Snyder, States, & Gabrielsen, 1979). It was found that 3.60 m/s was the highest velocity that allowed for the simulations to run to completion. Finally, 2.00 m/s was used as a lower value, corresponding to head-first impacts that resulted in no injury (Nusholtz, Huelke, Lux, Alem, & Montalvo, 1983; Nightingale, et al., 1997).

3.2.4 Assessment of Model Response for HFI

The contact force between the head and the impact plate were extracted from the simulations and compared between FB and TSM boundary conditions.

The intervertebral angles were measured in the cervical spine for each pair of vertebrae from C2-C3 to C6-C7. The kinematics of the most anterior and posterior corners of the inferior vertebral endplate for each vertebra were tracked and the relative angles were calculated (Fig. 26a).

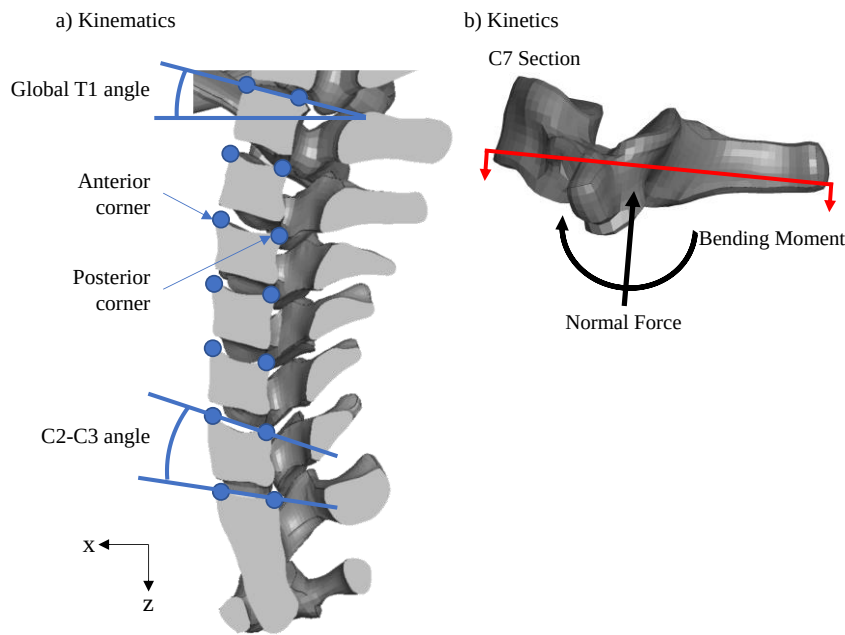


Figure 26: a) Intervertebral angle measurement location. b) Kinetics measurements cross-section locations (left), and the measured forces (right).

Global T1 kinematics were tracked to measure how the upper thoracic spine influenced the cervical spine during impacts. Rigid body translation (motion of the geometric center of T1) was used to track the motion of T1 relative to the global coordinate system. T1 rotation was measured against the global reference frame using the most anterior and most posterior corners of the superior endplate and the global 'X' axis.

A cross-section was analyzed for predicted force and moment at the C7 level, representing the forces in the lower cervical spine. The cross-section passed through the vertebral body and spinous process of

C7, and the normal force (perpendicular to the plane) and sagittal bending moment (Fig. 26b) were reported.

An additional investigation was undertaken to estimate hard tissue fracture from HFI. A critical-value plastic strain-based element erosion criterion from (DeWit & Cronin, 2012) was implemented in the FB and TSM models, and the models were re-run with element erosion included. The erosion model used plastic strain thresholds for both cortical (2D quadrilateral elements) and trabecular (3D hexahedral elements) bone representation, which were 1.78% and 9.5%, respectively. To quantitatively compare tissue failure between the models, the volumes of eroded bone elements were tracked, which included the volume of trabecular and cortical elements.

3.3 Study 3: Assessing the Neck Response of the Full Human Body Model with Whole PMHS Experimental Data

In Study 3, the response of the FB model to HFI was investigated and compared with full body PMHS experiments. Version 5 of the 50th percentile male GHBMC neck model (Darweesh M. , 2022) has been validated against experimental HFI studies (Nightingale, et al., 1997; Saari, Itshayek, & Cripton, 2011). GHBMC Version 6 uses the Version 5 neck model and includes enhancements to its biofidelity. Similarly, the full body has been validated against frequently occurring automotive collisions such as frontal impact (Gayzik, et al., 2011), but not head-first impact. Therefore, this study had two aims: 1) to determine if a computational model and an experimental full body PMHS test (Kerrigan, et al., 2014) produced similar plate reaction forces under similar velocity impacts, and 2) to compare thoracic kinematics (measured as T1 rotation) between a computational model and an experimental full body cadaveric test (Kerrigan, et al., 2014).

The secondary data use in this study were approved by the Office of Research Ethics at the University of Waterloo (file # 46987), wherein unpublished data from (Kerrigan, et al., 2014) was compared with the GHBMC results. The experimental full body cadaveric test from (Kerrigan, et al., 2014), involved dropping four inverted cadavers, with the cadaver heads hitting a plate covered in a closed-cell Polyvinyl chloride foam.

3.3.1 Numerically Modelling the Experimental Impact Surface

The foam used in the experimental tests (Kerrigan, et al., 2014) was characterized by dropping a 1.895 kg aluminum semi-spherical impactor with a diameter 145.8 mm onto a piece of 1” thick closed-cell

Polyvinyl Chloride (PVC) foam. The kinematics of the cylinder and the reaction force of a load cell beneath the foam pads were plotted. The aluminum semi-spherical impactor drops were at 4 different initial velocities (2.0, 3.2, 4.4, and 5.3 m/s) in order to capture rate effects of the foam relevant to head-first impact. Sample data at 3.2 m/s from these experiments are shown (Fig. 27).

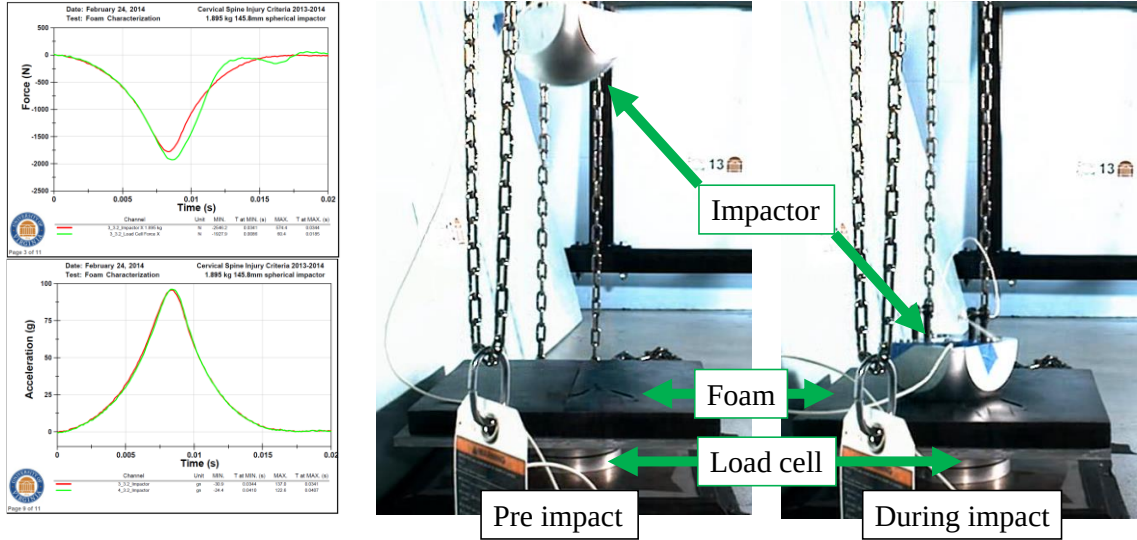


Figure 27: Force-time and acceleration time plots (left), and two images showing the semi-spherical aluminum impactor hitting the foam during characterization experiments (right), from the experiments of (Kerrigan, et al., 2014).

To model the closed cell PVC foam in LS-DYNA, the Material Type 57 (*MAT_LOW_DENSITY_FOAM) material was used, including parameters for a low density closed-cell Polyvinyl Chloride (Products, n.d.). Additionally, a stress-strain curve was extracted from the force-time and acceleration-time plots of the 3.2 m/s trial. The 3.2 m/s trial was selected because this loading rate most closely matched the cadaveric HFI initial velocities from (Kerrigan, et al., 2014). To extract stress from the force-time plots, the impactor contact area was approximated visually from videos of the experiments, and divided by the load cell force to approximate average stress. To extract strain (ϵ) from the acceleration-time plots, the acceleration was integrated twice to determine displacement (s), and the known initial thickness of the foam was (l_0). Equations 1-3 show how stress and strain were obtained from force (F), approximate area (A_{approx}), acceleration (a).

Equation 1

$$\sigma = \frac{F}{A_{approx}}$$

Equation 2

$$s = \int \int a \, dt \, dt$$

Equation 3

$$\epsilon = \ln \left(\frac{l_0 - s}{l_0} \right)$$

To test the foam model, the aluminum semi-sphere impactor experiment was replicated in LS-DYNA, which involved modelling: the aluminum semi-sphere impactor, the foam, a metal plate below it, and contacts between adjacent parts (Fig. 28).

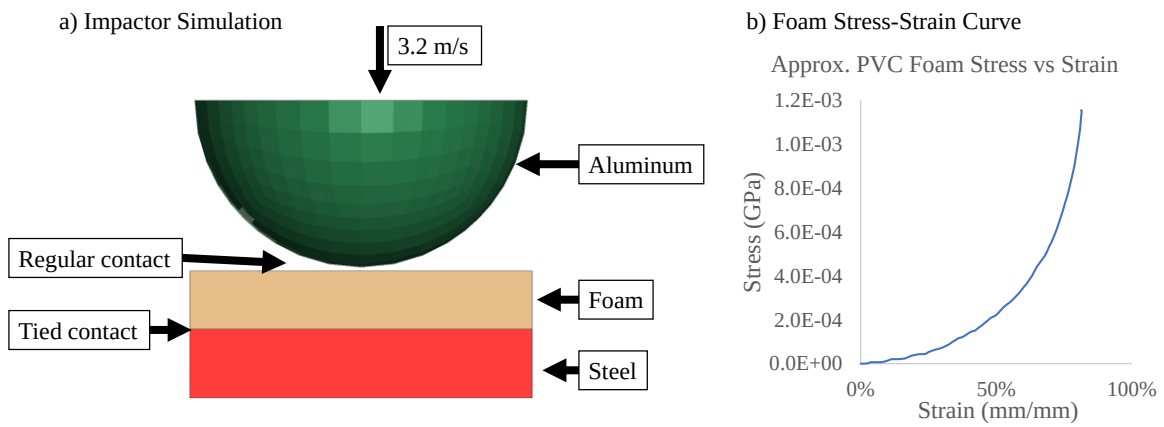


Figure 28: a) Setup of the impactor simulation including an aluminum semi-spherical indenter with an initial velocity of 3.2 m/s, a 1” thick foam pad with material properties determined and verified, attached to a steel plate, b) Stress strain curve used in the impactor simulation.

The aluminum was given a density of 2366 kg/m³ in order to replicate the mass used in the experiment (1.895 kg), and an initial velocity of 3.2 m/s was applied to match the experiment. The developed foam

material model was applied to a 1” thick rectangular part, and a tied contact card (*CONTACT_TIED_SURFACE_TO_SURFACE) attached the foam to a steel plate.

3.3.2 Repositioning the Full Body Model to the Initial Position of a PMHS, Modelling Head-first Impact to a Full Body Model, and Comparing the Response with the Experiments from (Kerrigan, et al., 2014)

Researchers (Kerrigan, et al., 2014) inverted and repositioned four PMHSs to have a torso angle of 20-27°, their C7/T1 disc rotated 25° to the ground, and a “level” eye line (Frankfort plane). The GHBMC neutral posture (FB_{NEUT}) had different torso angles, with closer C7-T1 angles, and Frankfort planes (Fig. 29) compared to experimental PMHS tests. However, the C7-T1 angle in the experiments and the Frankfort planes were meant to maintain a resting neck lordosis, while the GHBMC M50 (FB_{NEUT}) had a comparable posture to seated test subjects when neck length was accounted for (Corrales & Cronin, 2021). Therefore, the FB_{NEUT} head and neck posture was comparable with the experiments, but not the torso angle. To match the torso angle with experiments (Kerrigan, et al., 2014), the FB_{FLX} model from Section 3.2.1 was rotated 30° about the positive y-axis (FB_{FLX}+30°), which allowed the torso angle to be close to the experimental bounds, while allowing the Frankfort plane to be 4.6° away from the target of 0° (“level”). The C7-T1 angle was 16.9° away from the target, but fitting all three of the target values would have required extensive morphing because the GHBMC M50 is a subject specific model. As a result, the cervical spine was in flexion rather than extension prior to impact, which was expected to change the T1 rotation and possibly the element erosion.

