THESIS FOR THE DEGREE OF LICENTIATE OF ENGINEERING IN MACHINE AND VEHICLE SYSTEMS

Towards a human body model for prediction of vehicle occupant kinematics in omni-directional pre-crash events

EMMA LARSSON

Department of Mechanics and Maritime Sciences

CHALMERS UNIVERSITY OF TECHNOLOGY

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EMMA LARSSON

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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

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EMMA LARSSON
Department of Mechanics and Maritime Sciences
Division of Vehicle Safety
Chalmers University of Technology

ABSTRACT

As the vehicle fleet becomes more equipped with crash avoidance systems, the proportion of crashes preceded by evasive manoeuvres is expected to increase. In an evasive manoeuvre, occupant position and posture can be influenced by the induced loading. Therefore, there is a need to predict the occupant response from evasive manoeuvres. During evasive manoeuvres, the occupant kinematics can also be affected by muscle activity, and as such, taking the effect from active muscles into account in simulations of occupant response to evasive manoeuvres is important.

In this thesis, a method for activation of the neck and lumbar muscles in an active human body model, based on recorded muscle activity from volunteers, was enhanced and evaluated. The active human body model successfully predicted passenger kinematics in lane change, braking, and combined manoeuvres. As a step towards a model capable of predicting driver kinematics in evasive manoeuvres, the same method was adapted to control the shoulder muscles. The model with active shoulder muscles was evaluated in a simplified test setup. The active model successfully predicted peak elbow displacement for all loading directions.

Based on the results from the included studies, an active muscle controller based on directionally dependent muscle activity data can successfully predict kinematics from reflex response to loading in a finite element human body model. These findings represent an important step towards developing an active human body model able to predict occupant kinematics and muscle forces in omni-directional pre-crash events.

Keywords: Active Human Body Model; Pre-Crash Manoeuvres; Shoulder Muscle Control

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Emma Larsson

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LIST OF APPENDED PAPERS

Paper A	**E. Larsson**, J. Iraeus, J. Fice, B. Pipkorn, L. Jakobsson, E. Brynskog, K. Brolin, J. Davidsson Active Human Body Model Predictions Compared to Volunteer Response in Experiments with Braking, Lane Change, and Combined Manoeuvres *Proceedings of* *IRCOBI Conference 2019, Florence, Italy.* Author contribution: Formal analysis, Methodology, Visualisation, Writing – original draft
Paper B	**E. Larsson**, J. Fice, J. Iraeus, J. Östh, B. Pipkorn, J. Davidsson Development of a shoulder muscle feedback controller for human body models *Prepared for submission to Annals of Biomedical Engineering* Author contribution: Conceptualisation, Formal analysis, Methodology, Visualisation, Writing – original draft

CONTENTS

INTRODUCTION

For most of us, travelling by road is a given aspect of our everyday lives. However, despite the everyday nature of road travel, road traffic injuries are the leading cause of death among those aged 5-29, and are the eighth leading cause of death across all age groups, killing more people than HIV/AIDS or tuberculosis [1]. Beyond the devastating effects on an individual level, it also has a significant economic impact [2]. During 2010, the cost of road traffic accidents in the US alone was estimated to 1.6% of the US gross domestic product [2]. With an ever-increasing demand for personal mobility, and increasing motorisation of passenger transport, there is an urgent need to improve road traffic safety [1]. The UN has included road safety in their sustainability goals [3], in an effort to highlight the importance of this issue both from individual health and national economic perspectives.

Improvements to vehicle safety are continuous, with new safety technologies being continuously adopted in new vehicles [4]. Passive safety systems, e.g., airbags and seat belts, have prevented many road traffic injuries and road traffic deaths since their introduction [5]. More recent technologies are crash avoidance systems such as automatic emergency braking and lane-keeping assistance. Today, many vehicles are equipped with both of these systems [6], and both are included as a point of evaluation in several new car assessment programmes, e.g., Euro NCAP [7].

The development of passive safety systems rely on human surrogates, such as anthropometric test devices (ATDs), more commonly known as crash test dummies, or simulation models of the human body; human body models (HBMs). ATDs are mechanical surrogates of a human, developed to replicate human responses in specific load cases [8]. ATDs are reusable tools, and as such are designed to withstand many crashes with high repeatability. HBMs are virtual tools and are intended to represent the response in simulated crashes as close as possible to that of a human. HBMs typically include detailed representation of the skeletal system and adjacent soft tissues and can also include internal organs [8-11], and be used for tissue-level injury prediction. While ATDs have been important tools in the design of safe vehicles, HBMs have some advantages over ATDs in evaluating vehicle occupant protection. As ATD designs are limited by durability, repeatability, reproducibility and cost requirements, the ATDs are typically less representative of humans than state-of-the-art HBMs [12].

Many crashes are preceded by an evasive manoeuvre, such as braking, steering, or a combination of manoeuvres [13-17]. As vehicles increasingly are equipped with advanced crash avoidance systems, the number of crashes and their average impact velocity are expected to decrease [18], suggesting that evasive manoeuvres will precede a higher proportion of crashes. However, evasive manoeuvres such as braking or steering can affect the occupant's posture [19-21], potentially moving the occupant out of the standard seated position. In turn, these non-standard positions can affect the occupant to restraint interactions and affect the kinematics and injury outcome in a subsequent crash [22-25]. A posture where the occupant is leaning out of the seat belt has been shown to induce larger crash kinematics [23], and in a modelling study, it has been shown to increase injury risk [22].

