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# Muscle Responses in Dynamic Events

Volunteer experiments and numerical modelling for the advancement of human body models for vehicle safety assessment

by

JÓNA MARÍN ÓLAFSDÓTTIR

Department of Mechanics and Maritime Sciences
*Division of Vehicle Safety*
CHALMERS UNIVERSITY OF TECHNOLOGY

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

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Raw EMG signal from a sternohyoid muscle during a left frontal oblique 0.55 g sled acceleration

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## Abstract

Fatalities and injuries to car occupants in motor vehicle crashes continue to be a serious global socio-economic issue. Advanced safety systems that provide improved occupant protection and crash mitigation have the potential to reduce this burden. For the development and virtual assessment of these systems, numerical human body models (HBMs) that predict occupant responses have been developed. Currently, there is a need for increasing the level of biofidelity in these models to facilitate simulation of occupant responses influenced by muscle contraction, such as often experienced during pre-crash vehicle manoeuvres.

The aim of this thesis was to provide data and modelling approaches for the advancement of HBMs capable of simulating occupant responses in a wide range of pre-crash scenarios. Volunteer experiments were conducted to study driver and passenger responses during emergency braking with a standard seatbelt and with a seatbelt equipped with a reversible pre-tensioner. Muscle activity, kinematic, and boundary condition data were collected. The data showed that pre-tensioning the seatbelt prior to braking influenced the muscular and kinematic responses of occupants. Drivers modified their responses during voluntary braking, resulting in different kinematics than were observed during autonomous braking. Passenger and driver responses also differed during autonomous braking. The findings demonstrate that HBMs need to account for the differences in postural responses between occupant roles as well as the adjustments made by drivers during voluntary braking. The studies provide detailed data sets that can be used for model tuning and validation.

The modelling efforts of this work focused on simulation of head-neck responses. To facilitate the modelling of neck muscle recruitment, muscle activity data from volunteers exposed to multi-directional horizontal seated perturbations were analysed. The derived spatial tuning curves revealed muscle- and direction-specific recruitment patterns. The experimental tuning curves can be used as input to models or to verify spatial tuning of muscle recruitment in HBMs.

A method for simulating muscle recruitment of individual neck muscles was developed. The approach included a combination of head kinematics and muscle length feedback to generate muscle specific activation levels. The experimental tuning curves were used to define appropriate sets of muscle activation in response to head kinematics feedback. The predicted spatial tuning using the two feedback loops was verified in multi-directional horizontal gravity simulations. The results showed that muscle activation generated by individual or combined feedback loops influenced the predicted head and intervertebral kinematics. The developed method has the potential to improve prediction of omnidirectional head and neck responses with HBMs. However, further work is needed to verify these findings.

Overall, this research has increased knowledge about the muscle responses of occupants in dynamic events typical of pre-crash scenarios. The findings highlight important aspects that must be considered to enable active HBMs to capture a wide range of occupant responses. The data presented support the advancement of current and future HBMs, which will contribute to the development of improved safety systems that reduce the number of fatalities and injuries in motor vehicle crashes.

Keywords: Traffic safety, pre-crash dynamics, muscle activity, human body models, occupant kinematics, spatial tuning patterns, electromyography.

*I am the passenger*
*And I ride and I ride*

- Iggy Pop

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As I sit down to write the last piece of this thesis my mind starts to wander, past the many valleys and hills (mountains really) that have pathed this challenging but exciting journey. Many people have accompanied me, both in the depths of the valleys and on the hill tops, making the overall ride a rewarding one.

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Jóna Marín Ólafsdóttir
May 2017
Göteborg

# Nomenclature

| | |
|---|---|
| AEB | Autonomous Emergency Braking |
| ATD | Anthropomorphic Test Device |
| C1−C7 | Cervical vertebrae |
| CNS | Central Nevous System |
| EMG | Electromyogram |
| EU | European Union |
| Euro NCAP | European New Car Assessment Programme |
| FE | Finite Element |
| GHBMC | Global Human Body Models Consortium |
| HBM | Human Body Model |
| HIC | Head Injury Criterion |
| HUMOS2 | Human Models for Safety 2 |
| MB | Multibody |
| MVC | Maximum Voluntary Contraction |
| PMHS | Post Mortem Human Subject |
| PID | Proportional−Integral−Derivative |
| T1 | First thoracic vertebra |
| THUMS | Total Human Model for Safety |
| U.S. | United States |
| WHO | World Health Organization |
| WSU | Wayne State University, USA |

# List of publications

This thesis consists of an extended summary and the following appended papers:

**Paper A** Ólafsdóttir, J. M., Östh, J., Davidsson, J., Brolin, K. (2013). Passenger Kinematics and Muscle Responses in Autonomous Braking Events with Standard and Reversible Pre-tensioned Restraints. In *Ircobi* (pp. 602-617). Gothenburg, Sweden

**Paper B** Östh, J., Ólafsdóttir, J. M., Davidsson, J., Brolin, K. (2013). Driver Kinematic and Muscle Responses in Braking Events with Standard and Reversible Pre-tensioned Restraints: Validation Data for Human Models. *Stapp Car Crash Journal, 57*, 1-41

**Paper C** Ólafsdóttir, J. M., Brolin, K., Blouin, J.-S., & Siegmund, G. P. (2015). Dynamic spatial tuning of cervical muscle reflexes to multidirectional seated perturbations. *Spine, 40*(4), E211–9.

