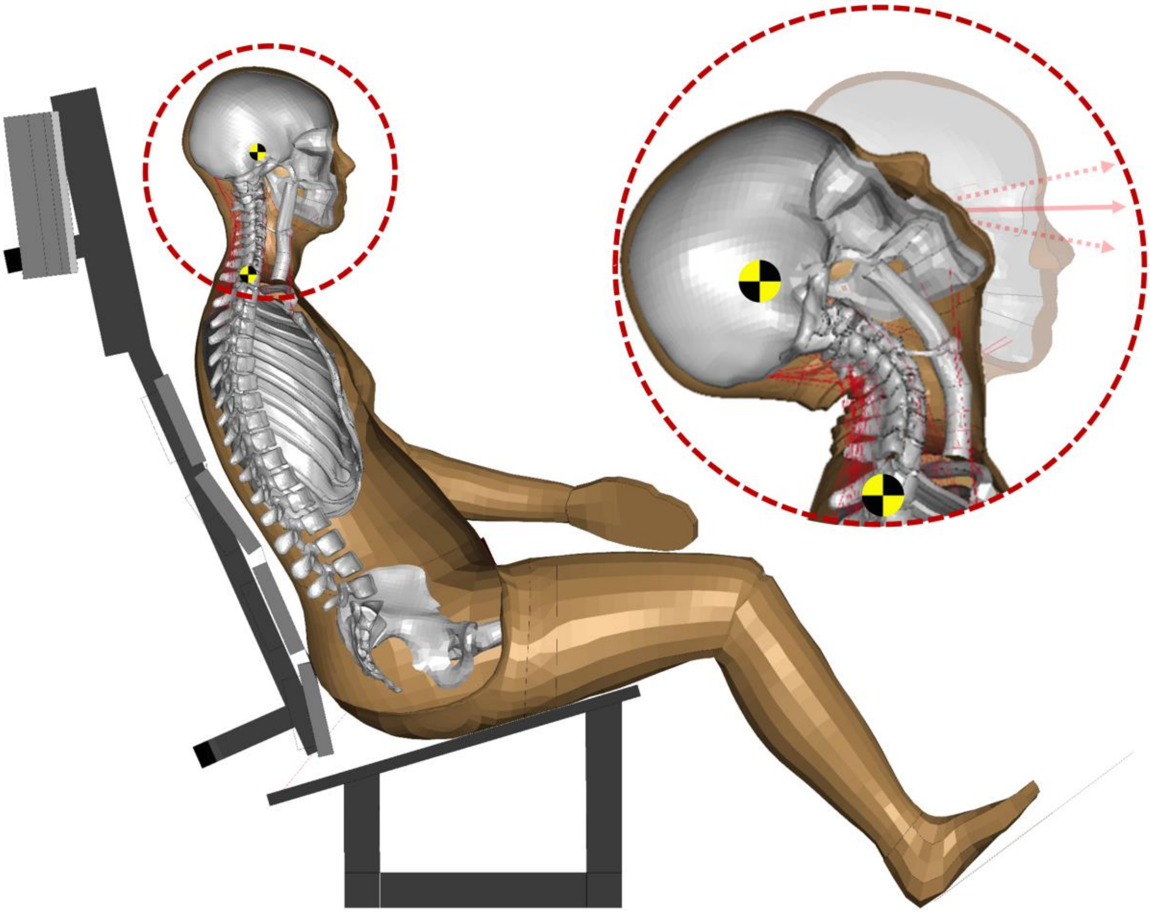

An Average Female Head-Neck Finite Element Model with Reflexive Neck Muscles

I PUTU ALIT PUTRA

DEPARTMENT OF MECHANICS AND MARITIME SCIENCES
CHALMERS UNIVERSITY OF TECHNOLOGY

Gothenburg, Sweden 2020
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THESIS FOR THE DEGREE OF LICENTIATE OF ENGINEERING

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Department of Mechanics and Maritime Sciences
Chalmers University of Technology
SE-412 96 Gothenburg
Sweden
Telephone + 46 (0)31-772 1000

Cover:
The ViVA 50th Percentile OpenHBM with Active Neck Muscle.

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"With every mistake, we must surely be learning…"
-George Harrison, 1968.

An Average Female Head-Neck Finite Element Model with Reflexive Neck Muscles

I PUTU ALIT PUTRA
Department of Mechanics and Maritime Sciences
Division of Vehicle Safety
Chalmers University of Technology

Abstract

Several factors potentially contribute to the risk of whiplash injuries; one of them is the neck muscle activities. Muscle activities in the neck have been shown to influence the head-neck kinematics during whiplash-like rear impacts. Thus, it is necessary to include the neck muscle responses when conducting a study of head-neck kinematics in a whiplash-like rear-impact condition. Therefore, as the first step, the development which focused on the implementation and optimization of a 50th percentile head-neck FE model with active reflexive neck muscles was conducted in the present thesis.

The active muscles were implemented in the existing ViVA OpenHBM, and the work was divided into three studies. The first study concluded that both neck link angular position feedback (APF) and muscle length feedback (MLF) control strategies improved the head kinematics agreement compared to the passive model, but overall, the APF controller performed better. The second study showed that the optimum controller gains and parameters could be identified using optimizations. The final study evaluated different ways to combine APF and MLF controllers. Further study and optimizations are needed to understand the best way to implement and combine MLF controllers with APF controllers.

The combined work increased the understanding of how to model active neck muscle controllers representing human reflexes during whiplash induced rear-impact. The optimization strategy used in the present thesis could be repeated in other head-neck models and in other regions of the human body in future work. In summary, an open-source head-neck FE model that represents a 50th percentile female in stature and mass with active reflexive neck muscle is now available and with future development will be used to study head-neck kinematics and its relation to whiplash injuries.

Keywords: *Finite Element; Human Body Model; Active Neck Muscle; Whiplash Injury*

Acknowledgments

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I also want to thank Professor Mats Svensson as my examiner. Mats, thank you for your feedback, guidance, and advice. Thank you for sharing your extensive knowledge of whiplash injury. I am looking forward to exploring the ideas that we have been discussing so far.

I am also would like to thank the ViVA II Project Partners for their supports: Adj. Professor Astrid Linder (VTI), Adj. Prof. Anders Kullgren (Folksam Research), Adj. Prof. Lotta Jakobsson (Volvo Cars) and Anders Flogård (Crash Scene Investigation). Also, people that had supported me during the beginning of the project: Prof. Karin Brolin, Dr. Fusako Sato, Adj. Assoc. Prof. Jonas Östh, Dr. Jona Olafsdottir and Erik Brynskog.

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Last but not least, I would like to thank my wife and life companion, Ayuwidia, and my son, Harsakanaka. Widia, thank you for your endless love and care, support, and understanding. I hope you still don't mind sharing me with the laptop for the next two years. Harsa, thank you for choosing me as your father. You keep reminding me about the importance of curiosity.

I Putu Alit Putra

Gothenburg, March 2020

List of Appended Papers

This thesis is based on the work reported and presented in the following publications:

Paper 1	**I Putu A. Putra**, J. Iraeus, R. Thomson, M.Y. Svensson, A. Linder, F. Sato. **“Comparison of Control Strategies for the Cervical Muscles of an Average Female Head-Neck Finite Element Model.”** *Traffic Injury Prevention. 2019, Vol. 20, No. S2, S116–S122* https://doi.org/10.1080/15389588.2019.1670818
	Contribution of authors*: **Putra:** Methodology, Software, Formal analysis, Investigation, Writing – original draft, Visualization. **Iraeus:** Conceptualization, Software, Supervision, Writing – reviewing and editing. **Thomson**: Conceptualization, Supervision, Writing – reviewing and editing. **Svensson**: Supervision, Funding acquisition, Writing – review and editing. **Linder:** Funding acquisition, Project administration, Writing – reviewing and editing. **Sato:** Resources, Writing – reviewing and editing.
Paper 2	**I Putu A. Putra**, J. Iraeus, F. Sato, M.Y. Svensson, A. Linder and R. Thomson. **“Optimization of Female Head-Neck Model with Active Reflexive Cervical Muscles in Low Severity Rear Impact Collisions.”** *Under second review. Annals of Biomedical Engineering.2020*
	Contribution of authors*: **Putra:** Conceptualization**,** Methodology, Software, Formal analysis, Investigation, Writing – original draft, Visualization. **Iraeus:** Conceptualization, Software, Supervision, Writing – reviewing and editing. **Sato:** Resources, Writing – reviewing and editing. **Svensson**: Supervision, Funding acquisition, Writing – review and editing. **Linder:** Funding acquisition, Project administration, Writing – reviewing and editing. **Thomson**: Conceptualization, Supervision, Validation, Writing – reviewing and editing.

	I Putu A. Putra, J. Fice, J. Iraeus, M.Y. Svensson and R. Thomson **“Simplified Approach to Combine the Control Strategies of Active Reflexive Neck Muscles for an Active Female Human Body Model”**
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	Contribution of authors*: **Putra:** Conceptualization**,** Methodology, Software, Formal analysis, Investigation, Writing – original draft, Visualization. **Fice**: Supervision, Validation, Writing – reviewing and editing. **Iraeus:** Supervision, Writing – reviewing and editing. **Svensson**: Supervision, Writing – reviewing and editing. **Thomson**: Supervision, Writing – reviewing and editing.

* Contributor Roles Taxonomy (CrediT) (Brand et al. 2015) was used to recognize the contribution of each author.

** Paper 1 was updated in this thesis, with the details attached in Appendix 1.

Conference Presentations of the Present Work

I Putu Alit Putra, Johan Iraeus, Robert Thomson, **Implementation and Calibration of the LS-DYNA PID Controller for Female Cervical Muscles**, Nordic LS-Dyna User Conference 2018, Gothenburg, Sweden. October 18-19, 2018.

I Putu Alit Putra, Johan Iraeus, Robert Thomson, Mats Y. Svensson. **Implementation and Calibration of Active Reflexive Cervical Muscles on Female Head-Neck Model**. The 15th Annual Injury Biomechanics Symposium the Ohio State University, Ohio – United States, May 19-21, 2019.

I Putu Alit Putra, Johan Iraeus, Robert Thomson, Mats Y. Svensson, Astrid Linder, Fusako, Sato, **Comparison of Control Strategies for the Cervical Muscles of an Average Female Head-Neck Finite Element Model**. The 63rd Annual Scientific Conference for the Association for the Advancement of Automotive Medicine. Madrid, Spain. October 15-18, 2019.

List of Abbreviations

APF	Angular-positioned feedback
C.G	Center of gravity
CCR	Cervicocollic reflex
EMG	Electromyography
FE	Finite element
HBM	Human body model
MLF	Muscle-length feedback
PHMS	Post mortem human subject
PID	Proportional integral derivative
VCR	Vestibulocollic reflex
ViVA OpenHBM	Virtual Vehicle Safety Assessment: Open-source Human Body Model
WADs	Whiplash associated disorders

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Part I Overview Chapters

Chapter 1 Introduction

1.1 Whiplash Associated Disorders (WADs)

Whiplash injuries are very common around the world. In Sweden, from 1985 to 1986, 139 patients were identified with soft tissues related injuries (the incidence was 1 per 1000 inhabitants, Björnstig et al. 1990). Meanwhile, a more recent study also in Sweden revealed that there were 3297 cases of acute whiplash injury during 2000 – 2009 (Styrke et al. 2012). A high number of whiplash injuries victims were also reported in France and Spain with 12% of 7558 drivers and 12.2% of 8720 drivers involved in crashes sustained whiplash injuries (Martin et al. 2008).In Greece, a review of three years of patient data reported all traffic accidents related patients (180 subjects) reported neck pain, and 13.9% experienced dizziness (Partheni et al. 2000). In Japan, an online survey with 4164 participants involved in traffic accidents revealed that 183 people had neck pain for more than six months, while another 333 subjects had minor neck pain requiring treatment within three months (Oka et al. 2017). A ten-year based study in Australia concluded that one-third of the 150.794 traffic accident compensation claims were for whiplash injury (34%) (Gisolf et al. 2013). Similarly, per 100.000 population in the US, 328 were treated with neck sprain (Quinlan et al. 2004). In Saskatchewan, Canada, the claim for whiplash injuries was 417 per 100.000 populations (Cassidy et al. 2000). Despite the high incidences of whiplash injuries worldwide, whiplash injury etiology remains unclear and not fully understood.