During an evasive manoeuvre, typically producing accelerations of around 1 g [26], the load levels are low enough [27] and the duration long enough to allow that the occupant to respond to the posture/position change with muscle activation [28-30]. It has also been shown in modelling studies, mainly for the knee-thigh-hip area, that muscle forces can play a role in the predicted injury pattern [22,31-33].

One crash type that can have a severe injury outcome is road departure crashes [2,34], where the vehicle exits the road. This type has been over-represented in the number of motor vehicle crash related fatalities in the US, accounting for 32-57% of all fatalities, while comprising 12-16% of the non-fatal crashes [2,34]. In run-off-road scenarios, low and high level accelerations in most directions can arise, and the duration can be relatively long [35]. The combination of potentially long durations and low accelerations could indicate that muscles may influence the occupant kinematics in these types of accidents as well.

Studies have shown that a common preferred seated position in future vehicles is a reclined position [36,37]. Other studies have indicated that a reclined position could potentially increase the likelihood of spinal injuries [24,38], or increase the risk of abdominal injuries due to improper interaction with the lap belt [39]. To allow for a reclined seated position, new safety systems are being investigated, such as repositioning the occupant to an upright position prior to a crash. The repositioning concept has been investigated using simulations with an HBM (SAFER HBM v9), accounting for active musculature [33]. In the study, it was found that accounting for active muscles, although extrapolated from a different manoeuvre, influenced the results, highlighting the importance of including muscle activity when evaluating similar concepts.

Based on the information above, it becomes important when evaluating the vehicle safety performance of future vehicles to include occupant responses to pre-crash events, such as evasive manoeuvres or repositioning, accounting for the effects of active muscles.

Muscle physiology

To maintain posture and to control movement, active contraction of skeletal muscles is used [40]. The central nervous system (CNS) activates skeletal muscles either by voluntary or reflexive contraction. In response to a stimulus, the early response would be dominated by reflexes, while later, the response could be attributed to voluntary control [41,42].

Two reflexes that could be important for how an occupant would recruit their muscles include the vestibulocollic reflex and stretch reflex. The vestibulocollic reflex [43] activates neck muscles upon sensing linear or rotational accelerations of the head and aims at maintaining the head position in space. The stretch reflex contracts a muscle if a change in length of that muscle is sensed by muscle spindles inside the muscle belly [44] and aims to maintain the original length of the muscle.

To activate a muscle, the CNS sends an electrical pulse (action potential) through the motor neurons [40]. When the action potential reaches the neuromuscular junction, it triggers a chemical reaction, which in turn triggers an action potential that travels through the sarcolemma, surrounding the muscle fibres. This second action potential triggers another chemical reaction, which in turn contracts the muscle fibres within a motor unit (a group of muscle fibres innervated by one motor neuron). In response to a single action potential, the tension in the muscle fibre is quickly increased and then more slowly relaxed, Figure 1. If another action potential is received before the fibres are relaxed, the muscle fibres contract again increasing fibre tension force (wave summation), Figure 2 (a). If the frequency is high enough, no relaxation occurs (tetanus), and maximum tension within the muscle fibres is achieved, Figure 2 (b).

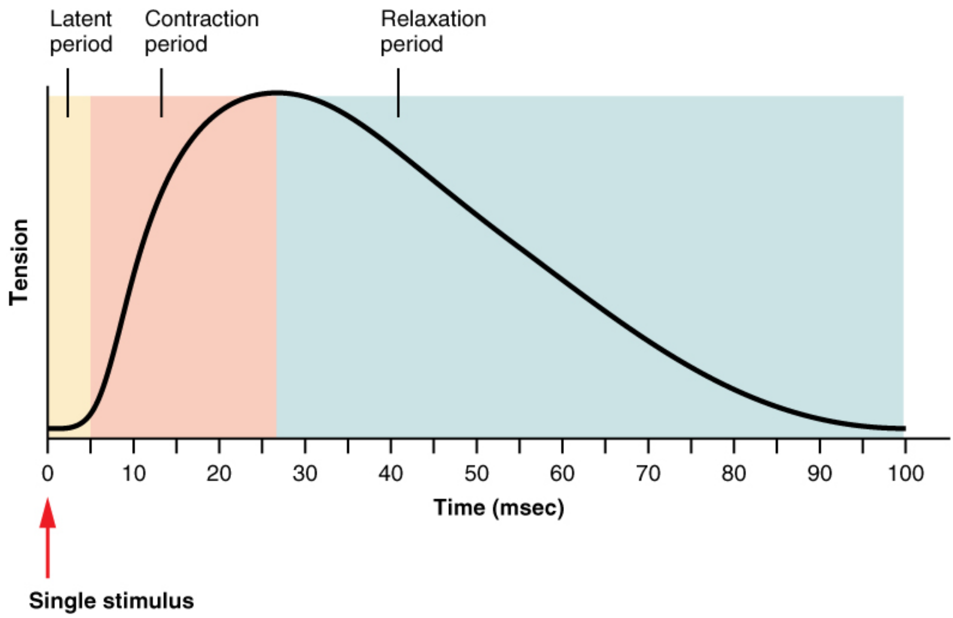

Figure 1. Muscle tension response to a single stimulus. Latent period, before any contraction in the muscle fibre occurs. Tension in the muscle fibre is built during the contraction period. In the relaxation period, the tension in the muscle fibre is relaxed back to no tension. Image from Anatomy and Physiology, Betts et al. [45] https://openstax.org/details/books/anatomy-and-physiology