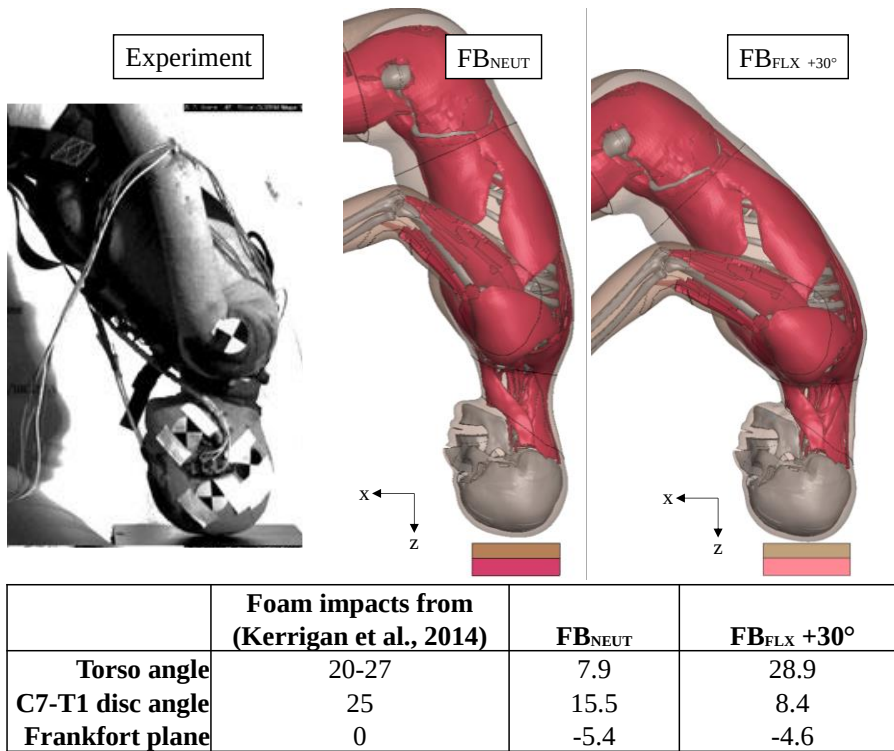


Figure 29: Image of HFI experiment with a post mortem human subject (PMHS) before impact from (Kerrigan, et al., 2014) (top left), a neutral postured GHBMC M50 pre-HFI (top middle), and a flexed posture GHBMC M50 rotated 30° pre-HFI (top right). Additionally, a table comparing the postural markers from the experiment from (Kerrigan, et al., 2014) and the GHBMC (bottom).

The postures from Section 3.3.2 (FB_{NEUT} and FB_{FLX+30°}) were impacted against the foam and plate modelled in 3.3.1 (Fig. 29). These impacts were at a velocity of 3.05 and 3.59 m/s, representing the velocities used in (Kerrigan, et al., 2014).

In these simulations, the head to foam contact force was monitored and compared with experimental forces. T1 rotation was measured against the global reference frame using the most anterior and most posterior corners of the superior endplate and the global ‘X’ axis. Similar to Section 3.2.4, the volume and location of eroded bone elements were tracked, which included the trabecular and cortical elements in the vertebrae.

Chapter 4

Results

Similar to methods, results in this document have been divided into three Studies, with Section 4.1 corresponding to the Study 1, Section 4.2 corresponding to the Study 2, and 4.3 corresponding to Study 3.

4.1 Study 1: Isolated Motion Segment Model to Investigate Response Under Combined Compression and Anterior-Posterior Bending and Relationship to Fracture-Dislocation

4.1.1 Obtaining the Fractures Associated with Distractive Flexion Fracture-Dislocations Using Different Ratios of Linearly Increasing Compressive Force and Bending Moment

All segments either rotated anteriorly or posteriorly before failing. The segments with pure flexion, or with more flexion than compression rotated anteriorly, while segments with pure compression, or with more compression than flexion rotated posteriorly (Fig. 30).

In pure flexion bending moment loading, all segments rotated in flexion until their interspinous ligaments began to fail. With increased bending moment, the capsular ligaments and ligamentum flavum also began to fail. Finally, between 20.0 and 21.0 Nm (for all except C6-C7, which had vertebral element erosion at 53.4 Nm), there was significant tearing in the ligaments, and the vertebrae rotated rapidly. During the rotation, there was often element erosion in the anterior-inferior corner of the superior vertebral body (the anterior vertebral margin). After this, the superior vertebra increased its rotational velocity (due to higher bending moment and less supporting structures), and translated away from the inferior vertebra.

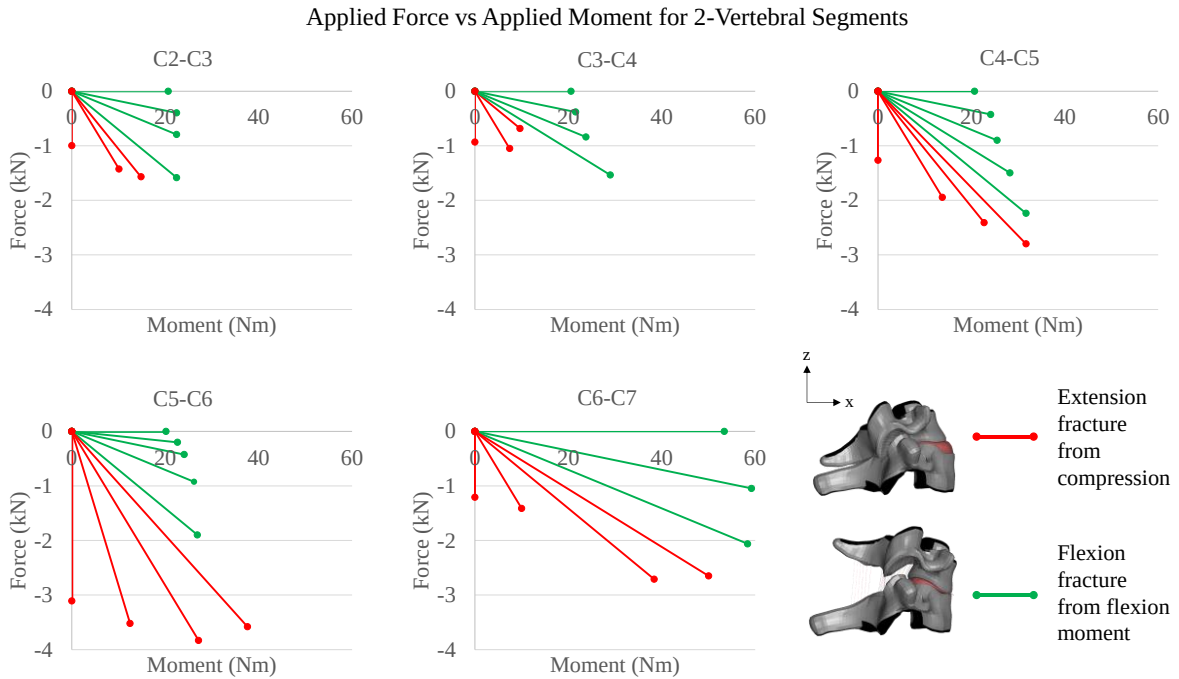


Figure 30: Effect of combined flexion and compression loading on fracture initiation for C2-C7 2-vertebrae segments. Each line (green or red) represents one simulation with linearly increasing applied compression force and flexion moment, and the end of the line represents the prediction of initiation of fracture. Red lines are used to show segments that rotated posteriorly and had fracture initiating due to extension bending, and green lines are used to show segments that rotated anteriorly with fracture initiating due to flexion bending.

In pure compression, all segments rotated in extension, due to the inclination of facet joint. Most segments rotated and were compressed until 0.93-1.26 kN, when the posterior facets began eroding. The only exception was C5-C6, which took 3.11 kN before the transverse processes and spinous process began eroding (also due to extension bending).

When the flexion bending moment and compression loading were combined, the vertebrae either rotated in flexion or extension, and had similar failure mechanisms to pure flexion or pure compression, depending on which direction they rotated (Fig. 31). Consequently, it took more compressive force for failure to occur in combined loading rather than pure bending moment or compression loading.

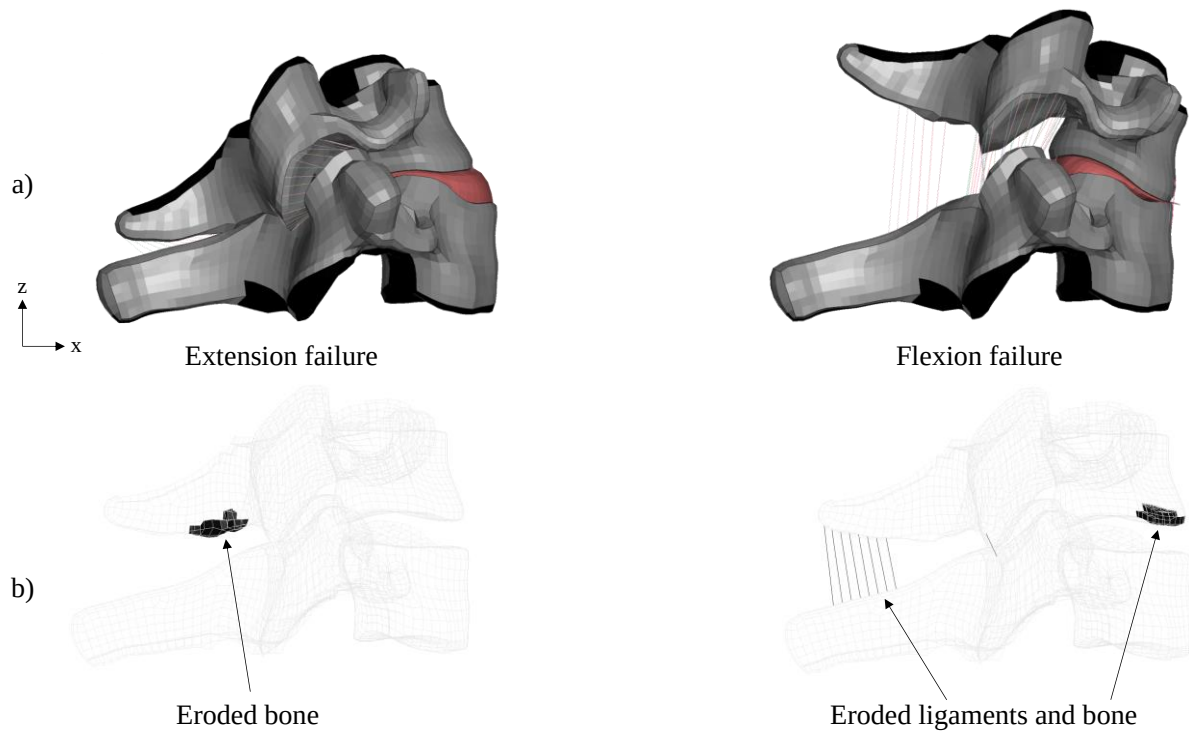


Figure 31: Extension (left) and flexion (right) failure in a C6-C7 force- and moment-based rotation with a) showing the rotation of the vertebrae with failed ligaments blanked out, and b) showing the eroded elements bolded in black.

4.1.2 Dislocating a Segment Model of 2 Vertebrae using Pre-Defined Motion, Based on Experiments from (Ivancic, et al., 2008)

The interspinous ligament begins to fail after 11.4°-13.8° of flexion rotation. At 15.6°-16.8° of rotation, more interspinous ligament had failed, and the capsular ligaments began to fail. At 19.5°-24.6°, the ligamentum flavum and the posterior longitudinal ligament began to fail. At 24.0°-29.5°, the superior-anterior edge of the inferior vertebra had element erosion. By the 30° of rotation, most of the interspinous ligament, ligamentum flavum, posterior capsular ligaments, and posterior longitudinal ligaments had failed. The progression of no failure at 10°, to some posterior ligament at 20°, to more posterior ligaments plus bone erosion at 30° is shown (Fig. 32), and the failure progression for all three vertebrae is listed (Table 2).