Several common symptoms have been observed in the patients that sustain whiplash injuries. Based on literature reviews (McClune et al. 2002, Sterner and Gerdle, 2004, Yadla et al. 2008, Sterling, M. 2011) the symptoms experienced by the whiplash patients were neck pain and stiffness, pain in the arm, headache, numbness or paresthesia, dizziness, visual and auditory problems, problems with memory and concentration as well as psychological issues such as anxiety. Females have been shown to have a higher risk (up to three times) to experience whiplash injuries (Carlsson et al., 2011). Another study by Kullgren et al. (2013) also concluded that the whiplash protection seats, found on the Swedish market, were less effective for females than for males.

Most whiplash injuries have been found to be associated with motor vehicle accidents. From different type crash directions, most whiplash injuries were associated with rear-impact collision (Krafft et al. 2002, Stigson et al. 2015).

Based on the above descriptions, it can be summarized that the high incidences of whiplash injuries were found almost in all continents; however, how the injury is manifested in the body is still not clear. Whiplash injuries were found to be mostly associated with rear-impact collisions with females having a higher susceptibility to sustain whiplash injuries compared to males. Therefore, further studies are needed to study whiplash injuries, especially to shed some light on how the injury occurs in females during a rear-impact collision.

1.2 Head-Neck Kinematics during WADs-induced Rear-Impact

To understand how whiplash injuries develop during rear-impact, an understanding of occupant head-neck kinematics is essential. During a rear-impact collision, the neck kinematics can be divided into several phases (Svensson et al. 1993, Linder, A. 2001, Linder et al. 2002). The first phase is called the retraction phase. The retraction phase is caused by the relative motion of the head and torso, and it produces an S-liked shape in the cervical spine. The relative motion between the head and trunk occurs because when the car is hit from behind, the occupant's torso is moved forward, but the occupant's head remains in the same position because of inertia. After that, the second phase called the extension phase occurs and the head usually contacts the head restraint. After this contact, the third phase called flexion phase (or rebound phase) occurs as the head and torso rebound from the vehicle seat head restraint and seatback cushion.

Presently, many hypotheses of how the whiplash injuries occur are connected to the retraction phase of the neck (Svensson et al. 1993, Grauer et al. 1997, Ono et al. 1997, Yoganandan et al. 2002, Ono et al. 2006). Therefore, any surrogate tools (for example, crash test dummy or human body finite element model) that are typically used to study the kinematics of whiplash injuries should be able to replicate this occupant S-like retraction motion.

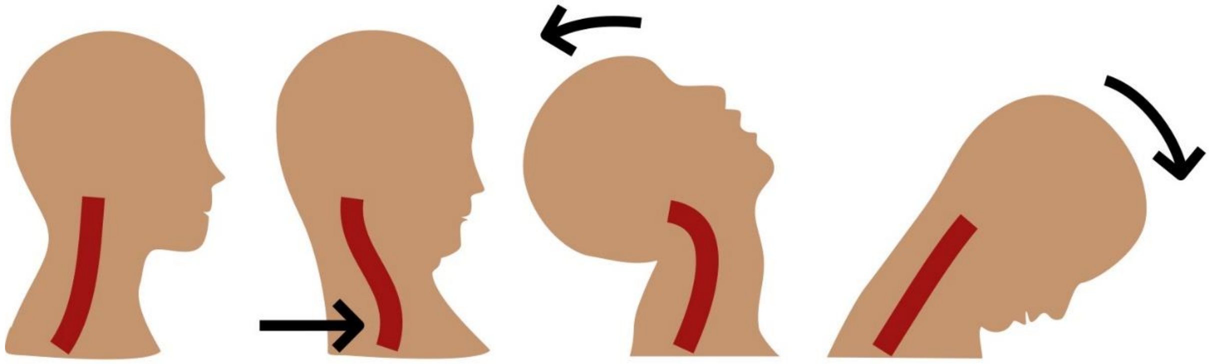


Figure 1.1 Occupant Motion during Whiplash Injuries Induced Rear Impact. The second figure illustrates the retraction phase with the neck S-shape.

1.3 Role of Cervical Muscle Activity in Head-Neck Kinematics

Besides vehicle and crash-related factors, such as head restraint design and impact severity (summarized by Carlsson, A. 2012), the head-neck kinematics of the occupant during whiplash induced rear-impacts are also found to be influenced by the cervical muscle activity (Brault et al. 2000, Siegmund et al. 2003, Blouin et al. 2006, Dehner et al. 2013, Mang et al. 2015). In terms of volume, cervical muscles are a significant part of the neck structure. It is also postulated that neck muscles could influence the whiplash injury risk by affecting other anatomical structures of the neck, for example the dorsal root ganglion and facet capsular ligaments (Siegmund et al. 2009).

Several studies have documented the influence of the neck muscle activity on the head-neck kinematics during whiplash-like rear-impact volunteer tests (Brault et al. 2000, Siegmund et al. 2003, Blouin et al. 2006, Dehner et al. 2013, Mang et al. 2015). Siegmund et al. (2003) conducted a volunteer study to analyze the effects of awareness on the subject muscles and kinematics responses during the whiplash-like perturbations and found that the Sternocleidomastoid (SCM) muscle activated earlier in aware volunteers. Unaware male subjects had higher head accelerations. Meanwhile, the surprised female subjects had larger head retractions compared to alerted subjects. Blouin et al. (2006) conducted a test with 65 volunteers that were exposed to a perturbation with and without a startle stimulus (the startle stimulus was induced by a124dB sound). They found that the startled subjects had earlier muscle activation, more substantial peak head acceleration with smaller peak displacements. A rear-end impact test with eight female subjects conducted by Dehner et al. (2013) found that the higher activity of sternocleidomastoid muscle contributed to the deceleration of head toward extension and acceleration toward flexion. Another study (Mang et al. 2015) found that subjects that experienced a pre-impact loud sound which inhibited the impact related startle response had a reduction of cervical muscle activity (C6 multifidus and C4 PARAspinal muscles) and a reduction in the peak head kinematic responses (6% reduction in extension angle, 9% in retraction and linear forward acceleration, 13% in extension) were observed.

In summary, muscle activities in the neck have an influence on the head-neck kinematics during whiplash-like rear impact perturbations. Therefore, it is necessary to include the cervical muscle responses when conducting a study of head-neck kinematics in a whiplash-like rear-impact condition.

1.4 Head-Neck Postural Control Reflexes

A detailed description of the human head-neck reflexes is beyond the scope of this thesis, but two main components, the Vestibulocollic reflex (VCR) and the Cervicocolic reflex (CCR) will be summarized here (Armstrong et al. 2008., Cullen and Goldberg, 2014., Cullen, K.E., 2012. Keshner, E.A., 2003).

The VCR activates neck muscles to maintain the head position and responds to head motion (rotational and translational) detected by the vestibular system. The human vestibular system can be categorized into two main components, called the peripheral and central parts. The detection of the head position is conducted by the peripheral components, which consists of semicircular canals and the otolithic organs (called saccule and utricle). The semicircular canals detect the head angular or rotational acceleration. Meanwhile, the otolithic organs detect the head in space by responding to gravitational acceleration, linear acceleration, and tilting of the head. The gathered information from the vestibular components is used as input to the vestibular nuclei in the brain (medulla). The CNS will process this input (along with other information from the eyes and cervical spine receptors) and then output different reflex mechanisms (one of the reflexes is the VCR). The reflex mechanisms then control the activation of the cervical spine muscles.

Meanwhile, the CCR activates neck muscles to reduce the motion of the head relative to the trunk and responds to changes in muscle length detected by muscle spindles. Muscle spindles, which can be found within the muscle belly, are observed in higher concentrations in the deep cervical muscles (Amonoo-Kuofi, 1983, Liu et al., 2003). They have a function to detect the muscle's length changes. The information on changing muscle length is transmitted to the CNS via the afferent nerve fibers. The CNS then processes the information and triggers the CCR. The CCR will activate the cervical muscles to maintain the head-on-trunk orientation.

Head and neck posture can be maintained though the combination of several input receptors such as the vestibular system, cervical muscle spindles, and joint articular receptors (Armstrong et al. 2008). All information is processed in the Central Nervous System (CNS) to activate different reflex mechanisms and triggering the corresponding muscles.

1.5 Finite Element Human Body Model with Cervical Muscles Reflexes

To date, several models have been developed with active cervical muscle controllers to simulate muscular reflexes. This thesis focuses on finite element (FE) models, so muscle control efforts on multibody dynamics (MB) models have been excluded from the present review. Three models called SAFER A-HBM (Östh et al. 2012, 2014a, 2014b, Ölafsdottir et al. 2019), THUMS version 5 (Iwamoto and Nakahira, 2015) and ViVA OpenHBM (Kleinbach, C.G. 2019) have been found to model the human's reflexes mechanism for controlling the cervical spine muscle activation.

Östh et al. (2012, 2014a, 2014b) developed a model to simulate passengers during braking. The model included active neck, lumbar, upper, and lower extremity muscles. To control the activation of the muscles, the authors implemented PID controllers. For the head-neck complex, the controller error signals were based on the head link angle and neck link angle. They grouped the neck muscles into flexion or extension groups based on the anatomical description of muscle function. From those studies, they concluded that the model with active muscle could capture the kinematic response of the volunteer's data used for the validation data.

More recently, Olafsdottir et al. (2019) developed a head-neck model with active muscles representing both VCR and CCR mechanisms. They developed an omnidirectional controller based on the neck link angle using PID controllers. The muscle activity was determined by spatial tuning patterns, physiologically based muscle grouping, that varied continuously based on impact direction. The model was evaluated using 1G loading in multiple directions. They found that the model with active cervical muscles reduces the head and neck motions compared to the model without muscle activation.

Another active FE HBM with active muscle reflexes called THUMS version 5 was developed by Iwamoto and Nakahira (2015). They implemented PID based active muscle controllers in all body regions, including the neck, and used a sigmoid function to model the firing rate. Their model was intended to study the head injury mechanism in lateral acceleration. They observed that the models with active muscle could capture the volunteer head kinematics better than the model without muscle activation.

Recently, Kleinbach, C.G (2019) enhanced an open-source FE HBM representing the 50th percentile female called ViVA OpenHBM. The author implemented an angle-based and length-based active muscle controller and validated the model using a low-speed rear impact of volunteer tests. The author concluded that the active muscle controllers increased the model agreement when the head CG, T1, and pelvis kinematics were compared. The author also emphasized that the head-neck rotation agreement needs to be improved further. Based on this study, it can be summarized that the author obtained promising results, although there was no evaluation for the intervertebral rotation in the cervical spine.

Only one model summarized in this section focused on predicted kinematics during a low-speed rear impact. Also, only one model was found to be open-source and represent females, which is a vulnerable occupant compared to males. To address these deficits, the current thesis will focus on the development of an open-source active FE HBM with active reflexive cervical muscles representing the human reflex mechanism to study whiplash injury kinematics in a low-speed rear impact.

Chapter 2 Objectives

The ultimate goal of the present research is to develop a validated Finite Element (FE) Human Body Model (HBM) that represents the average female population to predict the whiplash injury outcome during a rear-impact collision. By developing the female FE HBM, dynamic response and injury outcomes for females and males can be studied. Consequently, it could help understanding why females have a higher risk of sustaining whiplash injuries.