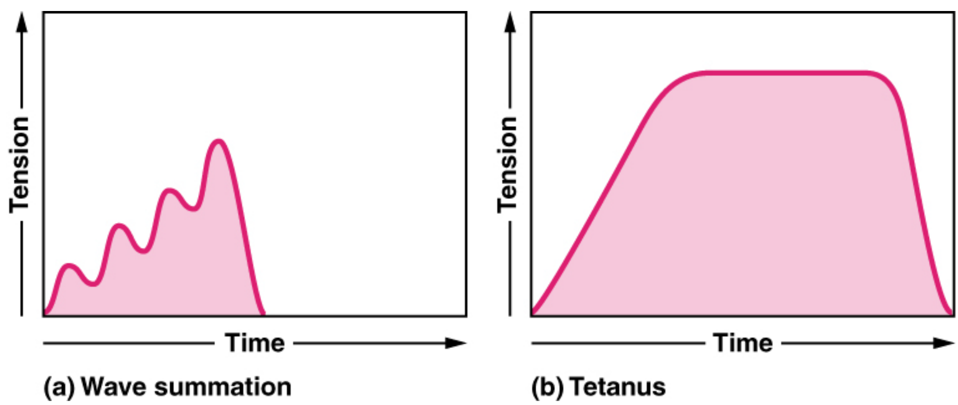

Figure 2. Muscle tension in response to several stimuli. (a) Increasing tension in the motor unit occurs if additional pulses are sent to the muscle before complete relaxation. (b) No relaxation occurs if the frequency of stimuli is high enough. Image from Anatomy and Physiology, Betts et al. [45] https://openstax.org/details/books/anatomy-and-physiology

Besides the increased tension in muscle fibres from increased action potential frequency, the strength of a muscle contraction can be increased by activating more motor units. The smaller motor units are activated first, leading to a non-linear relationship between the number of motor units activated and tension force produced [40].

The electrical activity related to muscle activation can be measured using electromyography (EMG) [40], either by placing electrodes on the skin or by placing needle electrodes inside the muscle. When measuring activity using EMG, the voltage of the action potentials that propagate in the muscle (and potentially surrounding muscles) are measured. EMG signals are often normalised to a maximum value, either by asking the subject to maximally contract that muscle (maximum voluntary isometric contraction (MVIC)) or by using the maximum recorded activity in the test that is being investigated.

Skeletal muscles produce movement by applying tension forces between bones (or, in some cases, skin) [45]. In many joints, movement is induced by rotation around the joints, and the muscles produce a moment around the joint, with a magnitude depending on the force in the muscles and the lever arms. The muscle that causes the movement is called an agonist. Synergists are surrounding muscles that assist the agonist, while antagonists are the muscles opposing the movement [45]. During movement, the antagonists might also co-contract with the agonist and synergists to stabilise the joint [46].

Active HBMs for pre-crash simulations

To model the pre-crash behaviour of occupants, HBMs have been fitted with active musculature and control systems to regulate muscle activation. There are several active HBMs (AHBMs) developed for use in pre-crash events. The current state-of-the-art models can predict both driver and passenger kinematics in braking and steering events. An important difference between a driver and a passenger is the drivers' ability to actively engage their arms during a manoeuvre. As such the difference between an AHBM modelling a driver and modelling a passenger is an active controller for the muscles spanning the elbow and shoulder joints.

The Global Human Body Models Consortium (GHBMC) (M50-OS v2 + Active) [47] and the Total Human Body Model for Safety (THUMS) (v6) [48,49] models have adopted similar control systems allowing for active control of the full body. Both models use several Proportional-Integral-Derivative (PID) controllers that respond to angle changes of joints in the body or angle changes between body parts. In the THUMS model, muscles in 16 regions are controlled by 36 PID controllers, while in the GHBMC model, muscles spanning the same 16 regions are controlled by 30 PID controllers. The THUMS has additional controllers in the ankles, elbows, and wrists. In the THUMS model, intermuscular load sharing was determined by anatomical descriptions from textbooks, while the intermuscular load sharing in the GHBMC model has been determined by studying the line of action in the model. The reference posture in THUMS was determined in local coordinate systems, e.g., head/neck joint angles are determined in a coordinate system attached to the torso, while a reference coordinate system has not been mentioned with regard to the GHBMC muscle controllers. The THUMS model, but not the GHBMC model, also includes additional PID controllers aimed at producing a bracing force in the hands and feet. Instead of these additional PID controllers, the GHBMC includes bracing by altering the reference posture. Both models were validated in the driver position using frontal sled decelerations, at 2.5 g and 5 g. In addition, a previous version in the THUMS model (v5) has been validated in a passenger position in lateral loading [50].

Another AHBM, based on THUMS (v3), has a hybrid muscle control system with both feedback and open-loop feedforward control [51], implemented in the lumbar region, neck, and arms. The feedback portion responds to muscle lengthening, while the open-loop feedforward has a pre-defined level of activation. Although a muscle length feedback controller does not have an explicitly defined reference posture, setting the reference posture with defined muscle lengths means that the posture maintained will be a local posture. The model was validated in the passenger position in lane change and braking [51].