**Paper D** Ólafsdóttir, J. M., Östh, J., Brolin, K. (2017). Modelling reflex recruitment of neck muscles in a finite element human body model for predicting omnidirectional head kinematics. *In preparation for journal submission.*

**Paper E** Ólafsdóttir, J. M., Fice, J. B., Mang, D. W. H., Brolin, K., Davidsson, J., Blouin, J.-S., & Siegmund, G. P. (2016). Neck muscle activation patterns in dynamic conditions. Updated extended abstract originally presented at the AAAM 60th Annual Scientific Conference and published in *Traffic Injury Prevention, 17*(S1), 219-224.

# Contents

# 1 Background

Motor vehicle crashes are one of the leading causes of death and injury. Over 1.2 million fatalities and up to 50 million injuries occur every year in traffic around the world (WHO, 2015). In the European Union, despite decades of improvements in making vehicles safer, infrastructure, and policy, more than 25.600 people were killed and 200.000 were seriously injured in traffic in 2014 (European Commission, 2015), with car occupants accounting for approximately 51% of the fatalities (WHO, 2015). Although occupant protection has improved remarkably over the past 50 years, further enhancements in vehicle safety are needed to reduce occupant fatalities and injuries.

Progress in occupant protection has been accomplished largely through improvements in vehicle crashworthiness and occupant restraint systems. Advancements in the performance of restraint systems have been achieved through increasingly sophisticated technology for seatbelts, airbags, and other components. For example, most modern vehicles have front seatbelts equipped with pre-tensioners that remove slack from the seatbelt in the early stages of a crash, as well as load limiters reducing chest loads in severe crashes. Airbag systems commonly incorporate multi-stage inflation that can be tailored based on information about the crash and the occupant.

These advanced restraint systems have been designed to achieve good performance with crash test dummies, also known as anthropomorphic test devices (ATDs). ATDs are mechanical models of the human designed to assess occupant injury risk in vehicle crashes. ATDs are valuable tools for safety system assessment, but they are highly simplified models of human anatomy and do not accurately represent the broad range of vehicle occupant size and shape. Furthermore, most ATDs are developed to simulate responses in a single loading direction.

As computation power has expanded, numerical simulation of ATD response in crash scenarios has become increasingly used in development of safety systems. Computational models of ATDs have reduced physical testing costs and improved restraint system optimisation, but the results are still limited by the lack of anatomical accuracy in the ATDs. Consequently, computational human body models (HBMs) have been developed to improve the representation of anatomy and response. HBMs span both lumped-parameter, multibody (MB) models and finite element (FE) models that represent detailed internal and external anatomy. Due to their relatively high fidelity, FE HBMs have become increasingly attractive for simulating restraint and vehicle interactions and assessing injury at the tissue level. Using recently developed morphing methods, anthropometric changes can conveniently be introduced in HBMs to represent a large range of the occupant population. HBMs thus provide an enhanced tool for development and virtual assessment of safety systems.

Recently, the focus of occupant restraint system development has expanded beyond the crash event to consider the critical moments preceding the crash, which may include abrupt vehicle manoeuvring as drivers attempt to avoid an imminent crash (Ejima et al., 2007; Stockman, 2016; Talmor et al., 2010). These events are expected to increase in prevalence and importance as vehicle manufacturers and regulators focus on the development of safety system solutions that prevent or mitigate crashes. Active safety technologies, that is, automated systems that provide warnings or interventions, can be expected to result in an increasing percentage of crashes that are preceded by an abrupt vehicle manoeuvre. A wide range of active safety systems are currently being deployed, ranging from driver assistance systems (e.g. electronic stability control) to driver warnings (e.g. forward collision warning) to dynamic interventions (e.g. autonomous emergency braking (AEB)). Although only 7% of U.S. 2016 vehicle series and 10 out of 18 series tested by Euro NCAP in 2016 included AEB as standard, the majority of U.S. automakers have committed to making AEB a standard feature by 2022, and EU regulation for 2015 has mandated that AEB is optional in all new passenger vehicles (Cicchino, 2017; Euro NCAP, 2017; European Commission, 2009). The number of AEB equipped vehicles is therefore expected to rise rapidly in the near future. Furthermore, systems that provide autonomous steering interventions have been demonstrated by several vehicle manufacturers.