The development of the 50th percentile female finite element (FE) head-neck model with reflexive neck muscles in low-speed rear impact collisions was the goal in the present thesis. The specific objectives of this work were addressed by formulating the four research questions below:

- **Research Question (RQ) 1**: “Which active muscle controllers, angle-based or muscle-displacement based controller, can detect the head-neck motion and simulate the volunteer head displacement time histories?”
- **Research Question (RQ) 2**: “How well can different optimization schemes, targeting head and or vertebral volunteer kinematics outputs, identify the active muscle controller parameters and reproduce volunteer kinematics?”
- **Research Question (RQ) 3**: “Can spinal curvature influence the head kinematics?”
- **Research Question (RQ) 4**: “How can we combine two muscle controller strategies in a single head-neck model?”

These research questions are specifically addressed by several studies presented in the following chapters of this thesis. Three studies were designed to answer the questions within the scope of a two research period, relying on available experimental data.

Chapter 3 Material and Methods

The present study was arranged in a way that the research questions in Chapter 2 could be addressed in a systematic way (Figure 3.1). The development of active reflexive neck muscle was conducted in the ViVA OpenHBM that represented the 50th percentile female (Figure 3.2) and was divided into three studies. The first study aimed to answer the first research question by comparing and understanding how the different muscle controllers (angle-based and muscle-displacement based controllers) influence the head kinematics of the model in a low-speed rear impact. To answer the second and third research questions, the effects of different optimization objectives were evaluated with respect to the head-neck kinematics of the model with the angle-based controller. Specifically, different cervical spine curvatures were implemented, and vertebral level kinematics were analyzed. An attempt to combine those two controllers to answer the last research question was conducted in the third study. The combination was done based on the experience gained from the first and the second study. In the present thesis, the focus of the analysis was limited to the effects of neck muscle activation on the head-neck kinematics of the model.

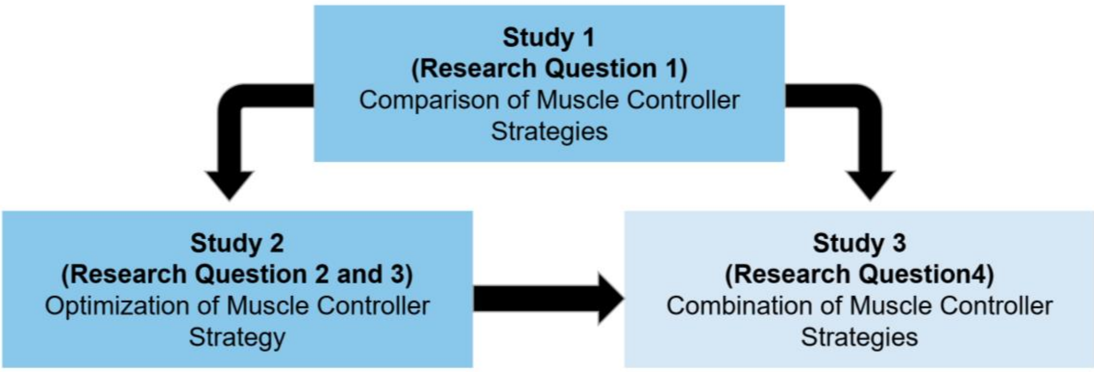


Figure. 3.1 Structure and Relation between Study in the Present Thesis

3.1 Overview of ViVA Open Human Body Model I (ViVA OpenHBM I)

The original ViVA OpenHBM, on which this work is based, was originally created by Östh et al. (Östh et al. 2016, 2017a, 2017b). The model was developed to represent the stature and mass of a 50th percentile average female (Figure 3.2). The model development was focused on the neck because the main intention was to use the model to study whiplash injuries in a low-speed rear-impact collision. Thus, the model has a detailed neck structure, while other body parts are simplified. The model has previously been validated against quasi-static loading component tests and whole-body rear-impact tests (Östh et al. 2017a, 2017b). In addition to the detailed cervical spine, a simplified cervical spine model was also developed (Östh et al. 2017b). The simplified model was created by removing the intervertebral non-muscular soft tissues of the neck. To compensate for the soft tissue removal, compliant joints (translational, axial rotational, lateral bending and flexion-extension) were added with joint stiffnesses based on published human in-vitro tests (Panjabi, MM. et al. 1986,2001 and Nightingale RW, et al.

2002). Only small differences in kinematics, evaluated using an objective rating evaluation (CORA) method (Gehre et al. 2009), were found when the detailed and simplified cervical spine models where compared in dynamic rear-impact test conditions (Östh et al. 2017).

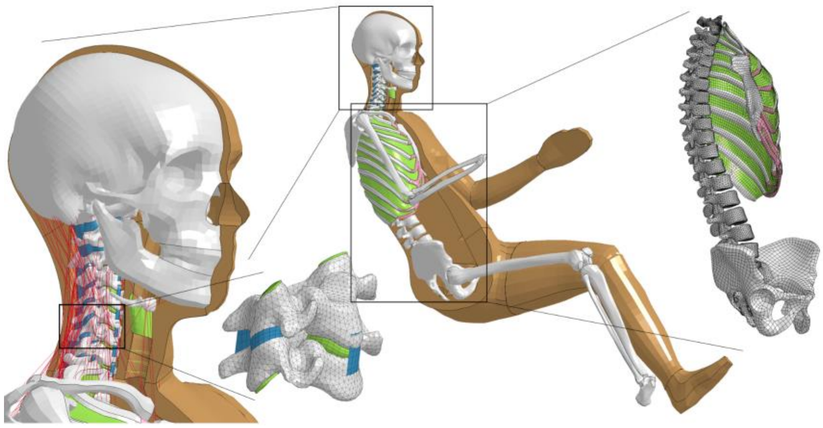


Figure. 3.2 ViVA I OpenHBM (Östh et al. 2017b). Source: Östh et al. ViVAOpenHBM I Project

3.2 Neck Muscle Modeling

The passive neck muscle modeling of ViVA OpenHBM I was described in Östh et al. (2017a). The neck muscles of the ViVA OpenHBM I model were implemented based on the Hill muscle model with PCSA from Borst et al. (2011). The origin and insertions of the muscles were based on anatomical descriptions from Standring (2008). There are 129 beam elements (resultant truss) of muscle elements to represent 34 muscles (Östh et al. 2017a). The LS-Dyna *MAT_156/*MAT_MUSCLE was used as the material model for the muscles.

The modelling of the active cervical muscles used in the present thesis was based on earlier research conducted by Östh et al. (2012,2015) and Olafsdottir et al. (2017,2019). The P.D. Controller, as defined by the PIDCTL function in LS-Dyna, was used to give activation signals to the neck muscles of the model.

In the present thesis, two forms of active neck muscle controllers are utilized. The first muscle controller is called as the Angular-positioned Feedback (APF) Controller. The APF controller activates neck muscle to maintain the head orientation relative to the global reference system. In this way, it has a similar in function to the VCR in humans. The controller vector (Figure 3.3) was defined based on the coordinate of the model head center of gravity (C.G.) and the center of the first thoracic (T1) spine vertebral body. This controller vector was sampled at time zero (t0). When the model head and neck are moving (due to impact loading), a new position of the head C.G. and the T1 were sampled and used to update the controller vector. An error angle, e(t), will be produced if there are any differences between the controller vector at t0 (reference angle) and the current angle, y(t). The projection of the controller vector is also used as the input to the spatial tuning pattern which define directionally specific muscle activations for each muscle group.

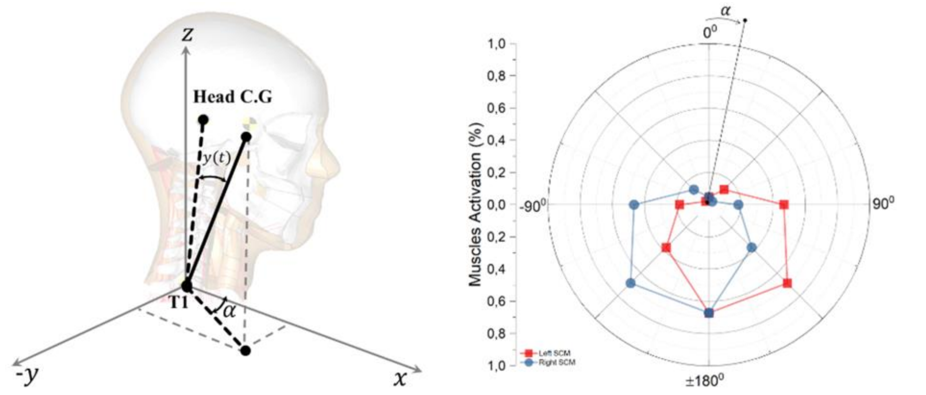
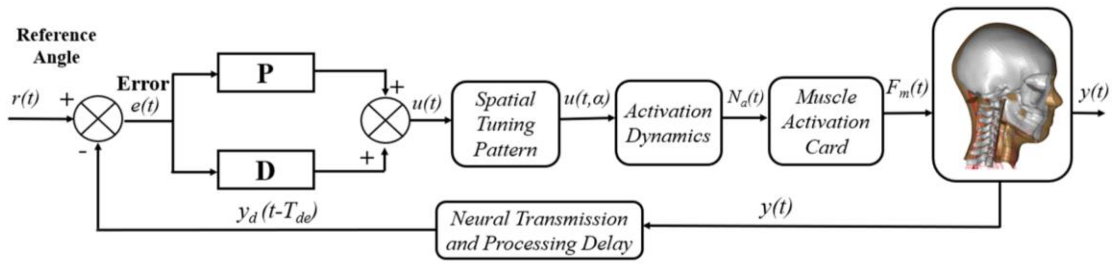


Figure 3.3 Controller Vector Calculation and Projection using Spatial Tuning Pattern

To mimic the neural processing delay, a delay was introduced in the error angle feedback signal, as presented in Figure 3.4. The delayed signal was then given to the PD controller. This signal was then compared with the reference angle to compute the control signal. To define the muscle activation, the computed-control signal was then given to the spatial tuning pattern.



$r(t)$	: Reference Angle calculated based on initial position of controller vector
$e(t)$	: Error Angle defined by Reference Angle $r(t)$ subtracted with Current Angle $y(t)$
$y(t)$	: Current Angle of the model calculated based on controller vector
$u(t)$	: Excitation signal as output from PD Controller
$u(t,\alpha)$	: Spatial Tuning Pattern-scaled PD Controller excitation signal
$N_a(t)$	: Muscle activation level
$F_m(t)$	: Muscle force generated by LS-Dyna Muscle Activation Card
$y_d(t-T_{de})$	: Current Angle $y(t)$ delayed using Neural Transmission and Processing Delay (T_{de})

Figure. 3.4 Controller Algorithm to Approximate the Feedback from Human Vestibular System

The second type of controller used in this thesis is called as Muscle-length Feedback (MLF) controller. It was developed to have similar function as the CCR which is to maintain head posture relative to body. The controller mechanism was similar to the APF controller, however, instead of changes in angle, changes in muscle length were used to derive the controller signal. As each muscle element has its own controller, there is no need for a muscle spatial tuning pattern in the MLF controller.

At time zero, the length of each muscle was calculated and used as the reference length. The length of each muscle was first calculated at a pre-defined time (set at time zero). Then, was after a delay mimicking the neural processing delay, the muscle length was measured again. When the length of the muscle changed due to a difference in length between the current and the reference length, the PD controller will produce an activation signal. The signal will be given to the corresponding muscles after being filtered and scaled using the activation dynamics filter. In the current study, each neck muscle element in the model has its own muscle length controller.

In the present thesis, the APF controller was used in all papers. Meanwhile, the MLF controller was used only in study 1 and 3. A preliminary attempt to combine these two controllers was described in Study 3.

3.3 Volunteer Kinematics Data and Spinal Alignment

The kinematics data of volunteer test based on Sato et al. (2014) was adopted to optimize the model head-neck kinematics responses in this thesis. Since detailed muscle response data was not available, the analysis was limited to the head and or neck kinematics. The volunteers were seated in a rigid seat without any head restraint. The seatback angle was 20-degree from vertical. The seat was mounted to a sled system that accelerated by the release of a compressed spring. The average delta velocity was 5.8 km/h, with peak acceleration equal to $42 m/s^2$.