In the SAFER HBM (v9) [52,53], lumbar, neck, and arm muscles are controlled by angle changes between body parts [54,55]. Muscles in six body regions are controlled by six PID controllers. Leg muscles can be activated with pre-recorded muscle activity data. In addition, muscle length feedback can be used to control the neck and lumbar muscles. In both neck and lumbar muscle controllers, the PID controller responds to angular displacement between two defined anatomical points, and intermuscular load sharing is based on directionally dependent muscle activation from volunteers. Neck and lumbar controllers aim at maintaining the posture in the global coordinate system. The model can predict full-body kinematics in braking for both driver and passenger position, and a previous version of the model has been validated in 1.1 g braking in driver and passenger position [55-57].

While the different models are all, to some extent, capable of modelling passenger and driver braking, the only model in which recorded muscle activity has been used in the development of the adopted controller is the SAFER HBM, where directionally dependent muscle activity is used in the neck and lumbar controllers [54]. Using recorded muscle activity is important to predict human-like muscle forces, because the human intermuscular load sharing cannot be determined solely from the muscles' geometrical location [58,59], humans co-contract surrounding muscles to maintain stability in the joints [40]. The co-contraction is essential for the shoulder joint, as it is the most freely moving joint in the body [40]. One modelling study found an improvement in predicted glenohumeral joint reaction forces of up to 45% when using recorded muscle activity in the muscle controller [60].

The neck and lumbar controllers in the SAFER HBM can respond to horizontal-plane loading. However, the reference posture is determined in the global coordinate system. This means that in an event where the occupant is subjected to large horizontal-plane rotations, the reference posture may become distorted. For instance, in a U-turn where the occupant is rotated 180^{o}, a forward-leaning reference posture would, after the rotation induced by the U-turn, become a rearward leaning posture. The THUMS and the hybrid controller models, on the other hand, use a purely local coordinate system. In an event where the reference body part is rotated, the reference posture will rotate with it. For instance, in a braking event where the torso is rotated forward, the head reference posture will also rotate forward.

Validation data for active HBMs

To ensure the applicability of AHBMs for use in designing safer vehicles, model tuning and validation are required. Typically, tuning and validation of an AHBM are through simulation of low-loading volunteer experiments, such as volunteer braking or steering tests [47,48,50,51,55-57]. Tuning and validation should be done with different sets of data; thus in every load case at least a tuning and a validation data set is required.

Numerous experiments with volunteers have been carried out, suitable for validation data for AHBMs. Several frontal deceleration tests in laboratory environments have been performed. Ejima et al. performed frontal decelerations in a sled test [61,62]. Head, torso, arm, and leg kinematics were recorded, and muscle activity was recorded from eight muscles bilaterally. EMG was not normalised to MVIC. Beeman et al. performed a sled test with volunteers, post-mortem human surrogates (PMHS), and an ATD in a driver position. Head, torso, arm, and leg kinematics were recorded, muscle activity and was measured with EMG in 20 muscles[63]. Mathews et al. performed frontal sled tests with paediatric and adult volunteers [64].

There have also been laboratory experiments with volunteers subjected to lateral loading. Van Rooij et al. exposed volunteers to lateral accelerations in a test vehicle, with the volunteers seated in the driver and passenger positions [65,66]. Head, torso, and extremity kinematics was recorded, and bilateral muscle activity was measured in four muscles, using EMG. Holt et al. performed lateral sled tests with paediatric and adult volunteers. Head, torso, and extremity kinematics were recorded. EMG was collected for four muscles bilaterally and three muscles unilaterally, and signals were normalised with MVIC. Arbogast et al. performed sled tests[67] with paediatric and adult volunteers. Head, torso, and extremity kinematics were recorded, while muscle activity was not recorded.

Tests performed in vehicles, either on a test track or on public roads, may present a situation to volunteers that is more representative of the daily riding conditions than a test performed in a laboratory. There have been several in-vehicle tests exposing volunteers to braking. Carlsson and Davidsson exposed volunteers in driver and passenger positions to autonomous braking when driving on regular roads [68]. In the driver position, the volunteers performed driver-initiated braking as well. Head and torso kinematics were recorded, while muscle activity was not recorded. In a test series by Österh and Ólafsdóttir et al. [28,29], drivers and passengers were exposed to autonomous braking when driving on regular, rural roads. In the driver position, the volunteers performed driver-initiated braking as well. Head and torso kinematics were recorded, and muscle activity was measured bilaterally in eight muscles (neck, lumbar, shoulder), using EMG. EMG was normalised with MVIC.

More recent tests have been performed in vehicles, exposing volunteers to lateral manoeuvres. Reed et al. performed braking, lane change, and combined manoeuvres with volunteers in the passenger position in three different vehicles [21]. Head kinematics were recorded, while muscle activity was not recorded. Huber et al. performed lane change and braking manouvres with passengers[10-12]. Head and torso kinematics were recorded for all volunteers, and for a subset of volunteers, EMG was measured bilaterally for eight neck and lumbar muscles. EMG singals were not normalised with MVIC. Ghaffari et al. performed lane change and braking manouvres with volunteers. Head and torso kinematics were recorded, together with EMG normalised with MVIC, from 19 muscles bilaterally [19,30].