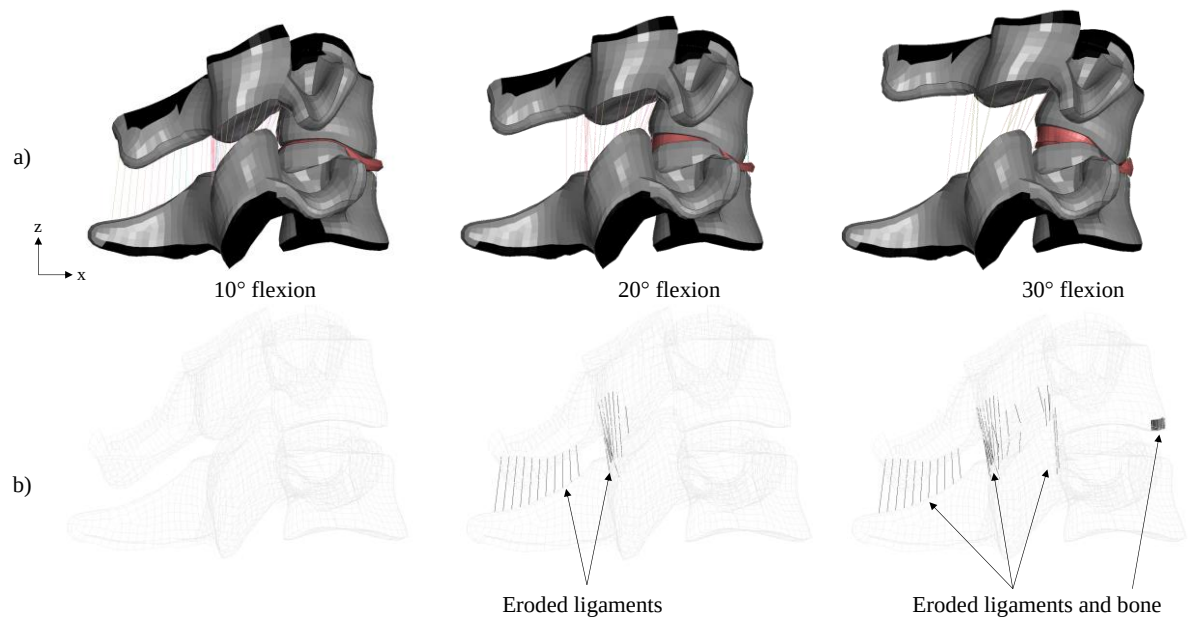


Figure 32: Failure progression in a C5-C6 displacement-based rotation showing 10°, 20°, and 30° of flexion, with a) showing the rotation of the vertebrae with failed ligaments blanked out, and b) showing the eroded elements bolded in black without rotation. As shown, there is: no element erosion from 10° of rotation; capsular ligaments, interspinous ligaments, and ligamentum flavum failure at 20°; and interspinous ligament, ligamentum flavum, posterior capsular ligaments, and posterior longitudinal ligaments failure, as well as vertebral body element erosion at 30°.

Table 2: Failure initiation angles for C4-C5, C5-C6, and C6-C7 vertebral motion segments during flexion rotation of displacement-controlled motion.

	Interspinous ligament begins failing	Capsular ligaments begin failing	Ligamentum flavum begin failing	Posterior longitudinal ligaments begin failing	Vertebral body element erosion initiates
C4-C5	13.8°	16.8°	21.0°	24.6°	29.5°
C5-C6	11.4°	15.6°	19.5°	21.6°	28.2°
C6-C7	12.0°	16.8°	23.4°	20.4°	24.0°

As can be expected from the 30° simulations, the 10° simulation had no failures during rotation, and the 20° had significant failure in the capsular ligaments, interspinous ligaments, and ligamentum flavum. After 30° rotation, the shear and subsequent anterior rotation allowed for both facets dislocate anterior to the inferior facets (Fig. 33c). Simulations that had 20° rotation only had enough clearance for one facet to be in a position anterior to the inferior facet (Fig. 33b). Finally, simulations with 10° of rotation did not have enough clearance for the superior facets to climb over the inferior ones. As a result, the superior facet failed in compression (elements were eroded) (Fig. 33a). A typical sequence of flexion, shear, extension for segments under 10°, 20°, and 30° is shown (Fig. 33), with 10° rotation segments incurring failure in the facet, and 20°, and 30° dislocating 1-2 facets.

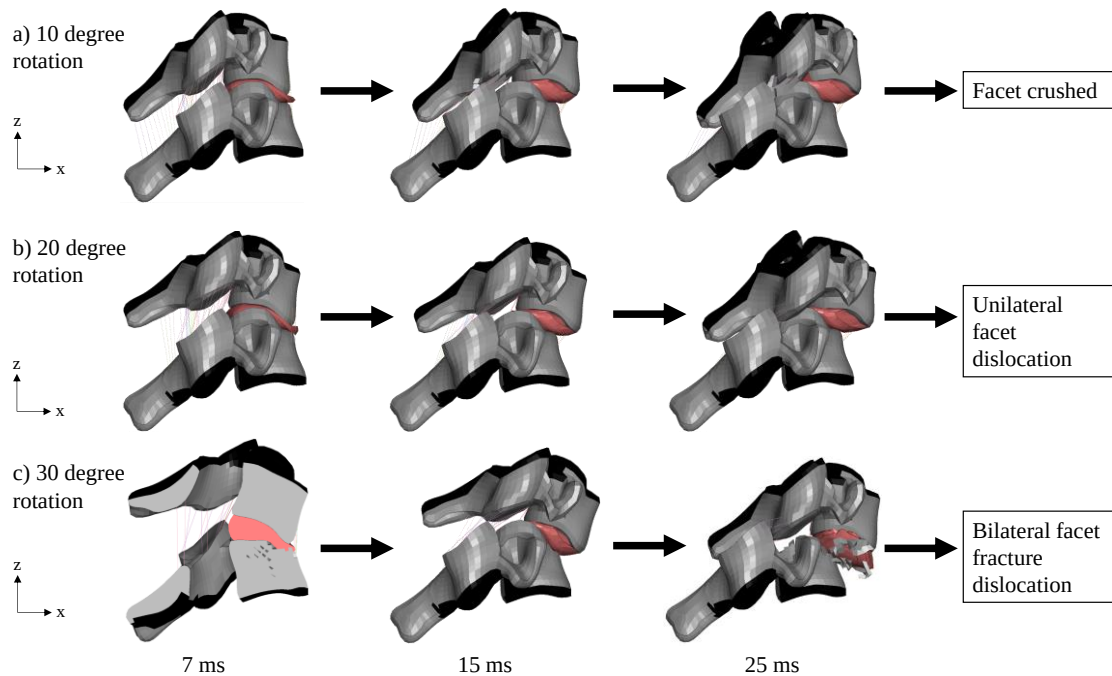


Figure 33: Progression of displacement-controlled dislocation (C4-C5 segment) for a) 10°, b) 20°, and c) 30° of initial rotation. Motion segments are shown at their specified flexion rotation at 7 ms (left), at the end of applied anterior shear at 15 ms (middle), and after extension rotation at 25 ms (right).

The amount of flexion (10°, 20°, or 30°) influenced peak flexion bending moment applied at the boundary (Fig. 34a). Flexion bending moments for 10° rotations were low (10.2-13.4 Nm) compared

to 20° and 30° rotations (21-22 Nm for C4-C6 and 30-31 Nm for C6-C7). The shear forces during the shearing motion of vertebrae (Fig. 34b) ranged between 1.1 kN and 1.9 kN. For C4-C5, the shear force increased slightly with increasing vertebral rotation (1.4 kN for 10°, 1.6 kN for 20°, and 1.8 kN for 30°). C5-C6 and C6-C7 had higher shearing forces at 10° (1.8 kN and 1.9kN respectively), compared to those at 20° (1.1 kN and 1.5 kN respectively), and 30° (1.2 kN and 1.4 kN respectively). The extension bending moments (Fig. 34c) ranged from 18.3 Nm to 30.0 Nm, and with the exception of C4-C5, less rotation resulted in higher bending moments.

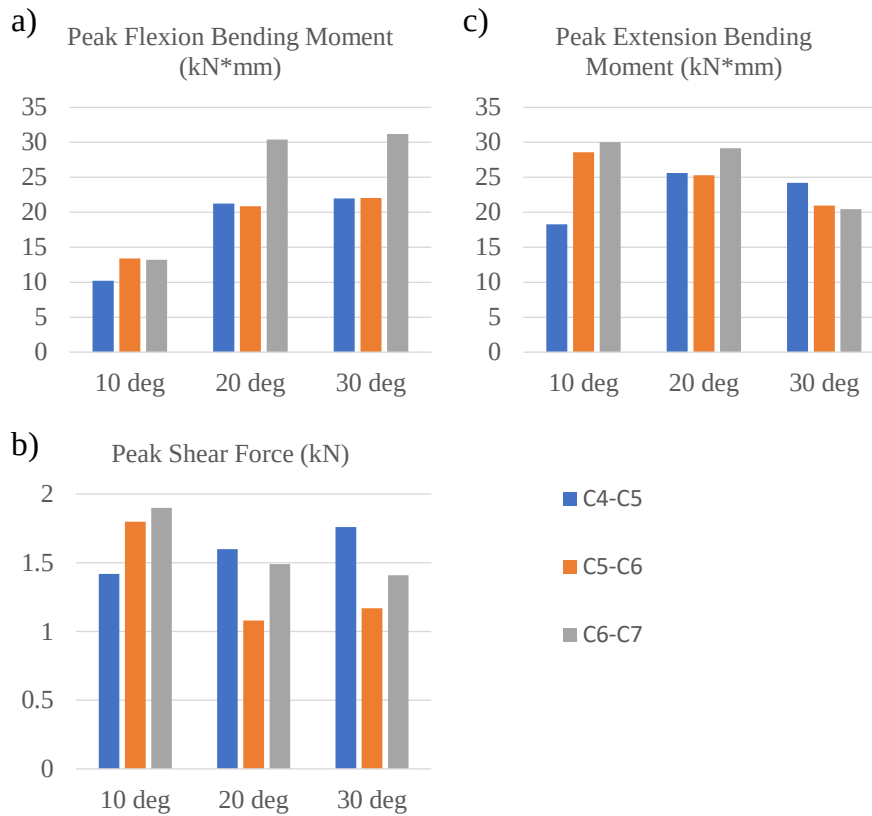


Figure 34: a) Peak flexion bending moment, b) peak extension bending moment, and c) peak applied shear force, all for simulations dislocating a lower cervical spinal segment model with 10°, 20°, and 30° pre-dislocation rotation.

Increasing from 10° to 30° of rotation resulted in more predicted tissue failure (Fig. 32) in the model. Consequently, during the shearing and extension rotation parts of the simulation, the different rotations led to different outcomes. For the 10° rotation, there was no failure in the models during the flexion rotation. The facets were crushed during shearing and extension rotation due to the ligaments being intact providing a constraint during the motion. When the flexion rotation was 20°, there was some interspinous ligament failure during flexion rotation. During shearing, the ligament failure allowed one superior facet to move anterior to the inferior facet. During the extension rotation, this produced a unilateral facet dislocation. Finally, in simulations with 30° of intervertebral rotation, there was an almost complete rupture of the interspinous ligament, in addition to element erosion of the anterior-superior edge of the inferior vertebrae. During shearing, one (C6-C7) or two (C4-C5 and C5-C6) facets were moved anterior to the inferior facet, which resulted in unilateral (C6-C7) or bilateral (C4-C5 and C5-C6) facet dislocations during subsequent extension rotation.

4.2 Study 2: Importance of Neck Boundary Condition and Posture on the Cervical Spine Response During a Head-First Impact

4.2.1 Repositioning of the Neck of a Human Body Model (HBM) from a Neutral Posture to a Flexed and Extended Posture

The positions of the vertebrae for the FLEX and EXT postures were within 0.9 microns of the target positions determined from the Cervical Spine Posture Predictor, measured at the superior anterior-posterior corners of the vertebral bodies and at the chin and infraorbital corner of the skull. The mesh of the repostured models all met the warpage (< 50°), aspect ratio (< 8), skew (< 70°), and Jacobian (> 0.4) targets for mesh quality.

4.2.2 Assessment of the Isolated Head and Neck Torso Surrogate Mass (TSM) Model Under Head-first Impact (HFI) in Three Postures

In the TSM_{NEUT} impact case, the initial contact of the head with the plate led to a rapid increase in force (Fig. 35) up to 12.2 kN in 2 ms (all simulations had this rapid initial head force), followed by a rapid decrease in force to 0 kN, and a rebound resulting in a second peak force (4.74 kN) occurring 7 ms after the first. Upon the initial impact, the cervical spine went into third order bending (cervical spine kinematics are plotted in the Appendix), with the upper and lower cervical spine increasing in lordosis, and the middle cervical spine (C4-C6) curvature becoming flexed. During this higher order bending,

peak axial force (Fig. 36) and flexion bending moment at the C7 cross-section were 2.93 kN and 28.1 Nm, respectively. Later in time, the flexed C4-C6 area of the spine transitioned back to a lordotic curvature. During the second head impact following rebound, the spine buckled in a first order mode (C2-C6 lordosis), and the axial force and flexion bending moment at C7 had larger maxima than the initial head impact (3.01 kN, 41.6 Nm).