The spine alignment of the original ViVA OpenHBM was used in all studies. However, the average spinal alignment based on X-ray analysis of volunteers from Sato et al. (2016) was also adopted as a comparison in Study 2. The vertebral alignment of the model was matched to an average of eight specific volunteers who had anthropometry that matched the average female population (Sato et al. 2016). The seat arrangements were similar to the sled used in the collection of perturbation data used as calibration in this thesis (Sato et al. 2014).

3.4 Optimization-based Parameter Identification

To answer the first and second research questions, optimizations to determine muscle controller gains, were conducted using LS-Opt (Stander et al. 2015). In the first study, the optimizations were designed to determine the values of the specific Proportional Gains (Kp), Derivative Gains (Kd), and Neural Transmission and Processing Delay (Tnd) for both APF and MLF controller. For the second study, the optimizations were extended to identify the time constants defining the muscle activation dynamics (Tna,a, Tna,d, Tne) in addition to Kp, Kd, and the Tnd. Metamodel-based optimization using Sequential Response Surface Method (SRSM) with Domain Reduction incorporated in LS-Opt (Stander et al. 2015) was used as the optimization method in Study 1 and 2. The optimization algorithm was an algorithm called the Hybrid SA (Simulated Annealing + Leapfrog Optimizer for Constrained Minimization). A Linear Polynomial Metamodel with D-optimal point selection was used to define the metamodel of the optimizations. The total number of simulation points per iteration was equal to 7 for Study 1, and 11 for the Study 2. The preset maximum number of iterations was ten and kept the same for Study 1 and 2. The reader is referred to Stander et al. (2015) for details.

3.5 Approaches to Combine Active Muscle Strategies

A preliminary study was conducted to combine two active muscle controllers using two approaches. This study was intended to initially determine the performance of APF and MLF controller in a single head-neck model. The first approach assumed that both APF and MLF controllers were controlling the activation of all neck muscles together (based on a review from Armstrong et al. 2008). In the second approach, the APF controller was used to regulate the activation of the superficial muscles. Meanwhile, the deep neck muscles were controlled by the MLF controller (See Study 3). The second approach was based on the fact that a high density of muscle spindles has been found in the deep cervical muscles (Amonoo-Kuofi, HS. 1983). The grouping of deep and superficial neck muscles was based on Borst et al. (2011). This study was designed as proof of concept. Therefore, the optimum gains were based on earlier studies and no optimizations simulations were conducted to address the combination of two controllers.

Table 3.1 Name of Simulation and Gains Combination

No	Name of Simulation	APF Controller Parameter	MLF Gains Controller Parameter
1	Passive Model	n/a	n/a
2	Combined-Control 1	Study 2 (Putra et al. 2020)	Olafsdottir et al. (2019)
3	Combined-Control 2	Study 2 (Putra et al. 2020)	Study 1 (Putra et al. 2019)
4	Distributed-Control 1	Study 2 (Putra et al. 2020)	Olafsdottir et al. (2019)
5	Distributed-Control 2	Study 2 (Putra et al. 2020)	Study 1 (Putra et al. 2019)

3.6 Simulation Boundary Conditions

In the current study, the optimization-based parameter identification was conducted using the ViVA OpenHBM Head-Neck model with simplified cervical spine. To mimic volunteer kinematics, the volunteers sagittal plane T1 kinematics were prescribed in the T1 of the model (Figure 3.5). The termination time of each simulation was 400ms included 100ms of initial quasi-static equilibrium settling with a gravitational acceleration equal to 9.81m/s^2.

3.7 Quantitative Ratings Evaluation

Throughout this thesis, the quantitative rating evaluation was conducted using CORA analysis (Gehre et a. 2009). The default corridor (5% inner limit and 50% outer limits) of CORAplus was chosen. To evaluate the head kinematics, the whole duration of the post equilibrium settling simulation (100-400ms, hereafter referred to as or 0-300ms) was compared. Due to the limitation of the reported spine kinematic data, the evaluation was only made during 0-180ms for the cervical spine kinematics. In addition of CORAplus 4.04, additional evaluations were conducted in Paper 1 using RSVVP Software (Mongiardini et al. 2013). Here, the Sprague-Geers metric in RSVPP was selected without any pre-processing steps.

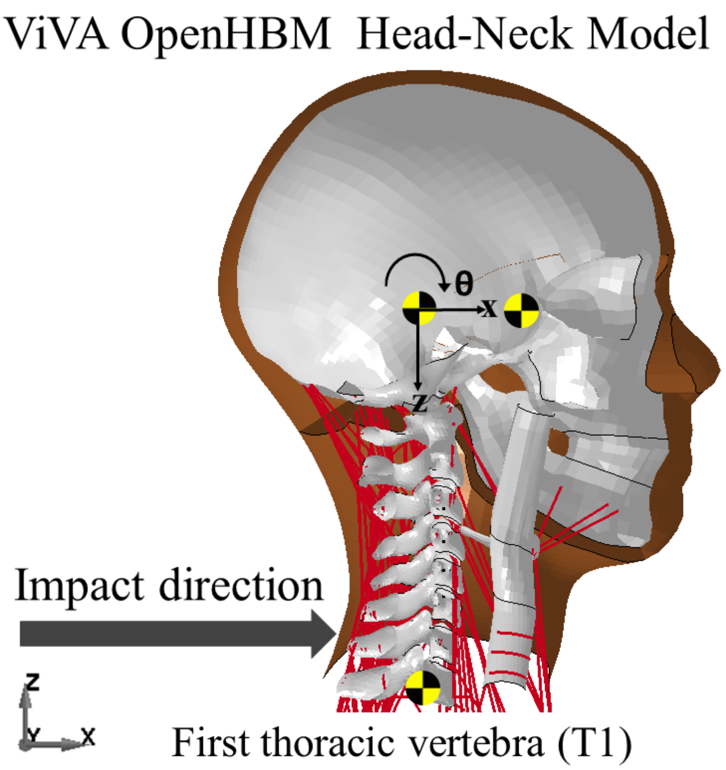


Figure 3.5 Simulation Setup

3.8 Software and Computational Environment

The pre- and post-processing software of the simulation models were conducted using LS-Prepost version 4.5 -x64 (LSTC 2012). LS-Dyna binary version 9.2 MPP double precision (LSTC 2016) was used for all simulations. Specifically, ls-dyna_mpp_d_r9_2_119543_x64_-redhat54_ifort131_sse2_intelmpi-413 was used for the passive models and model with APF muscle controller and ls-dyna_mpp_d_BETA_R9_121624_centos65_intel131_intelmpi binary was used to run the MLF model. The data analysis and graphics generations were conducted using OriginPro 2018b 64-bit (Study 1) and OriginPro 2019b 64-bit (OriginLab Corporation) (Study 2-3).

Chapter 4 Results

The summarized results based on Research Question 1 to 4 are presented below. The four research questions were addressed in three papers. Refer to appended papers for full details.

4.1 Kinematics Comparison between ViVA OpenHBM Head-Neck Model with APF and with MLF Controller (Research Question 1 - Study 1)

Both the APF and MLF controller changed the head kinematics (Figure 4.1). The model with the APF controller improved the agreement of horizontal head displacement (x-displacement) compared to the passive models (Figure 4.2a). In the vertical direction, active models could reduce the peak head displacements. However, no model (active or passive) could capture the first 100ms of the volunteer motion in the vertical direction (Figure 4.2b). When the head rotational displacements were compared, only slight differences in head motion was observed in the APF and passive models until impact time 100ms. The model with MLF controller had higher head rotational y-displacement compared to other models (Figure 4.2c). After that, the model with the APF controller could limit the maximum head rotational (Figure 4.2c). The model with the MLF controller could slightly improve the head kinematics from the passive model but had earlier non-realistic responses, not visible for the APF controller, that were not observed in the volunteer's head kinematics. These motions are best highlighted in Figure 4.2c.

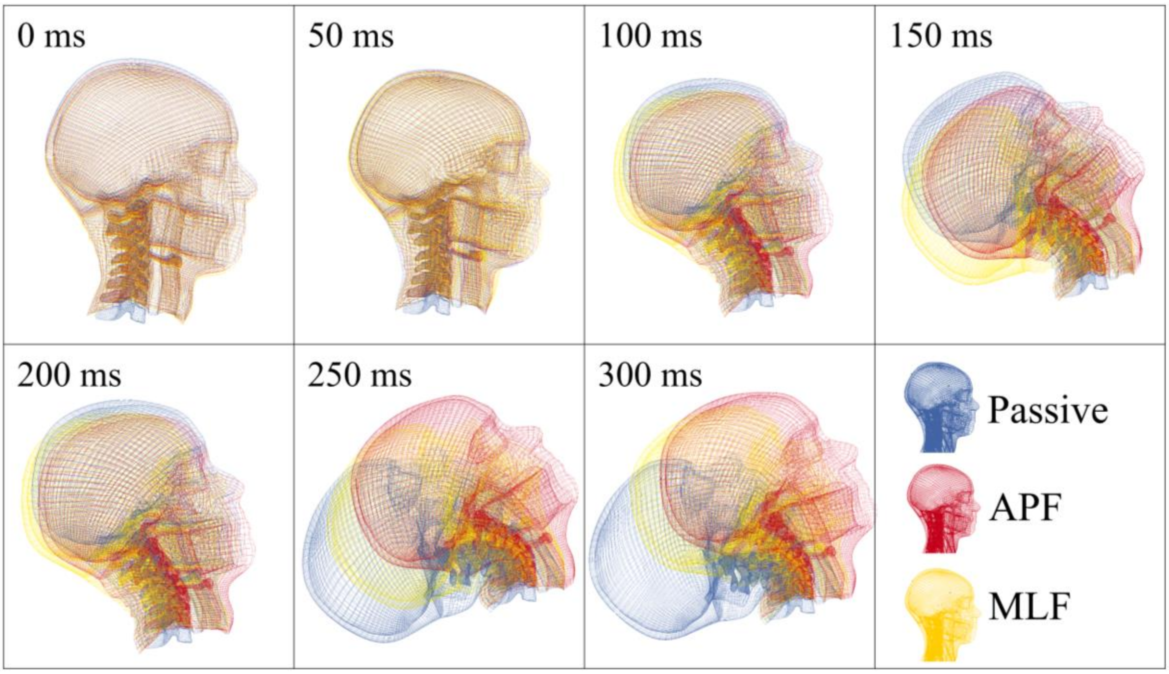
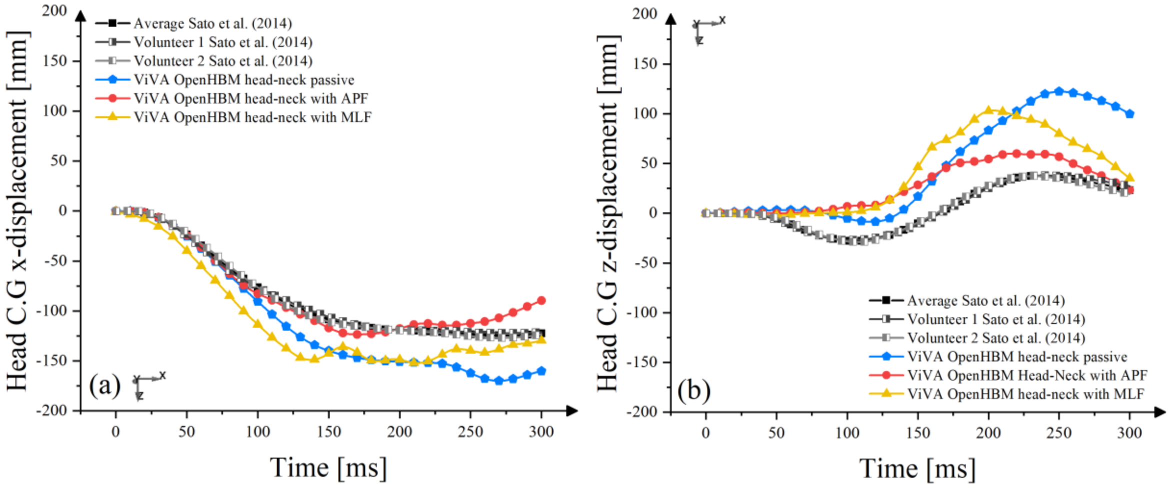