There are numerous tests with passengers and drivers in braking scenarios, as well as with passengers exposed to lateral loads. However, data with drivers exposed to lateral loading is scarce. In-vehicle tests are preferrable for AHBM tuning and validation because, compared to sled tests, they are more representative of real-world pre-crash maneuvers. If muscle activity has been recorded, MVIC normalised EMG signals are preferable to non-normalised signals.

Objectives

The motivation for this system development was to build a tool that can aid the design of safer vehicles. AHBMs can aid by accurately predicting the occupant posture and muscle forces when transitioning from a pre-crash event into a crash. This can be used to either develop safety systems that prevent the occupant from moving from the optimal position during a pre-crash event or by allowing evaluation of passive safety systems for an occupant in a realistic non-optimal position. In order to ensure applicability of the AHBMs, the models need to include representative muscle activity levels while at the same time producing representative kinematics.

The objective of this Ph.D. is to aid in the design of safer vehicles by further developing the active SAFER HBM, such that it can predict occupant kinematics in omni-directional pre-crash events. Five steps, listed below, have been defined to reach this objective. The first two have been addressed to date and will be presented in this licentiate thesis as a step towards a model capable of predicting driver kinematics in horizontal-plane pre-crash manoeuvres.

- Enhancing existing SAFER HBM as a passenger in horizontal-plane pre-crash manoeuvres, such that it can be used in simulations with large horizontal-plane vehicle rotations.
- Developing a method of controlling the shoulder muscles that can be used in the development of a model that can predict driver kinematics in pre-crash manoeuvres.
- Extending the AHBM capabilities to predict occupant response to vertical loading.
- Validating the AHBM in events with horizontal and vertical pre-crash scenarios.
- Developing/enhancing controllers to handle repositioning from a reclined to an upright posture.

The horizontal-plane response was included before any vertical response because horizontal-plane manoeuvres can occur in isolation, while vertical loads are often accompanied by horizontal-plane loads. The first of the objectives were selected as a starting point because neck and lumbar controllers are required for both passengers and drivers. In order to have the same prediction capabilities for both drivers and passengers, developing a new shoulder controller was selected as the second objective.

SUMMARY OF APPENDED PAPERS

Paper A

The aim of Paper A was to enhance the active neck and lumbar muscle controllers of the SAFER HBM v9 and compare the occupant kinematic predictions to volunteers in braking, lane change and combined manoeuvres.

Enhancements were made to the neck and lumbar controllers implemented in the SAFER HBM v9, where one controller was implemented to emulate reflexes from the vestibular system, i.e., angular position feedback (APF), and one to emulate the stretch reflex in muscle spindles, i.e., muscle length feedback (MLF). Enhancements were made to the APF part of the control system, where updates were made to the reference coordinate system in which the reference posture is determined. Three different reference coordinate systems were implemented in the HBM, and model performance was evaluated for the three different reference coordinate systems.

Whereas the original implementation aimed at maintaining the posture in the global reference system, the enhanced models aimed at maintaining a set posture in either 1) a completely local reference system, 2) the vehicle coordinate system, or 3) the gravity field but rotating with the HBM around gravity direction.

The three different APF controllers were evaluated in a combined lane change and braking load case. One of the APF configurations was compared to volunteers in braking, lane change, and combined lane change and braking. All three directions were evaluated using two different seatbelt configurations: a regular seatbelt and a belt with an electrical pre-pretensioner, yielding a total of six load cases. The kinematic predictions and muscle activation signals were objectively evaluated using CORA. The kinematic CORA results ranged from 0.78 to 0.88 for the active models and 0.70 to 0.82 for the passive configuration.

It was concluded from the study that the active muscles improve the predictions compared to using the model in a passive configuration for some load cases, while for other load cases, only small differences were seen. The largest difference between active and passive models was seen in combined lane change and braking with a standard seatbelt. The best correlation to volunteers for the active model was seen in combined lane change and braking with pre-pretensioned seatbelt.

Paper B

The objective of Paper B was to develop a method of controlling shoulder muscles in HBMs, based on human physiological data, to be used to model drivers in horizontal-plane evasive manoeuvres. The aim of the study was to predict human-like elbow displacements when exposed to dynamic loading to the elbow.

In the study, 179 beam elements were updated or added to the right shoulder of the SAFER HBM v10. The controller strategy from Paper A was adapted to the shoulder. The muscles spanning the glenohumeral joint were controlled with angular position feedback (APF), with intermuscular load sharing based on directionally dependent muscle activation data from volunteer experiments. The muscles spanning the scapulothoracic joint were controlled using muscle length feedback (MLF).

The model was evaluated by simulating a volunteer experiment in which a dynamic load was applied to the elbow in eight directions using a weight drop. Peak elbow displacement, time to peak elbow displacement, and detailed elbow kinematics of the model were compared to results from the volunteer experiments. A sensitivity study was performed to show the effect of varying the gains of the APF controller.

It was found that the active controller reduced peak elbow displacement for all directions, and for two of the gain combinations, the model was capable of producing peak displacements within one standard deviation of the volunteers, in all eight directions, with a time to peak within one standard deviation in four of the directions. The successful prediction of peak elbow displacement showed that the controller is ready to be implemented and evaluated in full-body driver simulations.

DISCUSSION

In this thesis, an active HBM was enhanced and evaluated. In the first study, the neck and lumbar controllers were enhanced, and results from the full-HBM with active neck and lumbar muscles were compared to volunteers in braking, lane change, and combined lane change and braking. In the second study, the controller strategy from the neck in the active HBM was applied to the shoulder muscles, and the effectiveness of the controller was evaluated by comparing the model and volunteer elbow kinematics.