Experimental studies have shown that occupant posture can change substantially when exposed to pre-crash manoeuvres such as emergency braking (Carlsson & Davidsson, 2011; Huber et al., 2015) or steering (Huber et al., 2013, 2015; Muggenthaler et al., 2005). These changes in occupant postures and the associated musculoskeletal responses may influence occupant responses and the restraint performance if a crash subsequently occurs. Consequently, simulation tools are needed that can accurately predict the dynamic response of occupants resulting from pre-crash vehicle manoeuvring. The currently available ATDs were developed to predict injury and be durable in high-energy crash loading but do not accurately match volunteer kinematics during lower-severity exposures (Beeman et al., 2012). ATDs lack musculoskeletal responses and as such do not simulate movements due to postural responses or voluntary actions. In contrast, HBMs that simulate muscle activation have shown good correlation with volunteer data in replicated pre-crash braking (Meijer et al., 2013; Östh et al., 2015). Advanced HBMs that can simulate postural control and muscle responses in a wide range of pre-crash scenarios and subsequent crash events will be increasingly important for the design and assessment of occupant restraints.

# 2 Muscle Recruitment in Pre-Crash Conditions

From the occupant's perspective, a pre-crash manoeuvre, such as emergency braking, can be considered as an external disturbance that compromises the stability of the body. Multiple physiological processes act to maintain stability during normal tasks associated with everyday life such as walking, throwing a ball, or riding in a car. When an external load is applied to the body these postural control processes are excited, triggering them to initiate correcting adjustments in order to retain stability and regain equilibrium. Feedback originating in various sensory organs is transmitted by afferent nerves and processed by the brain or spinal cord to generate motor commands actuated by skeletal muscles. Depending on the task and external loading condition, the contribution from the various input signals originating from the visual, vestibular and somatosensory systems are modulated by the central nervous system (CNS) (Keshner, 1995; Peterka, 2002; Schouten et al., 2008) and integrated with voluntary commands to produce body movement.

Somatosensory input includes sensory information from receptors in the skin, muscles and joints. Muscle spindles, found within the muscle belly, sense muscle length and lengthening velocity. Through its most direct pathway, i.e., a monosynaptic reflex arc, afferent signals from muscle spindles are transferred to the spinal cord and back to the muscle. This stretch reflex provides rapid posture compensation by acting to counter muscle lengthening that occurs due to external perturbations. Although found in most skeletal muscles in the body, spindles are particularly abundant in neck muscles and, along with the vestibular system, constitute a primary contributor to reflex stabilisation of the head (Keshner, 2009; Richmond & Abraham, 1975).

The other major reflex system controlling posture is driven by the vestibular organ in the inner ear. The three semicircular canals sense head angular acceleration (coding for angular velocity); otoliths detect linear acceleration and the head's orientation relative to gravity (Angelaki & Cullen, 2008; Fitzpatrick & Day, 2004). The CNS responds to these signals by activating muscles that counter the posture deviation produced by the movement of the head in space.

Cutaneous receptors in the skin sensitive to pressure change can also induce muscle contractions. For a vehicle occupant, a change in the force applied to the body by the seat could contribute to initial muscle responses at the onset of a manoeuvre (Forssbergr & Hirschfeld, 1994; Magnusson et al., 1999). The visual system also influences postural control during driving and riding in a vehicle, eliciting head motions to stabilise gaze through the optocollic reflex (Wylie, 2009). The visual system can estimate position and velocity of external objects. A high-intensity visual stimulus during a traffic incident can trigger a protective response characterised by muscle contractions throughout the body, i.e., bracing (Hault-Dubrulle et al., 2011) that results in a general stiffening of joints. Information obtained from the visual system may further contribute to anticipation of accelerations, prompting muscle responses.

In addition to these functional reflexes and volitional responses, unexpected visual, auditory, or tactile stimuli can trigger a startle reflex, characterised by rapid tensing of skeletal muscles throughout the body. Experiments have demonstrated that startle reflex contributes to muscle responses during unexpected rear-end sled accelerations (Blouin et al., 2006; Siegmund, Blouin, & Inglis, 2008). Startle inducing safety systems have been proposed to advance this response so that neck muscle activity and peak head kinematics during a rear end impact may be reduced (Mang et al., 2015, 2012). Tactile stimuli from a pre-tensioned seatbelt may also trigger a startle response (Paper A and B). Although muscle activation data from actual crashes are not available, these findings suggest that startle reflexes due to pre-crash stimuli might be common. In the future, pre-crash safety systems incorporating warnings or automatic vehicle motions may increase the prevalence of occupant startle responses.