Figure 4.1 Time-series Comparison of Head-neck Kinematics in Study 1 (Paper 1)



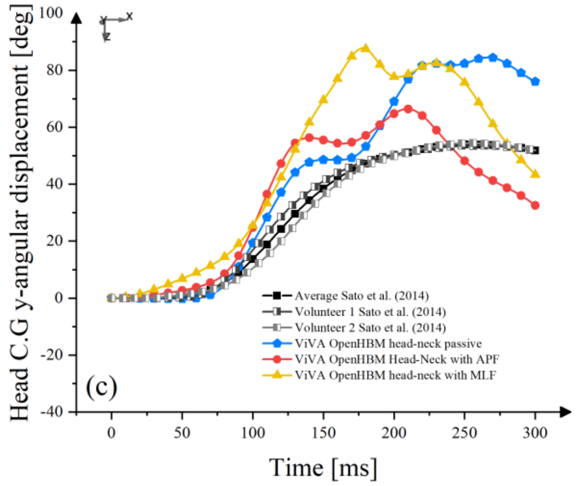


Figure 4.2 Head kinematics from simulation models and volunteer tests. (a) Head C.G x-linear displacement; (b) Head C.G z-linear displacement; (c) Head C.G y-angular displacement

4.2 Optimization-based Parameter Identification Results and Kinematics Comparison (Research Question 2 – Study 2)

Four different sets of controller gains, resulting in four schemes, were optimized in Study 2. The schemes were: Opt.1 model (optimized only for volunteer head kinematics), Opt.2 model (optimized only for cervical vertebral kinematics), Opt.3 model (optimized for both head and neck kinematics), and Opt.4 model (as Opt.3 but used one weighting factor for all cervical vertebral kinematics).

In general, all optimization simulations of these studies were numerically stable. The optimum value for Kp tended towards the lower bound of permitted parameter identification range except for Opt. 2. For Kd, the optimum values were at the lower boundary for most of the optimizations except for the Opt. 2 and the Cross-Validation optimization. The optimum values for the Tnd and the time constants describing the muscle activation dynamics were similar in all optimizations.

The parameter convergence and correlations were also analyzed in Study 2. An example of parameter optimization convergence is presented in Figure 4.3. All parameters converged before the last iteration. From the parameter correlation analysis, it was found that most of the parameters in all optimizations had a weak inter-parameter correlation (≤ ±0.29). Although, several medium correlations (±0.30 - ±0.49) were observed between parameters in some optimizations and only one strong correlation was found in Opt.2 (±0.50 - ±1.00).

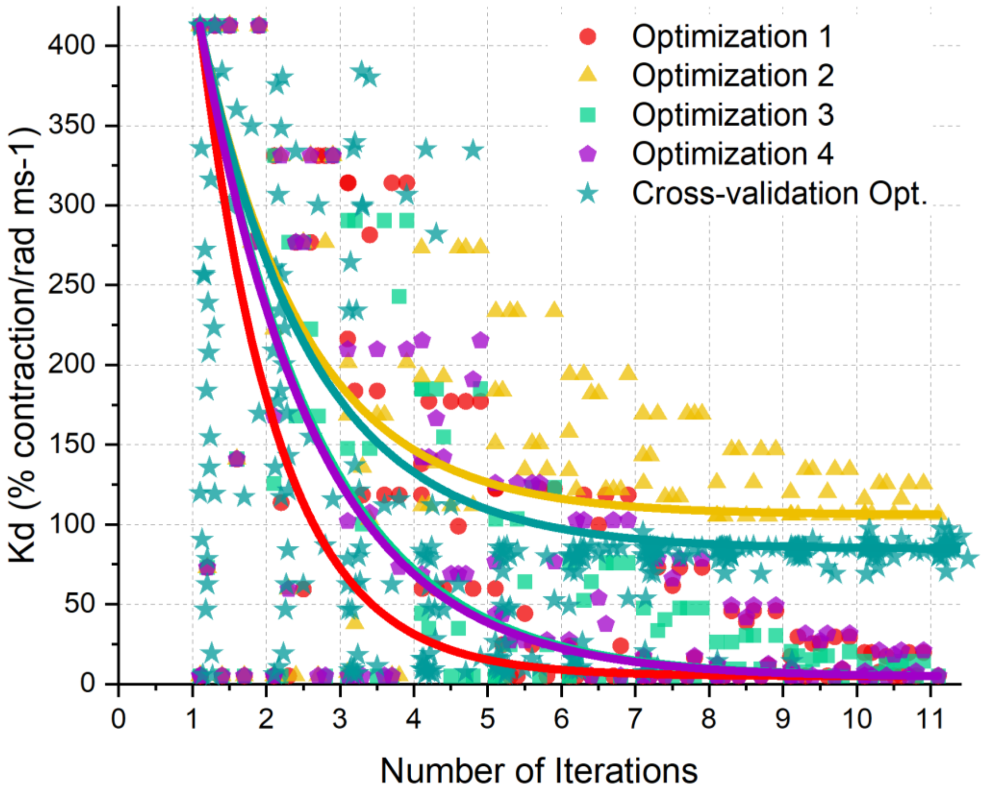
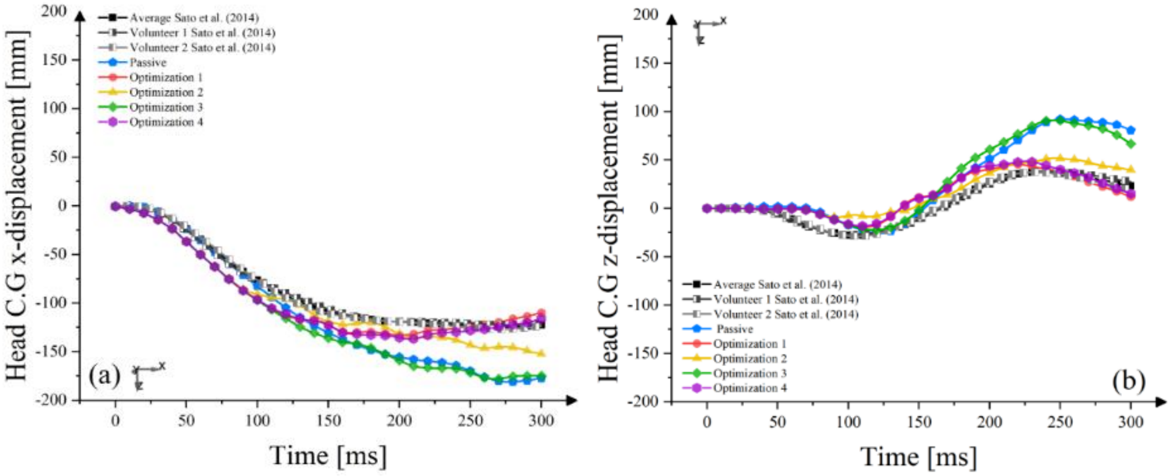


Figure 4.3 Parameter Convergence Plot (Study 2 - Paper 2)

Generally, the active muscle controllers altered the head kinematics by reducing the head peak displacements in all displacement directions except in the Optimization 3 approach (Figure 4.4). Head x-displacement in the Opt. 1, Opt. 2 and Opt. 4 approaches closely replicate the volunteer head motion until 220ms after impact. Meanwhile, Opt. 3 and the passive model had almost identical head horizontal motion. In the vertical direction, no models could fully capture the volunteer kinematics between 50-100ms after impact. The Opt. 1, Opt. 2 and Opt. 4 cases generally followed the volunteer head z-displacement trend, but with some discrepancies. Comparison of head rotational y-displacement highlighted that no models could mimic the volunteer motions until 125ms after impact although some models (Opt. 1, Opt. 2 and Opt. 4) could limit the head rotational motion.



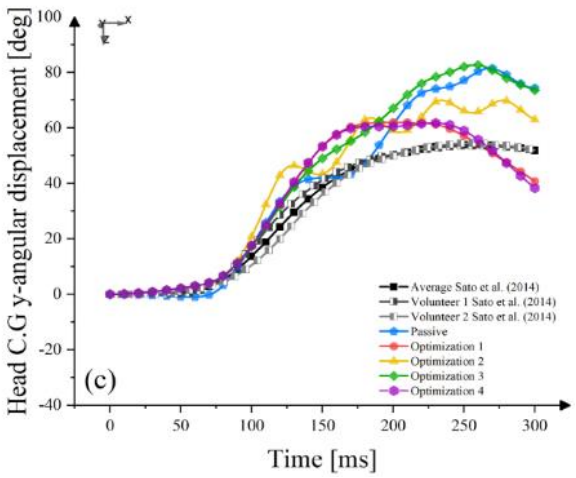
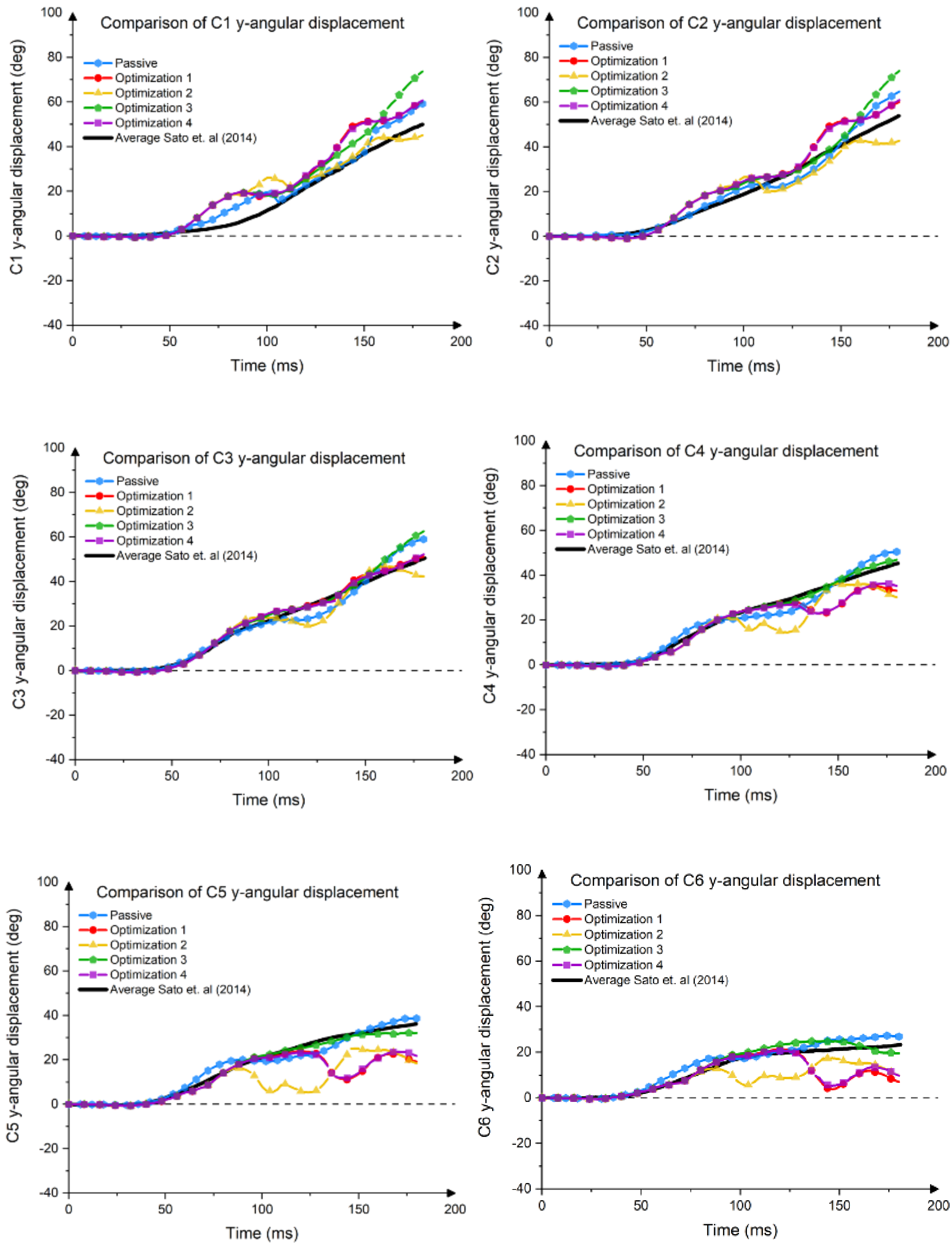


Figure 4.4 Comparison of Head Kinematics Calibration Simulation: (a) Head C.G x-linear Displacement; (b) Head C.G z-linear Displacement; (c) Head C.G y-angular Displacement.