The motivation for this work was to build a tool that can aid the design of safer vehicles. AHBMs can aid the design of safer vehicles by accurately predicting the occupant posture, kinematics and muscle forces when transitioning from a pre-crash event into a crash. This can be used to either develop safety systems that prevent the occupant from moving from the optimal position during a pre-crash event or by allowing evaluation of passive safety systems for an occupant in a realistic non-optimal position.

Evaluating kinematics

Injury risk has been indicated to change with the pre-crash posture of the occupant. As such, it is important to accurately predict the occupant posture, such that a representative position/posture is used for the crash simulation. To determine if the posture is representative, the kinematics need to be evaluated. In both Paper A and Paper B, the kinematics of the model have been compared to the volunteers.

In Paper A, the model was objectively evaluated by comparing the displacement time histories from model to volunteers using CORA, while in Paper B, the model was evaluated by comparing elbow displacement time histories as well as peak elbow displacement and time to peak elbow displacement to volunteers. Different measures of comparison were applied as different aspects of the controllers were evaluated. In Paper A, the time-displacement comparison was used to allow for evaluation of the full-sequence model behaviour. For a pre-crash manoeuvre simulation, accurate prediction of kinematics is required during the full sequence, as a subsequent crash could happen at any stage during the manoeuvre. For the shoulder controller, however, the aim was to investigate if the controller implementation produced a human-like directional response to dynamic point loads. Therefore, the peak displacement and timing was used for evaluation of different controller parameter settings, as it clearly shows model sensitivity to controller parameter settings. When evaluating the performance in a pre-crash manoeuvre, however, one of the objectives in this Ph.D., a metric more similar to that used in Paper A, will be used.

In Paper A, the head kinematics were compared to those of the volunteers, revealing that combining APF and MLF controllers lowered the correlation to volunteer kinematics, although these changes were small. Based on those results, the MLF controller did not add any value. However, as shown by Putra et al. [69] and Ólafsdóttir et al. [54], an MLF controller used for the cervical muscles prevents vertebral rotation and spine buckling. In Paper A, the spinal curvature was not evaluated. During further model development and validation, it would be of importance to include evaluation of the spinal curvature to ensure that non-physical vertebral rotations are avoided.

In Paper A, only the first turning phase in the lane change simulations was used for objective evaluation of the model, although the second phase was still included in the simulation. The second phase was excluded from the objective evaluation since the controllers were intended for simulation of reflex responses only, and the second phase occurred late enough in time for voluntary control to occur. In the second phase, the model showed a greater head lateral displacement compared to the volunteers. The model displacement peak magnitude is relatively similar during the first and second phase, while the volunteers displaced less in the second phase compared to the first phase. In another study where lane changes were performed both to the left and to the right [20], the same trend of lower displacements in the second phase was seen, however, more in one direction than the other. In that study, a difference in displacement magnitude was seen in the first phase depending on direction, where inboard excursions were greater than outboard excursions. In that study, it was hypothesized that the difference in displacement could be related to the volunteers trying to avoid the B-pillar. In the study used for the comparison in Paper A, [30], the volunteers moved inboard in the first phase and outboard in the second phase. Thus, the B-pillar potentially contributed to the lower displacements in the second phase. Therefore, before extending the model to handle long-duration or multiple events, it would be of interest to understand the cause of the lower displacement seen in the volunteer experiments, as this could guide the model development.

Evaluating muscle activation signals

Forces from muscles have been shown to alter injury patterns for the knee-thigh-hip area [31]. As injury patterns may be sensitive to muscle force, it is important to validate the muscle force levels in the HBM. Measuring forces in individual muscles in a human is difficult. Instead, muscle activity is measured using EMG, which measures the voltage of the action potential that activates the muscle. Subsequently, this leads to significant oscillation in the EMG signals, and a filter is usually applied to the signals. On the contrary, muscle activity signals in the HBM, on the other hand, are relatively smooth, as it responds linearly to displacements and displacement velocity. The discrepancy between how a human activates their muscles and what the controllers produce makes it difficult to compare muscle activity between volunteers and the HBM. An attempt to compare EMG signals to muscle activation signals was made in Paper A. In the study, CORA was used to compare the activation signal to individual EMG signals and then averaging the CORA scores. Although the comparison indicated how well the control signals correlated to the volunteers, the method would benefit from being improved, accounting for the differences in how signals arise.

Implementation of controllers

The implemented controllers were aimed at modelling a reflex response, as would be seen for an unprepared occupant. If the manoeuvre is driver-initiated, the driver will be prepared for the manoeuvre, and thus the implemented controllers might not be representative of driver behaviour. In some studies it has been shown that during crash avoidance manoeuvres or driver-initiated braking, drivers brace themselves against the steering wheel [28,70]. As proposed by Östh et al. [55], it would be viable to include the bracing by adapting the reference posture, an approach that might be feasible with the new controllers as well.