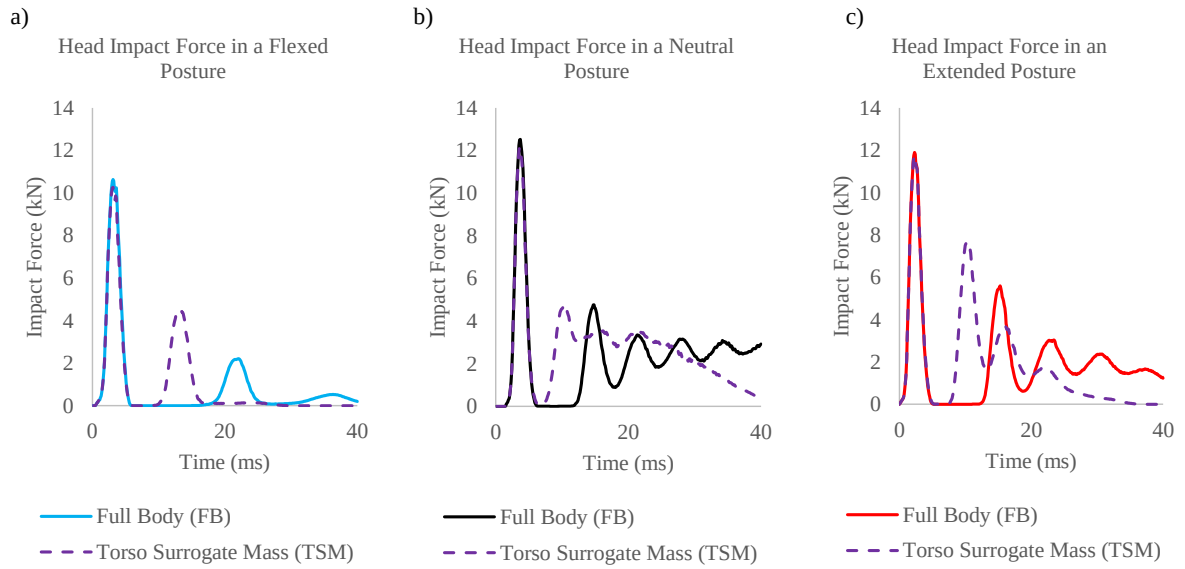


Figure 35: Head impact forces for 3.14 m/s simulations comparing full body (FB, solid lines) and torso surrogate mass (TSM, dashed lines) simulations, for a) a flexed posture, b) a neutral posture, and c) an extended posture. The first peak contains both FB and TSM overlapped traces.

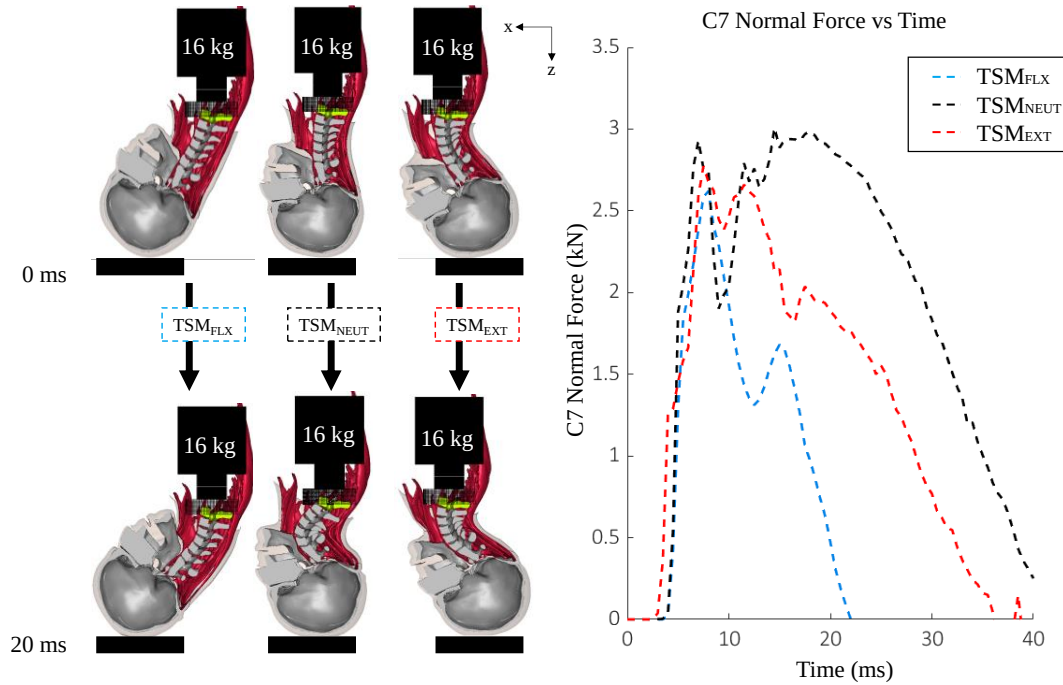


Figure 36: Mid-sagittal sections of the three TSM (torso surrogate mass) simulations (TSM_{FLX}, TSM_{NEUT}, TSM_{EXT}) at 0 and 20 ms (left) in addition to the C7 cross-section normal force for 3.14 m/s impacts (right). At 20 ms, the simulations typically had maximum intervertebral rotations, therefore the extent of the cervical bending and differences between postures could be demonstrated.

When element erosion (Fig. 37) was included for the TSM_{NEUT} model, the rapid increase in compressive force from the third order bending caused large amounts of erosion in the upper cervical spine. The element erosion began in the pedicles of C2, and spread to the inferior vertebral body, superior facets, odontoid, and transverse process by 5 ms after first head contact. At 6 ms, the C1 posterior arch began eroding. At 9 ms, ~25% of C2 solid elements were eroded, and C3 vertebral body began to erode as well. At 35 ms, there were so many eroded elements in C1-C2-C3 that adjacent soft tissue elements became loaded unevenly by the remaining elements, resulting in instabilities in the simulation, and an error termination due to out-of-range forces. At this time, there was a volume of 8,810 mm³ of eroded solid vertebral elements in C1-C2-C3. When the simulation had an error termination, the kinetic energy was 3% of the initial kinetic energy.

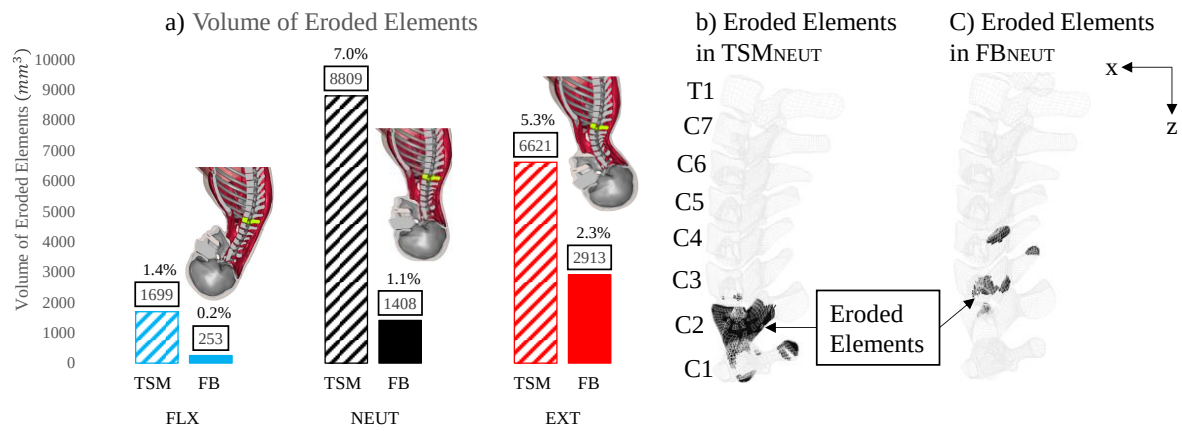


Figure 37: a) Volume of eroded solid vertebral elements for impacts at 3.14 m/s for FLX (flexed posture, left), NEUT (neutral posture, middle) and EXT (extended posture, right) simulations, comparing TSM (torso surrogate mass) to FB (full body) models with the percent of the full spine eroded included, b) eroded elements in TSM_{NEUT} colored black with C1-T1 labelled, and c) eroded elements in FB_{NEUT} colored black, all at 40 ms after initial head impact or the simulation termination.

The cervical spine of TSM_{EXT} started with first order lordotic bending upon initial impact (11.8 kN head impact force), and maintained the first order lordotic bending throughout the entire simulation, resulting in a maximum of 2.77 kN axial force and 53.3 Nm flexion bending moment at the C7 vertebra. A head rebound phase was observed in this impact with 7.67 kN of head force occurring 8 ms after the first head peak, resulting in similar axial loading at C7, but an extension bending moment due to the extension of C6-C7-T1 (2.66 kN and 72.8 Nm).

Like TSM_{NEUT}, TSM_{EXT} also had significant element erosion in the upper cervical spine. The element erosion began for TSM_{EXT} in the C4 facets and the C7 spinous process (at 5 ms), followed by C3 lamina (initiating at 6 ms). By 7 ms, the C2 inferior facets began eroding as well, followed by C1 anterior arch and the C2 inferior vertebral body. Similar to TSM_{NEUT}, by 32 ms, there were so many eroded elements in C2 that adjacent soft tissue elements became loaded unevenly by remaining elements, resulting in an error termination due to out-of-range forces. At this time, there was 6,620 mm³ of eroded vertebral

elements. When the simulation had an error termination, the kinetic energy was 22% of the initial kinetic energy.

TSM_{FLX} started and remained in first order flexion bending. During the first head impact (10.2 kN head contact force), 2.62 kN of axial force, and 18.3 Nm of flexion bending moment were measured at the C7 vertebra. After the second head impact (the second peak head force was 4.54 kN, 10 ms after the first peak), forces and moments at C7 decreased (1.68 kN and 15.5 Nm) due to the head CG being increasingly misaligned with T1 (Fig. 36). As flexion in the neck increased, the bending moment increased to a peak of 59.9 Nm at 39 ms.

When element erosion was included in TSM_{FLX}, the C4 and C3 spinous processes and C2 pedicles began eroding at 7 ms. At 8 ms, the C6 pedicles also began eroding. By 40 ms, there were spinous process elements eroded in C2-C3-C4-C5-C6, which combined with the C2 and C6 pedicles, amounted to a volume of 1,700 mm³ of vertebral elements.

Aside from the aforementioned TSM_{NEUT} and TSM_{EXT} cases with element erosion included, the TSM HFI loading was numerically stable and the analyses all ran to completion (40 ms), taking approximately 3 hours using 32 cores with Massive Parallel Processing (MPP) on an Intel Xeon 64-Bit Processor with 2200 Mhz clock speed. Note that the TSM_{NEUT} and TSM_{EXT} cases with element erosion included were only used to compare the amount of element erosion, all other comparisons were done with the simulations with no erosion. The two simulations with error terminations had higher element erosions at their terminations (32 and 35 ms) than all other simulations at 40 ms.

4.2.3 Assessment of the Full Body (FB) Model Under Head-first Impact (HFI) in Three Postures

In FB_{NEUT} (neutral posture head-first impact simulation), the HBM head contacted the rigid plate with a peak load of 12.5 kN, rebounded and hit again with 4.76 kN, 11 ms after the first contact, remaining in contact with the plate for the remainder of the simulation, up to 40 ms. All of the FB head rebound times were increased in FB compared to TSM (Fig. 35) because the less constrained FB spine allowed for more overall head motion between impacts. Upon the first impact, the spine went briefly into third order bending (as described for TSM_{NEUT}, but to a lesser extent), which caused the C7 normal force to reach 1.17 kN (Fig. 38). By 8 ms, the spine moved into lordosis, which it remained in for the remainder of the simulation. The second head impact and cervical spine lordosis caused the C7 axial force to reach

1.50 kN. The C7 flexion bending moment was 7.61 Nm at the first impact, and 7.89 Nm by the second head impact.