Some of the optimization schemes did not improve the intervertebral cervical spinal kinematics correlation to volunteer data (Figure 4.5). In the period 50-100ms after impact, all models over predicted the rotational motion of the C1. After that, two models (passive and Opt. 2 models) could replicate the volunteers' kinematics until 150ms after impact. For the C2-rotations, no active model could follow the volunteer motion up until 100ms after impact. Both active and passive models could match the volunteers' C3 rotations up until 100ms after impact. When the C4 rotations were compared, the Opt. 1 and Opt.3 models could mimic the volunteer motions up until 120ms after impact. This trend was also observed for the C5 and C6 rotational displacements. All models could reasonably follow the volunteers' C7 kinematics up until 140ms after impact. After this time, most of the models could still replicate the volunteers' C7 rotational y-displacement but not the Opt.1 and Opt. 4 models.


Comparison of C1 y-angular displacement
Passive
Optimization 1
Optimization 2
Optimization 3
Optimization 4
Average Sato et. al (2014)
C1 y-angular displacement (deg)
Time (ms)
Comparison of C2 y-angular displacement
C2 y-angular displacement (deg)
Comparison of C3 y-angular displacement
C3 y-angular displacement (deg)
Comparison of C4 y-angular displacement
C4 y-angular displacement (deg)
Comparison of C5 y-angular displacement
C5 y-angular displacement (deg)
Comparison of C6 y-angular displacement
C6 y-angular displacement (deg)
100
80
60
40
20
0
-20
-40
0
50
100
150
200

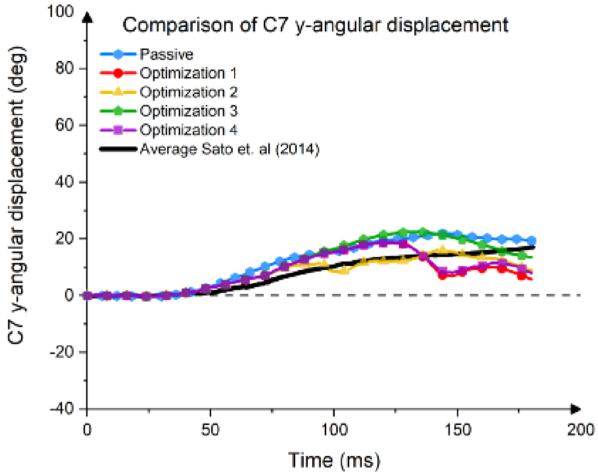


Figure 4.5. Comparison of Cervical Spine Kinematics (Study 2 - Paper 2)

4.4 Influence of Cervical Spine Alignment to Head Kinematics of Passive Model (Research Question 3 - Study 2)

The average female cervical spine alignment based on Sato et al. (2016) was more kyphotic (Figure 4.6a) compared to the original ViVA OpenHBM neck. In general, improved head kinematics were seen for the model with the updated cervical spine alignment based on a small population of volunteers (Sato et al. 2016). The difference in cervical spine alignment affected the head displacements in all directions, with a more pronounced difference observed in the z-displacement (Figure 4.6c). With the updated cervical spine alignment, the ViVA OpenHBM model could better reproduce the volunteer's upward motion, which peaked at 100ms after impact.

4.5 Kinematics of Model with Combined Muscle Controllers (Research Question 4 - Study 3)

Two muscle controllers were combined using two different approaches. The model with second approach (when APF controlled the superficial muscle and MLF controlled the deep muscle) followed the volunteer's head C.G displacement best in all directions compared to other models, although not entirely identical. All active models improved the head C.G displacements when compared with the passive model, except for the first scheme with adopted gains from Study 1. The best agreement of C3 y-rotational displacement was achieved by the second approach model, followed by the passive model. In C4 y-rotational displacement, almost all models had head kinematic responses close to the volunteer response except for the model with the approach and gains based on Olafsdottir et al. (2019). Until around 125ms, most models (except the model with first approach and older gains) could follow the volunteer motion with the model with first approach and newer gains followed the volunteer responses until 180ms.

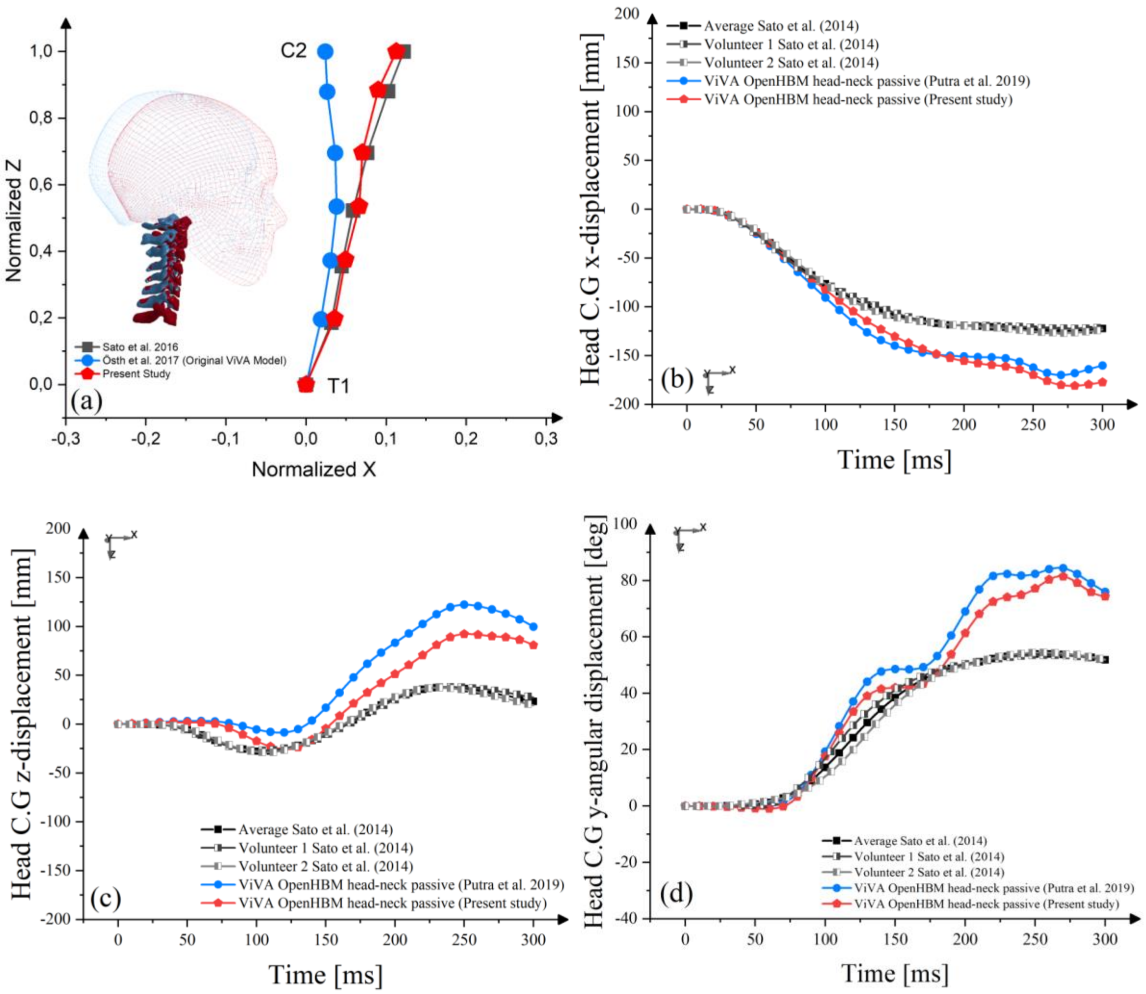
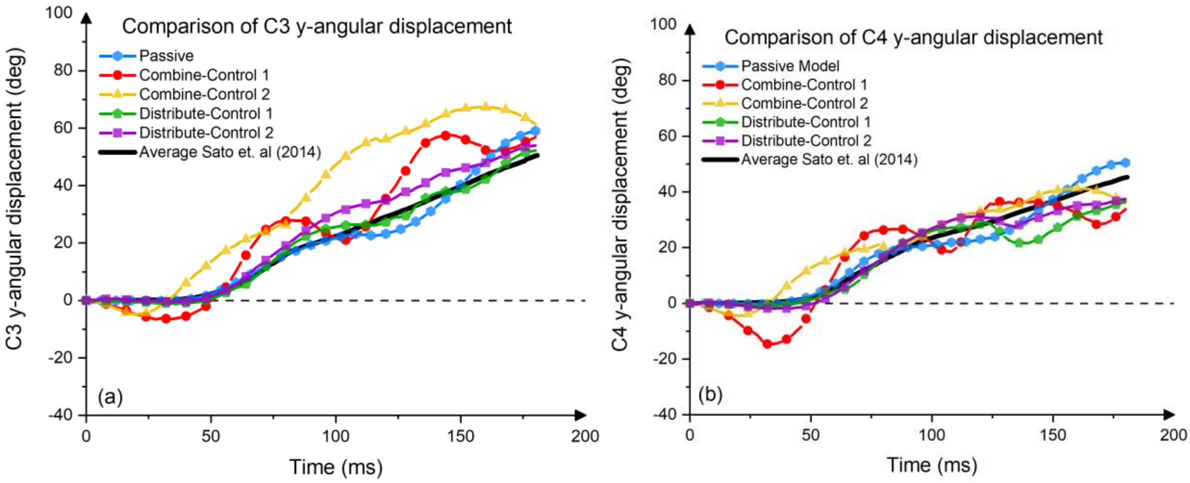


Figure 4.6. Influence of Cervical Spine Alignment to Head Kinematics of Passive Model. (a) Cervical Spine Alignment Comparison; (b). Head x-displacement; (c) Head z-displacement (negative displacement equals upwards motion); (d) Head ry-displacement.



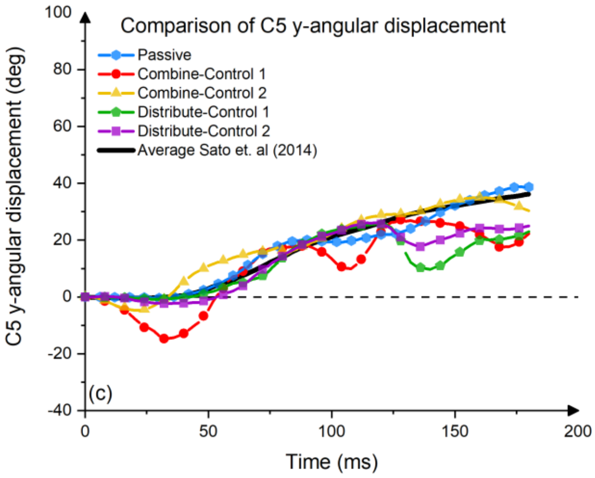


Figure 4.7. Cervical Vertebrae Rotational-y Displacement: (a) C3 rotational y-displacement; (b) C4 rotational y-displacement; (c) C5 rotational y-displacement.