The APF controller was initially created to emulate the vestibulocollic reflex [54]. This reflex responds to translational and rotational accelerations of the head and predominantly affects the neck muscles [43]. The original APF implementation maintained the posture of the head in the global coordinate system, which lets the controller respond to similar types of input as the human would. In Paper A, the reference coordinate system was updated to allow for simulations of cases with large vehicle rotations, such as U-turns. It was concluded that a hybrid reference system, partially connected to the global coordinate system and partially connected to the HBM, was the most suitable for the neck and lumbar muscle controllers. Updating the reference system was a pragmatic approach to allow for simulation of more complex scenarios but moved the controller one step away from the reflex it was intended to emulate. When adapting the APF controller from Paper A to the shoulder muscles in Paper B, a completely local reference coordinate system was used, as the shoulder muscles are not controlled by the vestibulocollic reflex. Changing the reference system moved the control system one step further away from emulating the vestibulocollic reflex, and instead the controller is more similar in function to the stretch reflex [44].

In previous implementations in the neck and lumbar muscle controllers, the stretch reflex was modelled using MLF. Since the implementation in the shoulder controller is similar in function to the stretch reflex, the MLF approach could have been used. However, with a pure MLF implementation, it is difficult to capture the co-contraction that humans use to maintain stability. In the shoulder, it has been shown that basing model muscle activation on recorded muscle activity and including co-contraction at a human-like level improved correlation to measured in-vivo glenohumeral forces by up to 45% [60]. In the implementation in Paper B, the scapula was controlled with MLF only, meaning that the co-contraction effect was neglected in scapula control. Ideally, the scapula would be controlled by an EMG-driven approach as well. However, the data available contained only scapula retractors and no scapula protractors due to difficulty in measuring protractor activity with surface EMG. To maintain a uniform method of controlling the scapula, the MLF approach was chosen for all scapula muscles.

Although the APF controllers were developed to emulate vestibular and stretch reflexes, the implemented controllers are simplified, representing pragmatic approaches instead of perfect analogies to the human motor/postural control system. A human could respond to numerous types of input in certain vehicle manoeuvres that are currently not included in the model's feedback system, such as tactile, visual, or auditory input.

In neither Paper A nor Paper B, the gains of the controllers have been tuned to better match the volunteers. As was shown in Paper B, controller performance can be sensitive to the gain selection. Putra et al. [71] tuned the gains of one controller, similar to the APF controller in this thesis, by optimisation. In that study, four optimisation loops were run with different constraints and objectives, and a difference in best gains was found depending on objectives and constraints. This shows the importance of carefully selecting what measures to tune the model towards. When tuning the controllers of the SAFER HBM, a suitable load case, as well as suitable objectives and constraints, must be selected. For the braking cases, as well as for passenger posture in lateral loading, there have been enough in-vehicle volunteer tests to perform both tuning and validation. For the driver in lateral loading, however, there are fewer tests available. For the shoulder, it might therefore be necessary to tune the controller only to a braking case and use a laboratory test setup for validation in the lateral direction.

In Paper A, it was found that the APF controller is limited in responding to rotational displacements, as the head is somewhat free to rotate without changing the relative positions of the two nodes in the SAFER HBM making up the reference vector; the head centre of gravity and T1. For the controller, a pure axial rotation of the head or neck would not be sensed at all, while lateral bending and flexion/extension of the neck would be indirectly controlled for, as these often produce a translation of the head centre of gravity relative T1. It is unlikely, although possible that isolated lateral bending or flexion/extension of the head without centre of gravity linear displacement would go unnoticed by the controller. When adapting the same approach for the shoulder in Paper B, this limitation is also carried over. However, for the humerus displacements, this is less of a concern, as the two nodes building the reference vectors are connected to the same structure; the humerus. Instead of the controller possibly being unresponsive to three rotations, as for the head, the shoulder controller is unresponsive only to internal/external rotations of the humerus. When modelling a driver, the hands will be connected to the steering wheel, and as such, any internal/external rotation of the humerus would also give rise to adduction/abduction or flexion/extension and thus give rise to a controller response.

In the feedback loops in both studies, there were several components: Proportional-Derivative (PD) controller, saturation, spatial tuning, activation dynamics, and baseline activity. Although all components are important for the controller, the sequence in which they should occur was less straightforward. The order was updated between Paper A and Paper B, Figure 3. In Paper A, baseline activity was placed last and was used if the control signal was below the baseline value, while in Paper B, the baseline activity was added on top of the control signal and was placed before the activation dynamics to prevent discontinuities in the signal. In Paper A, the saturation was placed last, while in Paper B, the saturation was placed before spatial tuning to ensure that the muscles always maintained the load sharing based on the volunteer data.