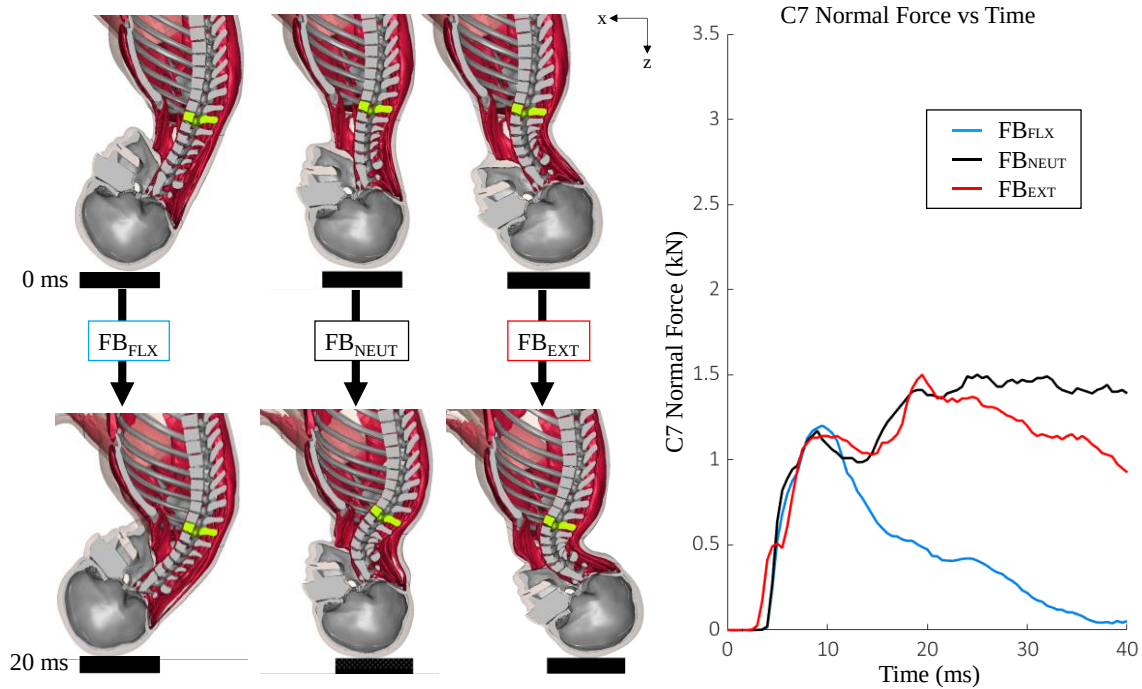


Figure 38: Mid-sagittal sections of the three FB (full body) simulations (FB_{FLX} , FB_{NEUT} , FB_{EXT}) at 0 and 20 ms (left) in addition to the C7 cross-section normal force for 3.14 m/s impacts (right). At 20 ms, the simulations typically had maximum intervertebral rotations, therefore the extent of the cervical bending and differences between postures could be demonstrated.

Between the first and second head impacts, the thoracic spine of FB_{NEUT} underwent flexion bending. In the middle of the cervical lordosis and thoracic flexion, T1 translated 9.41 mm anteriorly, and rotated 34.8° in the sagittal plane relative to the global coordinate system.

In FB_{NEUT} simulations with element erosion (Fig. 37), contact between the C3 facets and the C2 transverse process at 26 ms caused small amounts of elements to erode, which was caused by relative C2-C3 extension bending. By 28 ms, there was erosion in the C4 inferior facet and the C3 spinous process. At 29 ms, C3 pedicles began eroding, and at 32 ms, the C3 facets began eroding. By 40 ms, the erosion in these areas represented a combined volume of $1,410 \text{ mm}^3$.

The first head impact of FB_{EXT} resulted in a contact force of 11.9 kN, and 12 ms later, the rebound resulted in 5.59 kN of contact force. Cervical spine lordosis was slightly lower than FB_{NEUT} from C2-C4, and slightly higher in C4-C7, but was lordotic throughout the simulation. Lordotic kinematics resulted in similar normal forces to TSM_{NEUT} (1.14 kN at C7 during the first head impact, the maximum value occurring after the second impact at 1.50 kN). The peak C7 flexion bending moment was 3.37 Nm during the first impact, and due to C6-C7-T1 extension, there was a maximum of 7.76 Nm of extension bending during the second head impact. T1 translated 25.6 mm anteriorly and rotated 10.3° due to thoracic spine flexion and cervical spine lordosis.

FB_{EXT} had more predicted element erosion than FB_{NEUT} and FB_{FLX} . While small amounts of erosion were observed early in the simulation, at 17 ms, C3 began eroding in the facet joint region, which contained the majority of the element erosion at 40 ms. Additionally, there were small amounts of element erosion in C4 superior facets, C4 anterior vertebral body, C3 spinous process, and the C2 pedicle. In total, $2,910 \text{ mm}^3$ of vertebral bone was eroded.

FB_{FLX} had different results than FB_{NEUT} and FB_{EXT} . Due to the initial flexed posture of the neck, after the first impact of the head to the plate (10.6 kN of head impact force), the spine increased in flexion, and the head rotated anteriorly. Following the flexion of the head and neck, the second impact resulted in 2.21 kN, 19 ms after the first impact and the occipital bone of the skull contacted the plate. Moreover, throughout the simulation, the head CG alignment increasingly moved away from T1 (Fig. 38). During the first head impact, there was 1.20 kN of C7 normal force, followed by a gradual drop to 0.05 kN at 40 ms. Similarly, the first head impact of FB_{FLX} resulted in 9.39 Nm of flexion bending moment in C7, and the subsequent bending in the neck increased the flexion bending moment to 11.7 Nm. T1 translated 19.5 mm posteriorly, and rotated 16.6° due to thoracic and cervical spine flexion.

Finally, FB_{FLX} also had a different element erosion pattern than FB_{NEUT} and FB_{EXT} when element erosion was included. At 26 ms, C3 and C4 spinous process elements began eroding, followed by C5-C6 spinous process elements at 39 ms. In total, 253 mm^3 of vertebral bone was eroded at 40 ms.

The FB HFI loading was numerically stable, and the analyses all ran to completion (40 ms), taking approximately 12 hours using 32 cores with Massive Parallel Processing (MPP) on an Intel Xeon 64-Bit Processor with 2200 MHz clock speed.

4.3 Study 3: Assessing the Neck Response of the Full Human Body Model with Whole Body PMHS Experimental Data

4.3.1 Evaluating the Numerically Modelled the Experimental Impact Surface by Comparing Force-Time, and Displacement-Time Curves for an Aluminum Semi-Sphere Impact test

When modelling impact between a semi-spherical impactor and foam mounted on steel, the peak compression of the foam was 20.7 mm, which occurred 8.4 ms after impact (Fig. 39). The foam compression generated a peak reaction force of 2.07 kN, which resulted in the impactor rebounding off the foam.

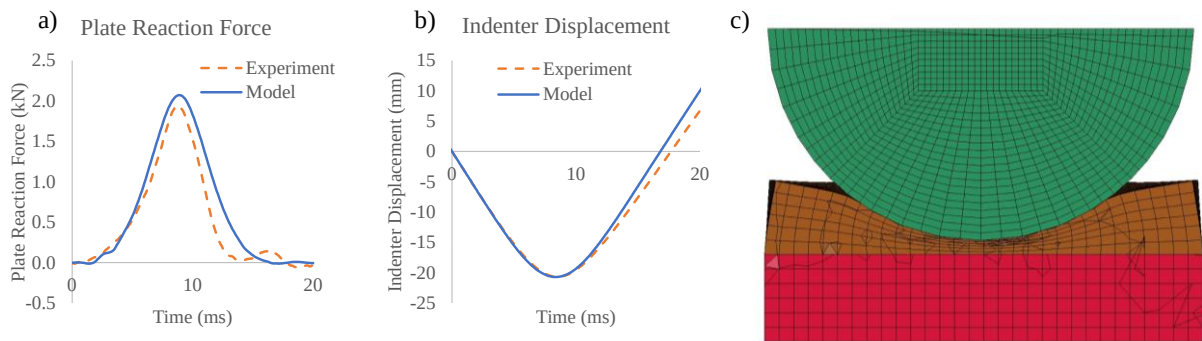


Figure 39: Measurements taken during simulated aluminum impact to foam modelled by *MAT_LOW_DENSITY_FOAM including: a) the reaction force measured by the plate (kN) vs time (ms), b) the displacement of the indenter (mm) vs time (ms), and c) a snapshot of a mid-sagittal section of the plate, foam, and indenter during peak deflection (8.4 ms).

4.3.2 Measuring the Head Force, T1 Rotation, and Element Erosion of a Flexed and Rotated Full Body Model Impacting PVC Foam at Two Velocities

Impacting GHBMCM50 FB_{FLX}+30° to the PVC foam pad resulted in peak head reaction forces of 4.27 and 5.96 kN for 3.05 and 3.59 m/s impacts, respectively (Fig. 40). Neither impact velocity resulted in full unloading and rebounding of the head seen in rigid plate impacts from Section 4.2, with peak unloading occurring at 24.4 ms with 1.15 kN (3.05 m/s impact) and 22.3 ms with 0.81 kN (3.59 m/s

impact) of head impact force. After unloading, the head was reloaded at to 2.32 kN at 38 ms and 33.9 kN at 33.5 ms (for 3.05 and 3.59 m/s impacts respectively). Both models unloaded towards 0 kN for the remaining of the 100 ms impact (Fig. 40).

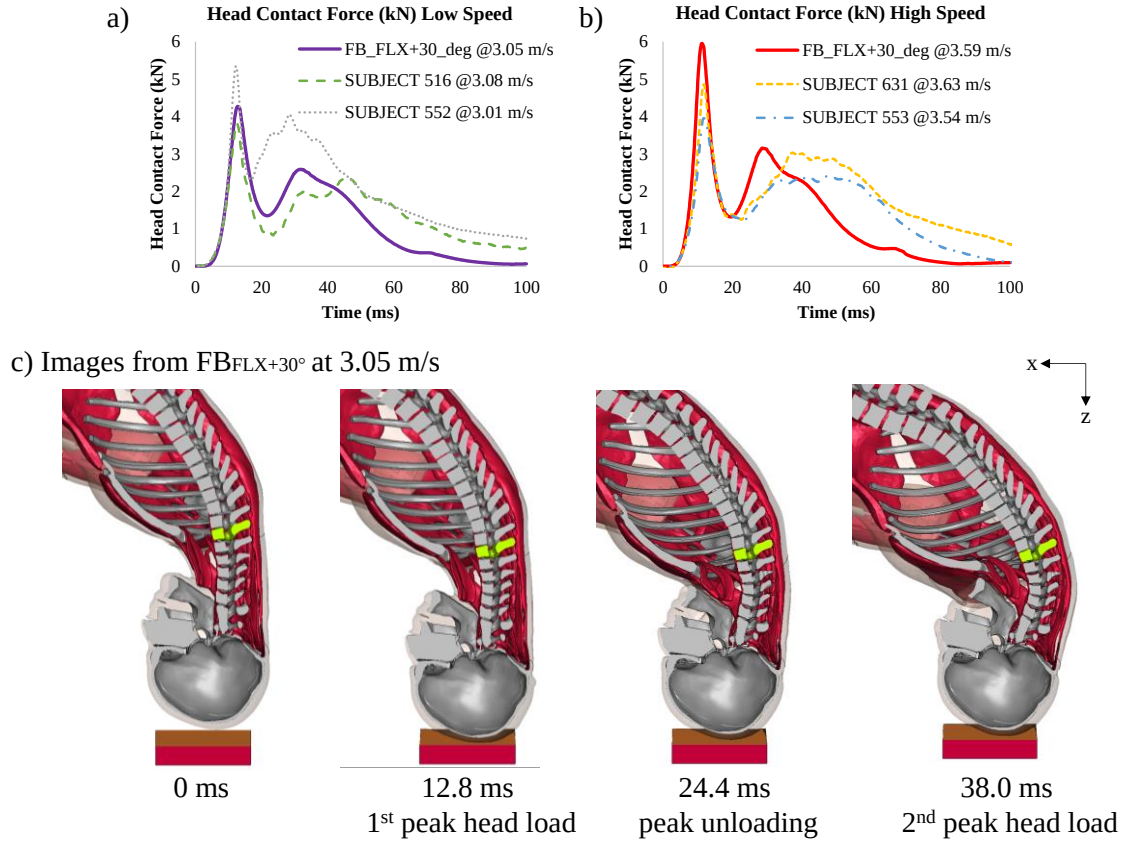


Figure 40: Head contact forces at a) low and b) high speeds for FB_{FLX+30} foam impact simulations compared to cadaveric subjects from (Kerrigan, et al., 2014), and c) mid-sagittal sections from pre-impact, first peak head load, peak unloading, and second peak head loading during a low speed simulation.