4.6 Quantitative Rating Evaluation (Study 1 and 2)

In Study 1 (Paper 1), the model with the APF controller could increase the CORA score compared to the passive models, although the rating of head z-displacement was only improved slightly. In Study 2 (Paper 2, Figure 4.8), the model that was optimized against volunteers' head displacement (Op1. Val) had the best agreement in all head kinematics components. Meanwhile, the cervical spine kinematics of the Opt.3 model had the highest rating, although it was almost identical to the passive model. When both head and cervical spine kinematics were averaged, the Cross-Validation model had the highest agreement and nearly identical with the Opt. 1 Val.

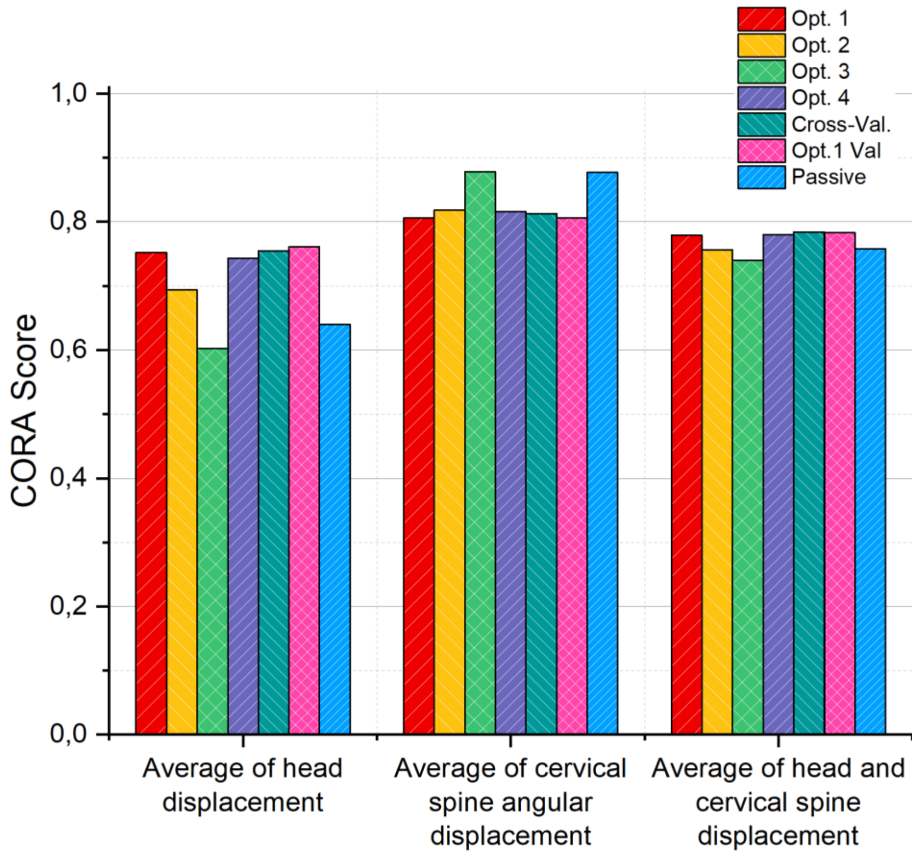


Figure 4.8. CORA Score of Study 2

Chapter 5 Discussion

The human reflex mechanism triggers muscle activity in the neck during a rear impact (summarized by Siegmund, G.P. 2011). Even though the current knowledge regarding human feedback and reflex mechanism is not well established, it has been highlighted that two reflex mechanisms called the VCR and the CCR exist and are integral to maintaining the head orientation (Armstrong et al. 2008., Cullen and Goldberg, 2014., Cullen, K.E., 2012. Keshner, E.A., 2003). Therefore, the active muscle modeling in HBMs should at least represent the behavior of these reflex mechanisms. To answer this challenge, the modeling of the VCR and the CCR reflex mechanisms were investigated and the influences of those reflex mechanisms in low-speed rear-impacts were analyzed in the present thesis. Hence, the present thesis contributes to the understanding of how modeling reflex mechanisms in whiplash-type motion affect the head-neck kinematics. This knowledge will also help in the research on potential whiplash injury mechanisms.

The implementation of a PID controller to represent the human neck muscle reflex system in the current study was adopted from Östh et al. (2012, 2015) and Olafsdottir et al. (2017, 2019). Another study from Iwamoto and Nakahira (2015) also implemented active muscles with closed-loop control representing the human muscular reflex system. But those previous studies have not focused on rear-impact collisions. Recently, Kleinbach, C.G. (2019) enhanced a similar model used in the present thesis by implemented an angle-based and muscle length-based active muscle controller. The author also validated the model using volunteer data in a low-speed rear impact. Although the model with active muscle controllers increased the model agreement when the head CG, T1, and pelvis kinematics were compared, the author did not conduct any evaluations for the intervertebral rotation in the cervical spine. The correct prediction of intervertebral rotation in the cervical spine is essential if a model will be used to study head-neck kinematics in whiplash-type motions.

The present thesis was arranged in four research questions. To answer the first research question, a comparison of two different muscle controllers was conducted as the first study (Paper 1). Based on the simulation results, it was found that the model with active muscle controllers improved the head C.G displacements agreement with volunteers' head kinematics. The improvement was achieved by reducing the peak displacements and following the volunteer time history displacements better than the passive model. Thus, it seems that the APF controller is more critical in the low-speed rear-impact scenario as it could simulate the head C.G displacement time histories better than the MLF controller. Non-realistic responses were also observed in the model with MLF controller. However, a drawback of the APF controller was buckling occurred in the model's cervical spine, which resulted in poor agreement of vertebral rotations with the volunteers. This buckling was nonphysical and caused a lower agreement of head vertical (-z) displacement between the model and the volunteer data. In this case, the MLF controller may help to prevent cervical spine buckling. The MLF controller could limit the APF controller's pulling force if it is beyond the normal range and consequently, avoids the nonphysical motion. Muscle length feedback control was observed by Happee et al. (2017) and Olafsdottir et al. (2017) to reduce cervical spine buckling, although their models were evaluated with lower severity load cases and also exhibited some non-realistic motions.

Based on the first study, it is important to continue research on exploring the combination of APF and MLF controllers. However, due to the computational cost of the MLF controller at that time, it was decided to continue the second study by exploring optimization methods to improve the APF controller performance. It was also hypothesized that a better understanding of the APF controller influence on head-neck kinematics as well as the optimization process to identify the controller parameters, the combination of APF and MLF controllers could be conducted in the most efficient and biofidelic way. The MLF controller itself is very complex because each muscle element has its own PID controller.

In the second study (Paper 2), besides the head's sagittal plane kinematics, the cervical vertebrae angular motion (flexion-extension) was also added as an objective of the optimizations. Combinations of these vertebral rotations and head kinematic data were used as the objective function of the optimizations. The optimizations successfully found the global optimum of each parameter. It was also found that the selected optimization parameters were less dependent of each other (Study/Paper 2). The optimization found that the derivative gain (Kd) was less influential on the model's kinematics. When the head C.G kinematics and the cervical spine kinematics were compared between the model and the volunteers, at least two APF controllers designs in Paper 2 had better agreement than the model in the first paper (Paper 1).

This second study also highlighted that the current method of approximating the VCR reflex by sampling a vector between the head C.G and the T1 center of the vertebral body was robust and could detect the changes in head motion well. However, it had less influence on the cervical spine kinematics. This was observed from the optimization results when the cervical spine kinematics were added as the objective of the optimization. Cervical spine buckling still occurred in the active model in the second paper, although it was much less than the model in Paper 1. Thus, an updated version of the current APF controller can be conducted by adding one or two additional vectors to capture the neck kinematics better. But, if this idea is implemented, the complexity of the model will increase, and consequently, more validation data are needed to optimize the controller's parameters.

In the same study (Paper 2), the initial cervical spine alignment of the model was changed to the average cervical spine alignment based on average female volunteers (Sato et al. 2016). The average cervical spine alignment was found to be more kyphotic than the original cervical spine alignment of the ViVA Model, which was more straight. The simulation results showed that the agreement of the passive model was improved in the vertical direction when the cervical spine curvature was adjusted. This result suggests that the cervical spine alignment is important to get biofidelic head and neck kinematics and is an important issue when differences in male and female response are studied.

An attempt to combine the APF and MLF controllers was conducted in Study 3 (Paper 3). Two simplified approaches were proposed to combine the APF and MLF controller. The first approach assumed that both APF and MLF controllers were controlling the activation of all neck muscles together (based on a review from Armstrong et al. 2008). In the second approach, the APF controller was used to regulate the activation of the superficial muscle. Meanwhile, the deep neck muscle was controlled by the MLF controller (See Paper 3). The second approach was developed based on the fact that a high density of muscle spindles was found in the deep cervical muscles (Amonoo-Kuofi, HS. 1983). Presently, this study was addressed as proof of concept. Therefore, the optimum gains based on earlier studies were used, and no optimizations simulations were conducted. The results from Paper 3 suggested that the gains from previous studies were not applicable when both APF and MLF controllers were used to control the same muscles. Therefore, the optimization-based parameter identification must be re-evaluated to derive both model parameters (APF & MLF). If the APF and MLF controller are used to control the muscles separately as in the second approach, gains from previous studies can still give a good reasonable result with an incremental improvement in the cervical spine kinematics. However, retuning of the controller gains is still needed in the second approach as part of the muscles that were controlled using the APF controller, now are controlled with the MLF controller. Most likely, this means that the gain for the remaining muscles needs to be adjusted (increased) to compensate for the muscle elements not controlled for.

In the same study (Study/Paper 3), the non-physical head rotation was observed in the active models (especially in the model with the first approach). The active models' head started to rotate backward after the settling with gravitational acceleration, but before the impact was started. This non-physical motion was currently assumed to be caused by the MLF controller because similar kinematics was not observed in the model with only the APF controller (Study 1 and Study 2). It is also supported by the fact that similar non-physical displacement was observed in the model with the MLF controller only (Study 1). Thus, this non-physical kinematics should be investigated further. However, due to the complexity of the MLF controller (it used 258 PID controllers), solving this problem was beyond the scope of the present thesis.

In the present thesis, kinematics data from two female volunteers seated in rigid seats (Sato et al. 2014) were used to derive the optimum muscle controller parameters. The reason was that the data based on Sato et al. (2014) were the most comprehensive published data that included cervical spine kinematics available during the present study. Since the kinematic responses were only based on two female volunteers, the kinematics data might not be representative of the average female population responses. In addition, the volunteers were also seated in a rigid seat, which might not be representative of a real-life scenario in which the occupant is sitting in an automotive seat including a head-restraint. It may be important to use volunteer tests that are seated in an automotive seat to derive the optimum parameters for the muscle controllers.

A separate MLF controller was implemented to control each muscle element in the present thesis. Consequently, there were 258 MLF controllers implemented. However, only one global gain was assigned to those controllers. This may be a limitation of the MLF controller in the present work. Therefore, it might be necessary to group the muscles into several groups and assign different controller gains. Another potential limitation may be related to muscle routing, as although the location of muscle origin and insertions were based on the anatomical description, the muscle lines of action are assumed to be straight. This approach may miss relevant complexities in the human CCR.

The methods to combine the APF and MLF controllers in the present study can be considered as a pragmatic approach. The knowledge of the human sensory-motor integration is still not well established (Armstrong et al. 2008., Cullen and Goldberg, 2014., Cullen, K.E., 2012. Keshner, E.A., 2003). Also, the human head-neck complex is controlled by separate systems that do communicate with each other. In fact, Blouin et al. (2007) found that multifidus muscles (one of the deep neck muscles) has a focused spatial tuning curve, which is more similar to the APF controller than the MLF controller. Therefore, further studies of controller tuning and the interdependency of APF and MLF characteristics are needed. A challenge for this approach is the availability of information that allows for the specific reflex actions to be identified and quantified.

Currently, the active muscle controller analysis conducted in the present thesis were mainly focusing on the effects of muscle activation to head-neck kinematics. It would also be beneficial to compare the muscle activation signals as the output from the active muscle controllers to the Electromyography (EMG) data from the volunteers. However, this comparison was not conducted in this thesis due to the lack of EMG data availability with a similar test setup. If this study can be done in the future, the activation signal from the muscle controller can be evaluated to make sure that it is within reasonable physiological ranges.