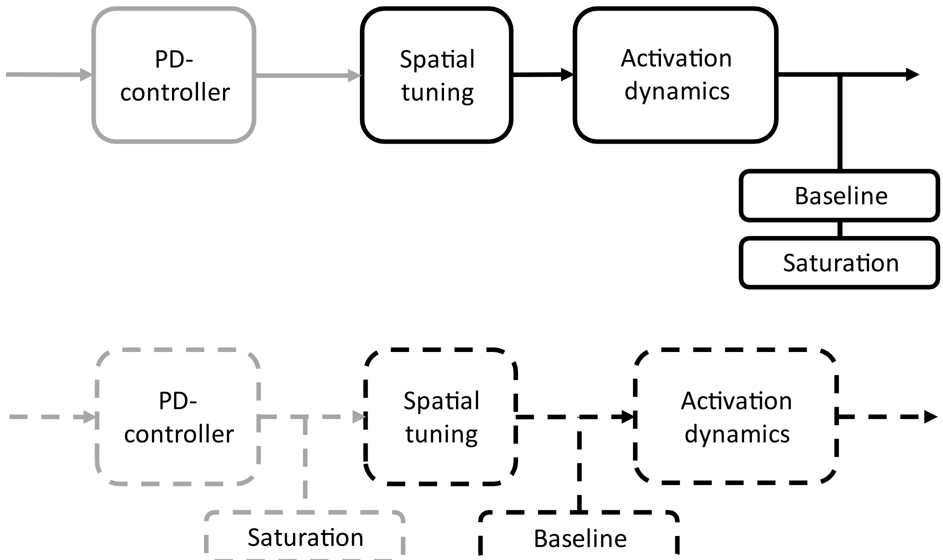

Figure 3. APF controller sequence in Paper A (top, solid lines) and Paper B (bottom, dashed lines). Grey lines indicate signals common for all muscles, while black lines indicate muscle-specific signals.

If the saturation was placed after the spatial tuning, an increasing signal requesting above maximum activity would saturate the agonist before the other muscles. Consequently, the other muscles would increase in activity while the agonist activity would remain at maximum, leading to distorted scaling where eventually all muscles would be at the maximum activity level. It could also lead to a situation where the controller no longer has the ability to control the movement, as the movement created by the controller might be very different in direction compared to expectations of the controller, creating a positive feedback loop. Thus, the saturation point was placed before the spatial tuning, and spatial tuning patterns were scaled accordingly. For a signal that never requires saturation, the placement of saturation is irrelevant, and volunteer EMG data reveal that for most muscles and most cases, the muscle activity of volunteers remain below MVIC [28-30].

To prevent discontinuity in the activation signal derivative, the muscle activation dynamics were placed after saturation, ensuring a smooth transition between ramping up and maximum contraction. Adding baseline activity represents the final operation before the signal reaches the muscle. Ideally, this operation should have been placed before reaching saturation, as adding it after reaching saturation lets the activity increase above maximum activation. As the saturation was placed on the global PD response signal, and baseline activity was added on a muscle group level, placing the baseline activity operation before saturation would have also required doing spatial tuning before the saturation. As argued above, reaching saturation post spatial tuning could lead to a positive feedback loop, and as such, the baseline activity was placed after saturation, even though it could allow for a signal requesting above-maximum activity from the muscle.

CONCLUSION

In this thesis, a method of controlling muscles in an HBM based on recorded muscle activity from volunteers has been developed and evaluated. The first objective was addressed in Paper A, where it was concluded that an HBM with active neck and lumbar musculature could predict passenger head and torso kinematics in horizontal-plane evasive manoeuvres while allowing for large vehicle rotations. The second objective was addressed in Paper B, where it was concluded that an HBM with an active shoulder muscle controller could successfully predict elbow peak displacement when subjected to dynamic loading to the elbow. The three muscle controllers included in the thesis, together with the elbow and leg muscle controllers already included in the

SAFER HBM, form a base for an AHBM capable of predicting driver and passenger kinematics in horizontal-plane pre-crash manoeuvres.

Based on the results from both papers, it can be concluded that an active muscle controller based on directionally dependent muscle activity data can successfully predict kinematics from reflex response to loading in an FE-HBM. These findings represent an important step towards developing an AHBM that can predict occupant kinematics and muscle forces in omni-directional pre-crash events.

FUTURE WORK

A natural step in continuing the development process of the active SAFER HBM will be to add more load cases to the HBM capability. Hence current implementations will need to be updated or extended. During the continuation of this Ph.D., the additional three aims will be targeted.

One common pre-crash/crash event is run-off-road [15], a scenario where accelerations in all directions can arise [35]. The current model is capable of handling horizontal plane loading only. Therefore, an important extension to the model will be the capability of handling vertical loading. This will likely require updates/extensions to several of the controllers and will also require additional data. The additional data would ideally be collected from volunteers exposed to low-level vertical loading in controlled tests in a representative vehicle environment.

When updates are in place, the controllers should be tuned to volunteers in representative load cases. After tuning, the HBM validation will require using a different set of volunteer tests. These two different sets of tuning and validation data will need to comprise representative load cases, including braking, lane change, and events with vertical loading. During the validation, it would be of interest to find a more suitable way of comparing the muscle control signal to the EMG signals, to compare muscle activity in the load cases where EMG signals are available.

To handle evaluation of future seating configurations, the model will be updated with a controller able to handle repositioning of the occupant from a reclined to upright posture. This update will likely require an update in the lumbar muscle controller, together with an update in the soft tissue modelling in the lower torso area.

Another functionality that is needed and planned for but will not be covered in this Ph.D. is the ability to represent a larger portion of the population. The current active SAFER HBM represents an average male only, while the passive SAFER HBM includes morphing capabilities to represent occupants of different sex, height, age, and BMI [72].

In Paper A and Paper B, the muscles were controlled by PD controllers. In many other active finite element HBMs used for vehicle safety, PD or PID controllers are used to control the muscles. One exception [73] uses a custom control approach. However, it still takes into account proportional and derivative changes of the muscle length. The PD/PID controllers are simple types of controllers and can be implemented directly in LS-DYNA [74]. For other applications where muscle activity is of importance, other types of controllers are often used, such as forward or inverse dynamics, or machine learning [75-78]. If the limits of what can be done with PD/PID postural control are reached, coupling LS-DYNA to a more advanced muscle controller may be an alternative which has been proven successful for leg muscle controllers [79].

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