For FB_{FLX+30} impacts, the flexed thoracic and cervical spines increased their flexion (kyphosis) throughout the impact (Fig. 40c). During the first head impact, there was an initial wave of compression, wherein T1 rotated slightly posteriorly (3.6° for the 3.05 m/s impact, and 4.0° for the 3.59 m/s impact). After the first head impact, the flexion of the cervical and thoracic spine resulted in T1 rotating

anteriorly until the end of the simulation (Fig. 41). This differed from the experiments due to the different initial C7-T1 angle. As a result, spinal kinematics and element erosion were different.

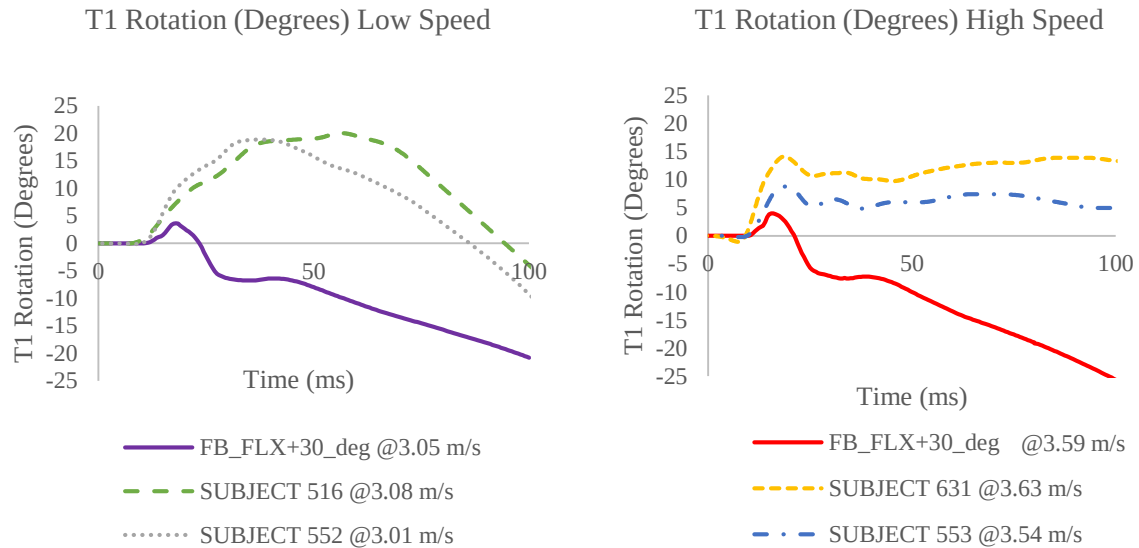


Figure 41: T1 rotation at low (left) and high (right) speeds for $FB_{FLX}+30$ foam impact simulations compared to cadaveric subjects from (Kerrigan, et al., 2014).

The impacts resulted in 1.67 and 1.98 kN of normal force measured at C7 (for 3.05 and 3.59 m/s impacts respectively), and 19.7 and 22.9 Nm of bending moment. As a result, when element erosion was included for the $FB_{FLX}+30$ foam impact simulations, there was T1-T2 vertebral body element erosion (beginning at 15 ms for the 3.05 m/s impact, and beginning at 13 ms for the 3.59 m/s impact). By 100 ms, the T1-T2 vertebral body element erosion made up 1,770 mm³ and 2034 mm³ of solid vertebral elements for 3.01 and 3.59 m/s impacts, respectively. Additionally, flexion of the cervical spine caused element erosion in the spinous processes of C2, C3, and C4. The C2, C3, C4 erosion initiated at 31.5 ms, causing 494 mm³ of elements to erode in a 3.05 m/s impact, and initiated at 29.5 ms, causing 548 mm³ of elements to erode in a 3.59 m/s impact.

Chapter 5

Discussion

5.1 Study 1: Isolated Motion Segment Model to Investigate Response Under Combined Compression and Anterior-Posterior Bending and Relationship to Fracture-Dislocation

5.1.1 Obtaining the Fractures Associated with Distractive Flexion Fracture-Dislocations Using Different Ratios of Linearly Increasing Compressive Force and Bending Moment

In pure flexion, all failure bending moments of segments (20.2-20.7 Nm) except for C6-C7 (53.4 Nm) were within the experimental range of failure moments from literature (13.9-29.5 Nm) (Nightingale, et al., 2007). Since compression was not simulated until the failure of the segment, but rather the initialization of erosion of the bone elements, only the pure compressive axial force of C5-C6 (3.11 kN) was within the experimental range (2.55-3.97 kN) determined by (Carter, Ku, Nuckley, & Ching, 2002) while the other segments had compressive force values below this range.

In combined loading, it was hypothesized that the amount of both compressive force and bending moment would decrease to cause a failure (in a linear combination). The hypothesis was based on experimental tests on porcine subjects (Prasad & Daniel, 1984) with composite axial force and flexion moment loading on the neck, resulting in a neck injury threshold known as the Nij 'Kite.' The segment level simulations in the present study had the opposite trend, wherein adding a small flexion moment to a compressive failure increased the amount of compression needed to fail the vertebra rather than decrease the compression to failure. Similarly, adding a small amount of compression to a flexion failure increased the flexion moment needed to fail the vertebra. This occurred because in force-controlled loading, the vertebrae only failed after a rotation occurred, and to rotate either in extension or flexion, the opposing force had to be overcome. The inclination of the facet joints caused compressive loading to rotate the vertebrae in extension-compression, and the lack of constraints in the simulations made pure compressive failure unlikely, and extension-compression failures more likely. Therefore, in combination loading, the two loads (flexion bending and compression) were opposed, resulting in either a failure in compression or flexion. Peak loading for both compression and flexion occurred when the opposing forces almost balanced each other out. Shown is a comparison of the Nij

‘Kite’ assumption about combined loading compared to the C6-C7 segment combined loading results (Fig. 42), demonstrating a linear trend in the Nij ‘Kite’ and a non-linear trend for the numerical data. While the tolerance of a vertebral segment and the whole neck differ, and the Nij ‘kite’ represents the whole neck, it was expected that trends at the whole neck would be present in the segment. The Nij ‘Kite’ is an injury metric for crash dummies, therefore the force measured in steel and plastic components is typically much higher than in a vertebral motion segment. Simulations with more flexion loading caused segments to fail in flexion, whereas simulations with more compression caused segments to fail in extension.

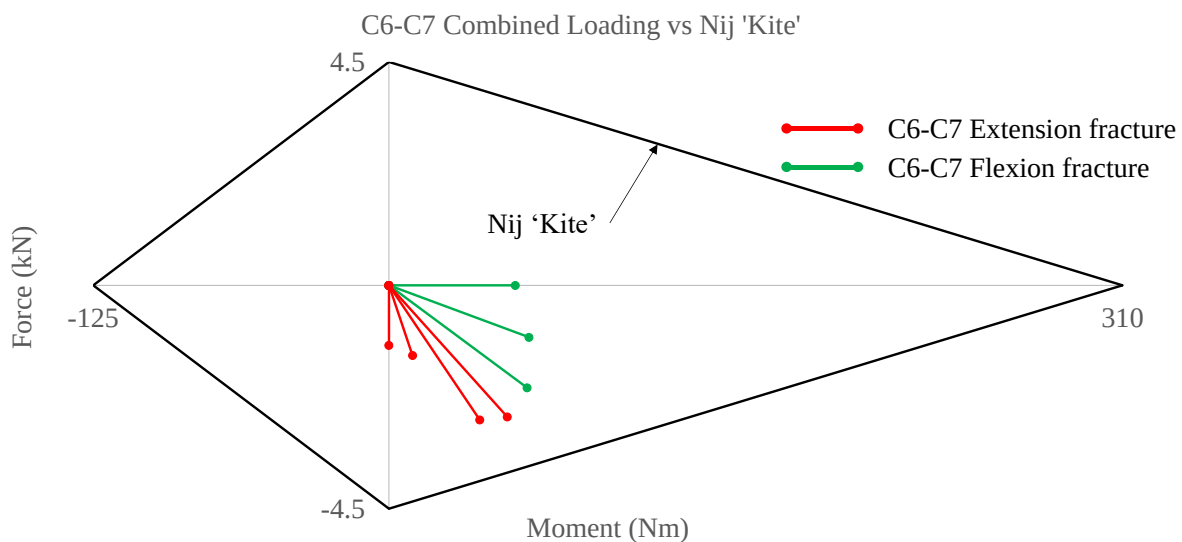


Figure 42: Simulation results of C6-C7 combined compression and flexion compared to the Nij criterion for neck injury. While the Nij predicts that combined compression-flexion loading causes failure to occur at lower values than pure compression or flexion (i.e. a linear combination), the GHBM motion segment model predicts an increase in both flexion and compression to failure during combination loading (possibly a non-linear combination). Note that the tolerance of a vertebral segment and the whole neck differ, but the trends were hypothesized to be the same.

5.1.2 Dislocating a Segment Model of 2 Vertebrae using Pre-Defined Motion, Based on Experiments from (Ivancic, et al., 2008)

During the initial flexion rotation, the 20° and 30° models for all three segment levels predicted ligament and bone failure associated with stage 1 distractive flexion fracture-dislocation, as described by Allen et al. (Allen, Ferguson, Lehmann, & O'Brien, 1982). After a shearing motion and rotating back in extension (both after the initial flexion rotation), the 20° and 30° models for all three segment levels had either one or both facets perched over the inferior facets, representing stage 2-4 distractive flexion fracture-dislocation injuries. Here, the sequence of events resulting in distractive flexion fracture-dislocation injuries has been assumed as a rotation causing ligament and sometimes bone failure, followed by a shear and then a rotation-back causing the dislocation. While this assumption may not be the sequence of events leading to injury that occurs in a head-first impact, forces and moments that obtain distractive flexion fracture-dislocations in a segment were a useful gauge to determine if loading during HFI was close to injurious.

The models with 10° of flexion rotation had element erosion in the facets due to compressive forces at the facet joints, stopping the facets from perching after the posterior rotation phase. This occurred because the 10° of flexion rotation was not sufficient to move the facets out of the way, and the shearing motion caused them to collide and predict fracture. During real-life fracture-dislocations, the facets are often fractured (Nightingale, Myers, & Yoganandan, Neck Injury Biomechanics, 2015), but the fracture model used in this study (element erosion) improperly predicts post-fracture behavior in compressive fracture by deleting the elements.

Increasing the rotational angle before shearing increased the applied flexion bending moment, as can be expected. With C5-C6 and C6-C7, increased initial rotations reduced the force required to shear the vertebrae, whereas with C4-C5 the opposite occurred, where more rotation required more force to shear the vertebrae. For C5-C6 and C6-C7, increased rotations caused more ligaments to fail during rotation which made the motion segment require less force to shear. Additionally, increased rotations allowed for the inferior facets to impede less upon the movement of the superior facets. For C4-C5, the intervertebral disc provided more shear force resistance as rotation increased.

5.2 Study 2: Importance of Neck Boundary Condition and Posture on the Cervical Spine Response During a Head-First Impact

Peak head impact forces ranged from 10.2–12.5 kN for all models, which was comparable with experimentally measured head impact forces (5.6–11.6 kN) against rigid plates (Nightingale, et al., 1997), when the forces were scaled by the head masses. In general, the impact forces were related to the head mass, and the neck engaged later in the impact. The lower head force predicted by the flexed models (TSM_{FLX} and FB_{FLX}), compared to the neutral and extended postures, was due to the increased thickness of the flesh at the point of contact in the flexed models (8 mm) compared to the neutral and extended models (6 mm for both). It is important to note that the difference (2.26 kN) in head forces between the flexed models and the other postures was low when compared to the experimental variability (Nightingale, et al., 1997) between all specimens (6 kN). After the initial peak of head force, all model configurations predicted a rebound phase where the head moved towards the cervical spine and the inertia of the thorax resulted in further compression of the neck. This was comparable to some of the reported experimental responses (Yoganandan, et al., 1986). In general, cervical spines with an initial lordosis (EXT and NEUT postures) went into lordotic bending, while the models with a flexed cervical spine posture (FLX) went into flexion bending, for both TSM and FB. This was attributed to the preferred deformation mode of the neck under compression of a lordotic initial posture being increased lordosis with increased compression due to the cervical spine morphology (facet joints and facet joint angles) and initial posture (Nightingale, et al., 1997; Saari, Itshayek, & Cripton, 2011).