The present study has contributed to a better understanding of how to model and calibrate the active neck muscle controllers approximating the human VCR and CCR reflex during whiplash induced rear-impact. The main intention of the current study was to create a robust model, not an exact duplicate of the human reflex system. The optimization strategy used and explored in the present thesis could also add some knowledge if the same process is conducted in the future. Based on the studies in this thesis, the model that was called the Opt.1 Val model mentioned in Study 2 will be the model that can be used for future research.

In summary, an open source head-neck FE model that represents a 50th percentile female stature with active reflexive neck muscle is now available and with future development will be used to study potential whiplash injuries mechanism.

Chapter 6 Conclusions and Future Directions

6.1 Conclusions

Based on Study 1, it can be concluded that the angle-based input is the most effective and straightforward input to detect the head motion in a low-speed rear-impact collision. The single angular input was conveniently implemented in a controller that could be defined with available experimental data. The results of the active models were comparable to the volunteer responses. But the model with the APF controller could replicate the volunteer head displacement time histories better than the MLF controller. Consequently, an overall better agreement in head kinematics compared to the volunteers was achieved by the model with the APF controller.

Based on the study conducted in Paper 2, it can be concluded that reliable parameter identification can be obtained via optimization of the active muscle controller gains. The optimization using the published volunteers head kinematics as the objective showed the best agreement in head kinematics.

The influence of cervical spine alignment was also studied in Paper 2. It can be concluded that the cervical spine alignment influenced the vertical head displacements more in than other displacements.

Based on Paper 3, it can be concluded that combining APF and MLF controllers is not trivial, although this study gives a preliminary insight and idea of how to combine different muscle controller strategies in a single head-neck model. With further studies such as gain tuning and study of interaction between controllers, the distributed control could likely increase the biofidelity of the HBMs.

6.2 Future Directions

To fulfill the goal of understanding why females have a higher risk of sustaining whiplash injuries and how muscle activity influences the injury risk, four future studies are proposed (Figure 6.1). The first study following this thesis should focus on the validation of the muscle controller strategy by adopting a different volunteer test setup. Due to the limitations in the current third study, two routes can be followed for the validation study, either using the active muscle controller from Study 2 (APF Controller of Optimization 1 Validation) or Study 3. If the muscle controller from Study 3 will be used, new optimizations simulations must be conducted. The subsequent study should be a comparison between female and male kinematics to understand the differences between the two sexes. To do this study, the development of male FE HBM should be conducted. Both male and female FE HBM can be equipped with active reflexive neck muscles. Following the second study, accident reconstruction simulations based on real-world accident data will be conducted using female and male active HBMs. The whiplash injuries outcome using global injury criteria (such as Neck Injury Criteria, NIC) and tissue-based local injury criteria (for example, transient pressure in spinal root ganglion, facet joint capsular ligament strains) can be analyzed. Finally, based on the previous studies' knowledge, design principles to develop a gender equivalent whiplash protection seat design can be identified in the fourth future study. Ultimately, the present research will contribute to reducing the incidence of whiplash injuries.

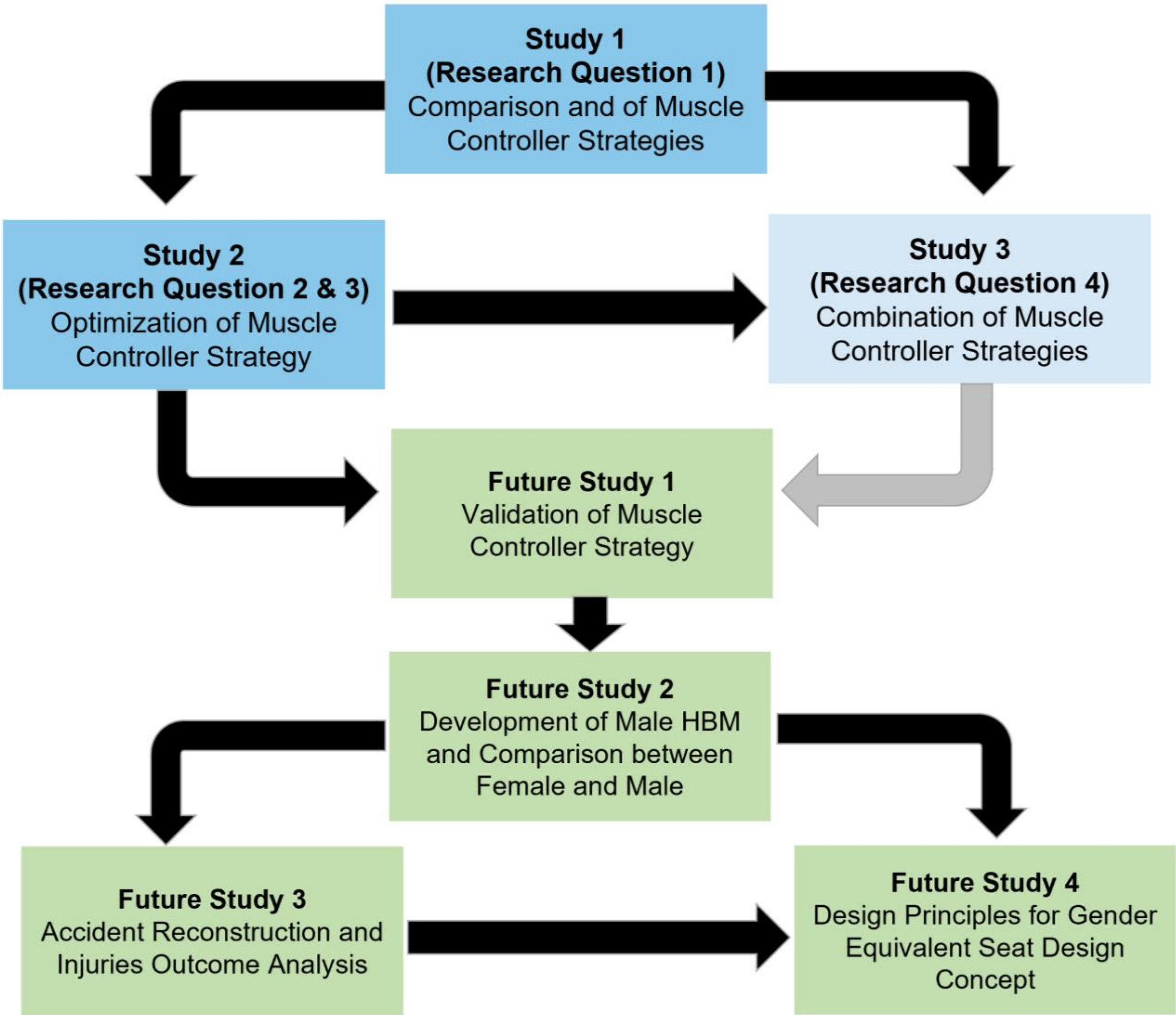


Figure 6.1 Future Direction of the Present Study

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Appendix 1 Updated Version of MLF Controller in Paper 1

A1.1 Problem

In Paper 1 (Putra et al. 2019), the Muscle-length Feedback (MLF) controller was implemented with a muscle spatial tuning pattern (Olafsdottir et al. 2015). Consequently, the activation level of each muscle was limited by the maximum activation level based on the directionally specific pattern for each muscle group. Another problem was the reference length of each muscle was defined based on original muscle length before the settling in gravity acceleration, not after the settling. This could cause the delay time as the result of optimization became slower than it should be. Therefore, the objective of the current appendix was to report the modifications that were being made in the MLF controller which was previously published in Paper 1 (Putra et al. 2019). This updated version of the MLF controller was integrated in Study 1 in this thesis.

A1.2 Method

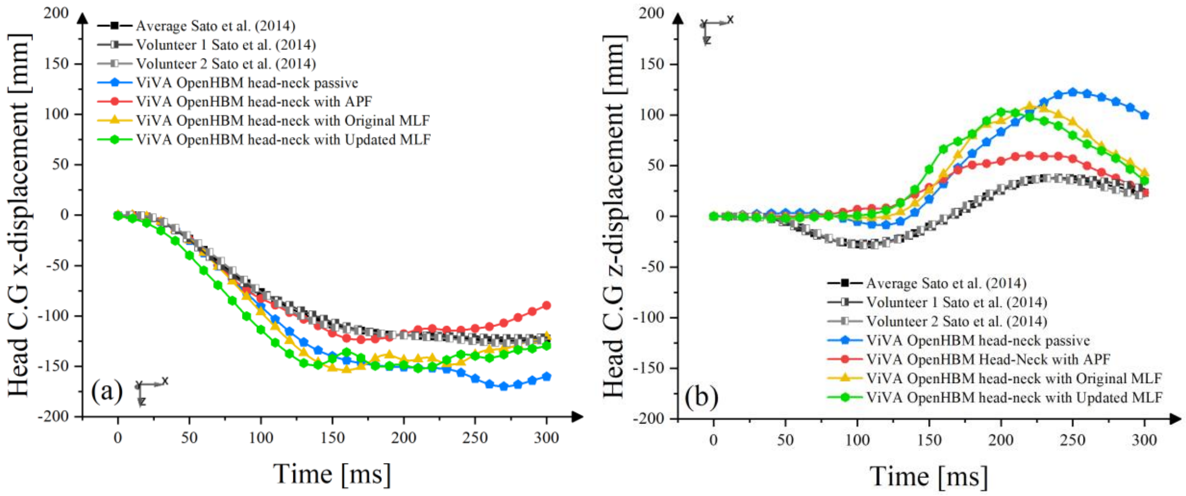
Modifications were conducted by removing the spatial tuning pattern in the controller algorithm, and the reference length of each muscle was changed to 100ms. 100ms was the time for the model settling. A new optimization simulation was conducted with the updated configurations. Meanwhile, other methodologies remained identical, as reported in Paper 1.

A1.3 Results

As expected before, the updated MLF had a faster delay than the original MLF (Table 1). Meanwhile, the Proportional gain and Derivative gain remained almost identical, with only small differences between the original and the updated controller. When head displacements were compared between two models, the model with Updated MLF had earlier peak displacements in all directions (Figure A.1). In head-x displacement, the model with updated MLF had an earlier response up until 100ms compared to the model with the original MLF controller. A similar trend as in the head x-displacement was observed in head rotational -y displacement. Meanwhile, the curves time-histories trend between the two models were still almost identical to each other.

Table A.1 Gains Comparison between Original and Updated MLF

Active muscle control strategy	Proportional gain (Kp)	Derivative gain (Kd)	Neural delay
Original MLF	0.45 (%contraction/mm)	6.71 (% contraction/mm s^{-1})	53.21 (ms)
Updated MLF	0.46 (% contraction/mm)	6.10 (% contraction/mm s^{-1})	19.94 (ms)



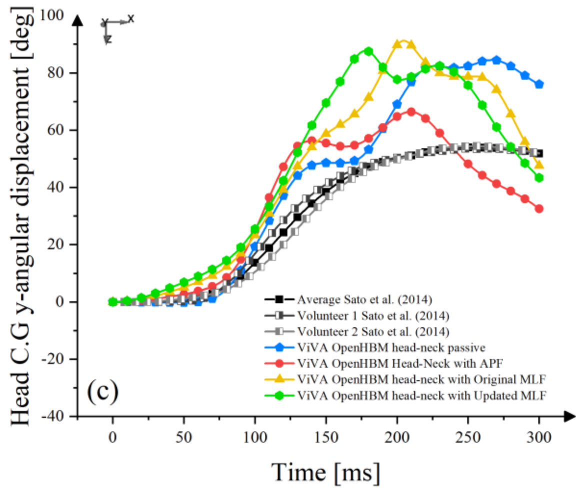


Figure A.1 Comparison of Head kinematics between models and volunteer. (a) Head C.G x-linear displacement; (b) Head C.G z-linear displacement; (c) Head C.G y-angular displacement

A1.4 References

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Part II Appended Papers

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