The FB models had approximately half the neck force and moment compared to the TSM models regardless of impact velocity and posture. This was explained by the high compliance of the full body, in particular the thorax, when compared to the low compliance of the rigid lateral and anterior-posterior constrained TSM boundary condition. C7 axial forces in 3.14 m/s TSM simulations (2.62–3.01 kN) were comparable with T1 forces measured in experimental head-first impacts (Nightingale, et al., 1997) to a rigid plate using a TSM boundary condition (1.97–4.19 kN).

In all FB simulations, the thoracic spine deformed to an increased flexion bending, T1 rotated and translated, and the soft tissues in the torso deformed, all contributing to energy dissipation leading to lower neck forces. In part, the higher dissipated energy in the FB models when compared to the TSM models can be quantified as the internal energy of the tissues within the head and neck. In FB HFI, less energy was absorbed by the head and neck tissues relative to the TSM models, demonstrated in plots

of the internal (strain) energy of all elements in the head and neck (Fig. 43). The reduced internal energy in the head and neck tissues for the FB models was attributed to a significant amount of the impact energy converted into internal energy in the thorax and extremities. Similar to the initial impact force, the initial internal energy was associated with the head mass impacting the plate such that the initial energy at each impact velocity was similar for all load cases. The 3.14 m/s simulations had ~15 J of internal energy (~20.5 J for 3.60 m/s, and ~6.5 J for 2.00 m/s) due to the initial head impact to the plate. However, the internal energy in the neck was lower in FB simulations than TSM, and this energy occurred later in time.

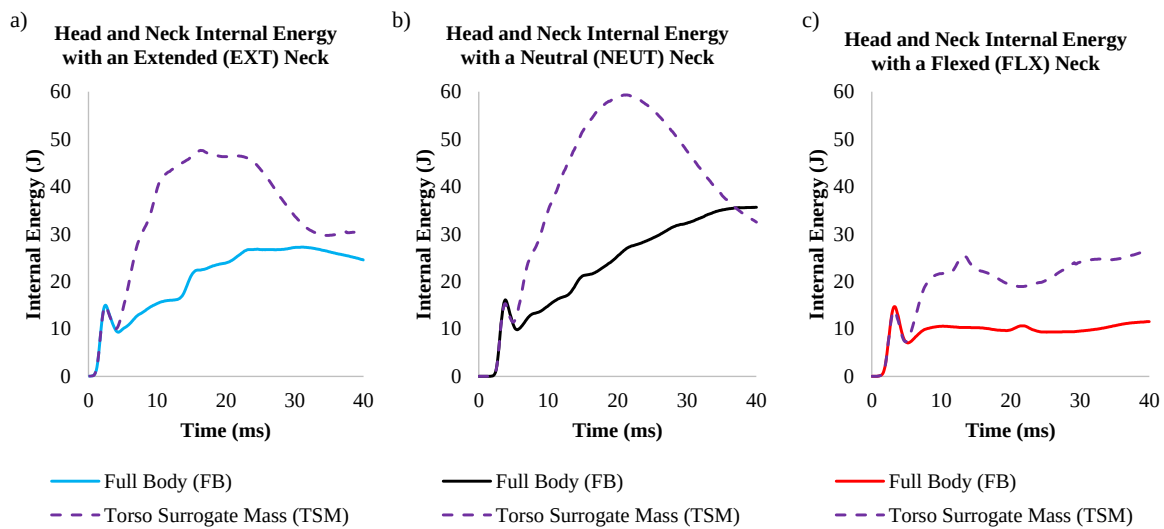


Figure 43: Internal energy in the head and neck of 3.14 m/s simulations comparing FB (full body) and TSM (torso surrogate mass) simulations in three postures: a) a flexed neck (FLX), b) a neutral neck (NEUT), and c) an extended neck (EXT).

After the initial peak of neck force, the FB_{FLX} and TSM_{FLX} models predicted unloading of the neck through rotation and anterior translation of the head. In contrast, the FB_{NEUT} , FB_{EXT} , TSM_{NEUT} , and TSM_{EXT} models predicted a sustained high load after the initial peak on the neck throughout the impact owing to the increased lordotic buckling, and the head not escaping the loading.

In simulations with element erosion included, TSM had significantly more vertebral erosion than FB in all postures (Fig. 37), and at any given velocity. The increased erosion indicated mechanical overload failure of these elements and was attributed to the higher loading in the neck for the TSM simulations,

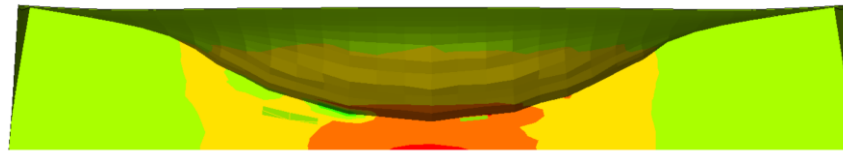
which was due to the T1 constraint imposed by the TSM. To investigate further, simulations of TSM impacts without translational or rotational constraints imposed were undertaken, and demonstrated that the TSM was unable to rotate and translate within the short duration of the impact (<40 ms) still leading to high forces in the neck. Therefore, the inertia of the added mass of the TSM constrained to T1 resulted in high neck forces and significant element erosion, regardless of external constraints on the mass. The early element erosion in TSM simulations also occurred in low speed (2.00 m/s) impacts. Conventionally, full body head-first impacts are expected to induce spinal fractures at velocities greater or equal to 3.14 m/s (McElhaney, Snyder, States, & Gabrielsen, 1979). Having significant element erosion at 2.00 m/s in impacts to TSM (in all postures simulated) may indicate that TSM over-predicts axial loading in the neck in HFI. The element erosion observed in 2.00 m/s FB_{NEUT} (144 mm³ of vertebral elements) simulations was comparatively insignificant to that in the 2.00 m/s TSM_{NEUT} simulation (2,010 mm³ of vertebral elements), with FB_{NEUT} aligning more closely with convention (McElhaney, Snyder, States, & Gabrielsen, 1979). TSM_{NEUT} having significant C1, C2 and C3 element erosion agreed with experimental HFI to a rigid plate and a TSM in a neutral posture (Nightingale, et al., 1997), which reported injuries at all cervical vertebral levels, with 4 out of 5 injured specimens having fractures in C1, C2, or C3.

5.3 Study 3: Assessing the Neck Response of the Full Human Body Model with Whole Body PMHS Experimental Data

Using the approximate stress-strain curves for a closed-cell PVC foam, the force-time and the displacement-time plots in the spherical impactor simulation had good agreement with the experiment. The timings of the peak forces (8.9 ms for the model, and 8.8 ms for the experiment), were almost identical, and the values of the peak forces (2.07 kN for the model, and 1.94 kN for the experiment), had a 7% difference. Similarly, the peak displacements (20.7 mm for both the model and the experiment) occurred at almost the same time (8.8 ms for the model, and 8.5 ms for the experiment). Differences in peak force can be attributed to the stress-strain approximation approach. Since only one loading rate was used, and an average value was taken over a large area, the average stress for all local strain rates was used. In the simulation and in the experiment, the local strain rate at a given location in the foam can be either higher or lower than the average strain rate. The spatial variability in strain rate and stress from the semi-spherical indenter simulation are shown (Fig. 44). An improved model would be strain rate dependent, and take stress-strain data from a more uniformly (spatially) loaded specimen.

a) Strain Rate

Time = 9.6
Contours of Mean Strain-Strainrate
min=-0.108658, at elem# 20011655
max=0.0654399, at elem# 20011779
section min = -0.0104148, near node# 20003368
section max = 0.0530262, near node# 20000481

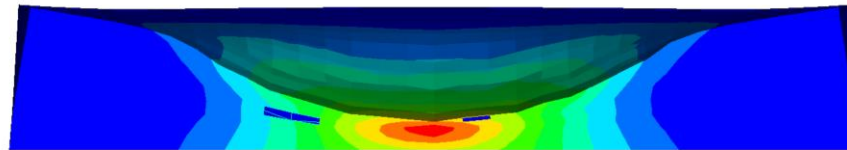


Mean Strain-Strainrate

6.544e-02
4.803e-02
3.062e-02
1.321e-02
-4.199e-03
-2.161e-02
-3.902e-02
-5.643e-02
-7.384e-02
-9.125e-02
-1.087e-01

b) Von-Mises Stress

Time = 9.6
Contours of Effective Stress (v-m)
min=1.866e-07, at elem# 20009903
max=0.00169149, at elem# 20011762
section min = 3.20335e-06, near node# 20003349
section max = 0.00164114, near node# 20011052



Effective Stress (v-m)

1.691e-03
1.522e-03
1.353e-03
1.184e-03
1.015e-03
8.458e-04
6.767e-04
5.076e-04
3.384e-04
1.693e-04
1.866e-07

Figure 44: Spatial variance in stress and strain rate, including: a) mean strain rate per element of simulated closed cell PVC foam 9.6 ms into impact with a spherical impactor, demonstrated on a mid-sagittal section. The strain rate ranged from -109 to 65.4 1/s (compressive strain rate is negative) within the foam, and b) the von Mises stress of simulated closed cell PVC foam 9.6 ms into impact with a spherical impactor, demonstrated on a mid-sagittal section. The von Mises stress ranged from 0.00 to 1.69 MPa.

In the unloading of the indenter, the differences between simulated and experimental force were more pronounced (Fig. 39). The experimental indenter was slower to rebound, and the experimental force was quicker to drop. The slower rebound and quicker drop in force can be attributed to hysteresis, which was not accounted for in the constitutive model used. Moreover, the experimental force demonstrated oscillations after the force reached zero. Since there were no corresponding oscillations in the displacement of the indenter, the oscillations were likely in the test fixture and were only present after the indenter lost contact with the foam.

Consequently, the head-to-foam contact force in the head-first impact simulations showed good agreement with the experiments, with both low-speed and high-speed impacts having bimodal force-time histories in experiments and simulations, which decreased until 100 ms after the second peak. The

peak force value of the low-speed simulation was within the experimental range, and the high speed was higher than the experimental average (1.5 kN higher). Given that there were only two experiments at each velocity, similar qualitative results point towards a good replication of the head-plate setup. Moreover, when scaling by body mass, the peak head impact force of the 3.05 m/s HFI simulation was 23% lower than the experimental average of the 3.10 and 3.08 m/s impacts, while the 3.59 m/s simulation was 7% higher than the average of the 3.54 and 3.63 m/s impacts.

The head force in Study 2 was mostly based on the head mass because the impact was to a rigid plate, so peak force occurred before significant loading was measured in the neck. Due to the foam slowing down the head loading in Study 3, the cervical spine was loaded up as the head was loaded up. Given that in Study 2 it was determined that the posture affected the neck loading, the posture then also may have affected the head force in Study 3. The difference in experimental and computational initial posture of the cervical spine can account for the difference in peak head force.

For experiments and simulations, T1 began rotating in extension until a peak value was reached (Fig. 41). In low speed (3.01 and 3.08 m/s) experiments, T1 rotated towards and past the original position by 100 ms, whereas in higher speed (3.63 and 3.54 m/s) experiments, T1 stayed at a similar rotation for the remainder of the impact. In simulations, there was much less extension rotation of T1 compared to the experiments, and there was more flexion rotation subsequently. The difference in T1 rotations between the experiments and the simulations can be attributed to the initial postures. Whereas the simulation attempted to match the initial head angle and torso angle of the experiment, additionally matching the C7-T1 angle would have required extensive morphing. Given that the initial postures of the cadaveric necks were lordotic, the neck lordosis and thorax kyphosis during impact allowed the cadaveric T1 to rotate in extension. Full body simulations with an initially lordotic spine, but a more upright torso (FB_{NEUT} posture), had double the T1 posterior rotation compared to the experiments (Appendix B), performed as a supplemental study. Therefore, the experimental C7-T1 angle, and the T1 rotation both lie between the two numerical cases, suggesting that future computational studies with matching head angle, C7-T1 angle, and torso angle will have T1 kinematics closer to the experiment.

Finally, while the head impact forces were lower in Study 3 compared to Study 2 due to the softer padding, there was still hard tissue element erosion in T1-T2 due to compressive loading and bending moment during the impact, as well as flexion failure in posterior vertebral elements in C2-C4. The element erosion began between the first and second head force peaks at both speeds, at a time when in

impacts to rigid plates, the head would be rebounding and moving laterally. Since the head was in contact with the foam the entire simulation, the head was pocketed in the foam, constraining it. As a result, in the lower loading conditions there was still significant element erosion. For both speeds, the combination of bending moment (~ 20 Nm) and compressive force (~ 2 kN) was at or close to the failure limit for C2-C6 motion segments observed in Study 1. Given that fracture-dislocation occurs most frequently in C5-C7 (Nightingale, Myers, & Yoganandan, Neck Injury Biomechanics, 2015), and element erosion occurred first elsewhere, the model requires further development to reproduce the most common injuries from epidemiology. Including active musculature and reposturing to replicate pre-impact bracing (Al-Salehi, Cripton, & Siegmund, Pre-Impact Cervical Spine Posture of Human Subjects in Headfirst Free Falls, 2023) are expected to change the location of element erosion.

Chapter 6

Conclusion and Recommendations

6.1 Conclusions

6.1.1 Study 1: Isolated Motion Segment Model to Investigate Response Under Combined Compression and Bending and Relationship to Fracture-Dislocation

In Study 1, cervical spine segments composed of two vertebrae were simulated using applied flexion moment, compressive force, and both, to determine when failure (element erosion) associated with fracture-dislocation would initiate. The flexion simulations predicted failures at the same bending moments reported in flexion experiments. However in compression loading, element erosion, corresponding to failure initiation, occurred at a lower displacement compared to the experiments. In this study element erosion was used to model failure, which is limited in predicting post fracture response. Therefore, the simulation results were only considered up to the initiation of fracture, not a large-scale decrease in compressive force as reported in the experiments. In combined loading, the segment rotated anteriorly when loading was predominantly flexion, or rotated posteriorly when the loading was predominantly compression. In both cases of combined loading a higher amount of both compressive force and flexion moment was required for failure to initiate, relative to single-mode loading. The failures obtained during pure flexion, as well as flexion-compression combined loading, when rotation was anterior, matched reported failures associated with fracture-dislocation, which included disruption of posterior ligaments and fracture at the anterior aspects of the vertebral body. The outcome of this investigation demonstrated that the motion segment predicted flexion injuries, but predicted an earlier onset of failure in compression loading compared to experiments.

A second displacement-controlled model was investigated to generate fracture-dislocations in the motion segment. Displacement control allowed for the separation of flexion and shear loading conditions. Flexion rotations of 20° and 30° produced posterior ligament failure, and often hard tissue element erosion in the anterior aspects of the vertebral body, the most commonly observed fracture-dislocation. Applying anterior shear following flexion led to unilateral or bilateral perched or dislocated facet joints. The amount of compression and flexion load corresponding to fracture-dislocation was quantified for comparison to the loading conditions in the full neck models. This is the first study to obtain a fracture-dislocation at the segment level using numerical models.

6.1.2 Study 2: Importance of Neck Boundary Condition and Posture on the Cervical Spine Response During a Head-First Impact

In this study, the GHBMC M50 human body model was used to investigate the differences in head and neck response under HFI using a head and neck with simplified boundary condition (TSM), and using the FB human body model. The maximum predicted head impact force was similar for both models, since it was primarily related to the head mass; however, the FB had a lower neck force compared to the TSM attributed to the torso compliance in the FB model. Owing to higher neck loads in the TSM, it was found that element erosion occurred at a velocity below the conventional injury threshold velocity (3.14 m/s) reported in experimental studies, suggesting the axial loading in TSM is higher than in real-world HFI scenarios. The effect of posture (flexed, neutral, and extended neck posture) on the head and neck response under HFI was also investigated. It was found that models with initial lordosis, increased the lordotic bending during impact, leading to first order buckling of the cervical spine. With a flexed (kyphotic) initial posture, the neck moved further into flexion during head-first impact, leading to a lower neck force. As a result of the lower neck force, flexed impacts had less deformation in the neck elements, and less element erosion.

6.1.3 Study 3: Assessing the Neck Response of the Full Human Body Model with Whole Body PMHS Experimental Data

In Study 3, full body HFI was simulated with boundary conditions of experimental full body tests (Kerrigan, et al., 2014) where four subjects were inverted and dropped onto a closed cell PVC foam, backed by a steel plate. To simulate the experiments, a material model of the foam was developed and showed good correspondence to a drop weight characterization experiment. The GHBMC was impacted at 3.05 and 3.59 m/s to be within ± 0.05 m/s of the lower speed (3.01 and 3.08 m/s) and higher speed (3.54 and 3.63 m/s) impacts.

Scaling by body mass, the head force traces were qualitatively similar to the experiments, and the peak values were also comparable. The scaled peak head impact force of the 3.05 m/s HFI simulation was 23% lower than the experimental average of the 3.10 and 3.08 m/s impacts, while the 3.59 m/s simulation was 7% higher than the average of the 3.54 and 3.63 m/s impacts. The variation in head forces is consistent with other head-first impact simulations, and additional tests are required to quantify the variability in the full body experimental tests.

To approximate the experimental subject postures, a rotated FB_{FLX} model (FB_{FLX+30°), and the FB_{NEUT} model from Study 2 were used. While FB_{FLX+30° and FB_{NEUT} either had a comparable torso angle or neck lordosis while both having similar head angles, achieving all of the experimental targets would require significant reposturing and morphing of the subject specific GHBMCM50 model. In the experiments, the initially lordotic cervical spine and kyphotic thoracic spine caused T1 to rotate in extension as the spine increased in cervical lordosis and thoracic kyphosis. In the simulations, the T1 kinematics demonstrated a high sensitivity to initial posture. The T1 of the FB_{FLX+30° model began rotating in extension as the spine was initially compressed on impact, but the overall spine flexion caused T1 to rotate in flexion. In FB_{NEUT} , the T1 extension rotation was in the same sense as the experiment, but had more rotation than the experiment. Therefore, two different postures of the GHBMCM50, which either had a similar torso angle or neck lordosis to experiments but not both, bracketed the experimental T1 rotation.

The model with element erosion included predicted more fracture than the experiments due to the different cervical spine postures. While the C7 loading in the FB model was similar to the fracture-dislocation load predicted from the isolated motion segment in Study 1, failure occurred in the FB at T1-T2 and C2-C4, releasing the impact energy and therefore not generating a fracture-dislocation in the C5-C7 section of the spine. In summary, this study produced similar loading to cadaveric HFI experiments, kinematics that bracketed the experimental T1 rotation, and vertebral forces which were found to cause fracture-dislocation in Study 1.

6.2 Limitations

Although Study 2 was the first study to investigate the effect of posture on the neck response under HFI in an FB model, the thoracic spine posture was not modified. It is possible that the initial posture of the thoracic spine could play an important role in the overall spine deformation in HFI. Moreover, only one configuration of legs and arms, which put the center of mass of the body anterior of the head and neck was investigated. While measured moments were lower in FB than TSM despite the eccentric center of mass, the effect of the extremity position was not investigated. This study did include passive muscles, but it did not include active musculature, which likely increases stiffness in the neck (Hu, Yang, Chou, & King, 2008), and possibly the thorax. Increased stiffness in the neck and thorax can likely influence the neck kinetic and kinematic response under HFI. Future work includes investigating the effect of muscle activation on the neck response under HFI. Additionally, the M50 model is a subject specific

model that does not capture population variability or may differ from population averages. For example, the neck length of the M50 model has been shown to be 10.8% longer than the population average for mid-stature young males (Corrales & Cronin, 2021).

Only an average stature young male was investigated; posture and boundary conditions may differently affect other population groups thought to be more vulnerable to injury (e.g. average females and the aged population). Future planned work includes the investigation of postural effects to alternate population groups under HFI. The skeletal tissues were modelled using a metal plasticity model, and bone failure was modelled using element erosion. Element erosion leads to energy being unphysically removed from the system (Corrales, Bolte, Malcolm, Pipkorn, & Cronin, 2022) and this is a limitation of the present study. For example, when trabecular bone is crushed in experiments, the comminuted material does not disappear but continues to do work on surrounding materials. On the other hand, with no element erosion, unphysical amounts of energy are kept in the system, which should be released as the bones crush. Therefore, with element erosion included and not included, the upper and lower bounds of energy loss were modelled, with the true results likely lying between these extremes. Future work should incorporate a model with an appropriate bone compressive fracture model, such as a smoothed particle hydrodynamics implementation (Ngan, Rampersadh, Rycman, & Cronin, 2024). Additionally, the metal plasticity models were limited in predicting fracture in bones due to their lack of orthotropy and asymmetry (Khor, Cronin, Watson, Gierczycka, & b, 2018), as well as a damage and fracture models (Cronin, Watson, Khor, Gierczycka, & Malcolm, 2022).

6.3 Recommendations

From Study 1, it is recommended that a full neck force-moment study be undertaken to quantify the effects of combined loading on fracture for the whole neck. To build confidence in the ability of the full neck model (or osteo-ligamentous spine) to produce a fracture-dislocation, the experiments of (Myers, et al., 1991) which reliably reproduced bilateral facet dislocation quasi-statically can be reproduced. For both of these studies, it is recommended that the SPH post-fracture model (Ngan, Rampersadh, Rycman, & Cronin, 2024) and possibly an improved bone fracture criterion (Cronin, Watson, Khor, Gierczycka, & Malcolm, 2022) be implemented.

In Study 2, it was concluded that the TSM and FB boundary conditions produced different kinematic, kinetic, and element erosion results for different postures. It is therefore recommended that the full-body be used when available to study head-first impact, or a TSM that more accurately captures the

boundary condition of a full body. Currently, no modified TSM attempting to replicate the FB loads and kinematics exists. Other factors such as muscle activation, head padding/pocketing, or off-axis loading should also be studied with a full body, with the intention of determining factors associated with fracture-dislocations. While effects from muscle activation, pocketing, and off-axis loading have been studied with TSM models, the full body may result in differences that have not been detected in TSM head-first impact.

Finally, in Study 3 the GHBM 50th percentile male was compared with the experiments of (Kerrigan, et al., 2014), but the pre-impact posture was not fully matched. A validation of the full body model should be performed in head-first impact. The FB model should be repostured to match the postural targets from the experiments (Kerrigan, et al., 2014), and spinal kinematics and kinetics can be compared. It is also recommended to continue the work of (Kerrigan, et al., 2014) and perform full body cadaveric HFI to more specimens to build corridors to better validate models and understand injuries associated with HFI.

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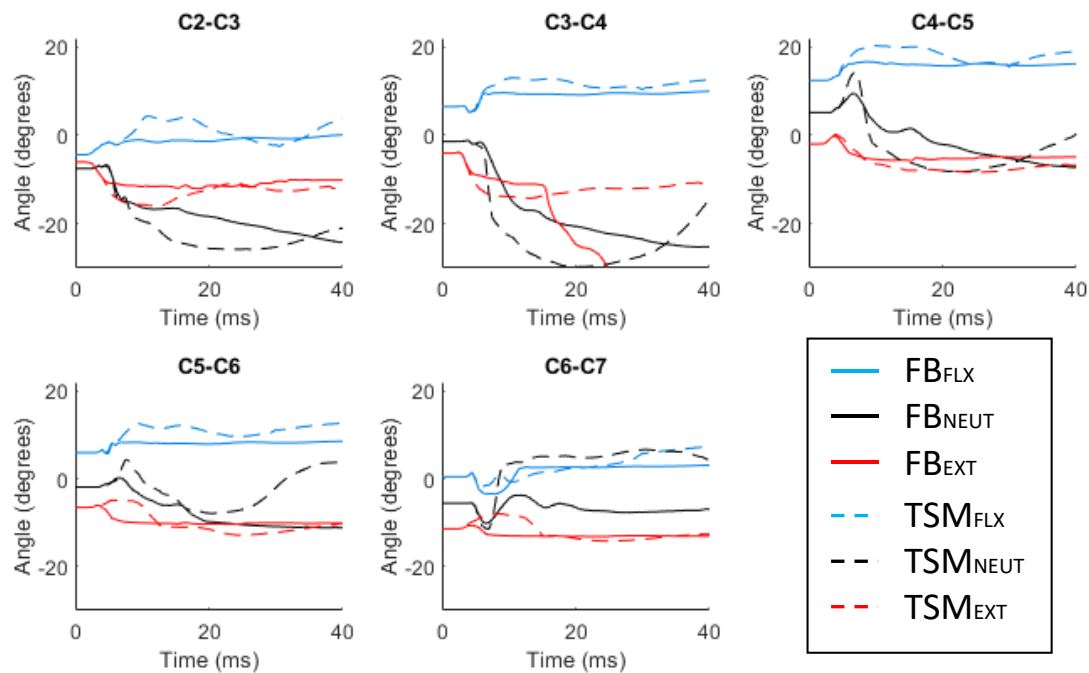
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Appendix A: Intervertebral Kinematics for Full Body and Torso Surrogate Mass Human Body Models Under Head-First Impact in Three Postures

Intervertebral Angle (degrees) vs Time (ms)



Appendix B: T1 Rotation During Head Impact to a Closed-Cell PVC Foam Compared to Experimental Results of (Kerrigan, et al., 2014)