Occupant postural responses in pre-crash conditions are thus influenced by muscle reactions that can be triggered and modulated by multiple sensory inputs and voluntary actions. Activated muscles apply forces to skeletal structures to which they connect, resulting in either a net moment around a joint to generate or restrict movement, or increased joint stiffness as antagonistic muscle pairs co-activate. These changes in joint moments and stiffness affect the dynamic response of the occupant during vehicle manoeuvring. Researchers have exposed volunteers to replicated pre-crash accelerations with a preconditioned muscle state, relaxed or tensed, to investigate the influence of voluntary bracing on kinematics (Ejima et al., 2012, 2007, 2008; Rooij et al., 2013). In these studies, maintaining a tensed (or braced) muscle state before and during loading resulted in less head and upper body displacements for both passenger and driver posture in frontal and lateral exposures. Although these experiments were conducted in laboratory environments that lacked some characteristics of actual pre-crash scenarios, they demonstrate that a preconditioned muscle state, such as co-contraction from bracing, influences the kinematic response of occupants to dynamic events.

Sitting restrained in a car seat is a relatively stable posture. The majority of the body is supported by the seat and seatback, as well as the footwell and seatbelt. The head, however, is usually unsupported, has more freedom to move, and is more exposed to induced motions from vehicle manoeuvring. The recruitment of over 20 muscle pairs in the neck is mediated by the CNS to generate appropriate multidirectional forces to oppose the induced head motion. Due to the multiple muscles that are effective in more than one degree of freedom (no fixed agonist-antagonist relationships), individual neurons innervate multiple neck muscles to generate functional muscle synergies (Sugiuchi et al., 2004). These synchronised patterns of activity have been reported for several neck muscles in isometric conditions (Blouin et al., 2007; Gabriel et al., 2004; Keshner et al., 1989; Siegmund et al., 2007; Vasavada et al., 2002). They revealed that neck muscles have variable preferred directions of activation and spread in activity during voluntary exertion in the horizontal plane. Furthermore, the level of contraction varies across neck muscles during induced head extension from seated perturbations (Siegmund et al., 2007). Consequently, simulation of neck muscle activation in pre-crash manoeuvres require a quantitative understanding of activation patterns at the individual muscle level. However, little is known about how individual muscles are activated in dynamic conditions.

# 3 Measuring and Modelling Muscle Activity

Skeletal muscles consist of bundles of muscle fibres, elongated muscle cells that contain sacromeres, the contractile elements of the muscle (Figure 3.1). Muscles are innervated by motor neurons that transfer nerve impulses from the CNS to each muscle fibre through neuromuscular junctions. When a nerve impulse reaches the neuromuscular junction, an electrochemical event is initiated that ultimately leads to a local change in voltage in the membrane surrounding the muscle fibre. This end plate potential initiates an action potential (flux of ions through the membrane) that propagates from the neuromuscular junction across the whole muscle fibre. The propagation of action potentials lead to a release of calcium ions into the cytosol of the muscle fibre. When the calcium ions intervene with regulatory proteins in the sarcomeres, the contractile proteins within the sarcomere, myosin and actin, bind to form cross bridges and contraction begins. The level of contraction in the muscle can then be increased by increasing the firing rate (rate of action potential transmission) of active motor units or by recruiting additional motor units.

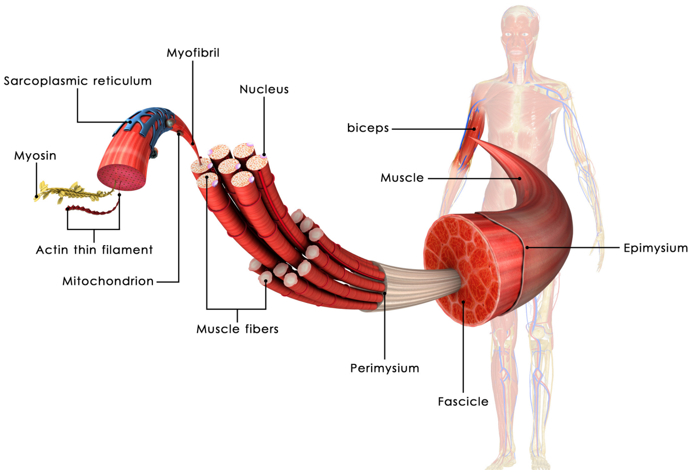


Figure 3.1: *Skeletal muscle structure*

The action potential propagation in the muscle fibres can be measured using electrodes placed either on the surface of the skin or inside the muscle. The measured signal, the electromyogram (EMG), is composed of the superposition of all action potentials detectable by the electrodes recording the signal (Winter, 2009). The EMG therefore represents the electrical activity in the muscle and is measured in volts. Electromyography does not measure the generated muscle force directly but rather the state of activation of the contractile elements surrounding the electrodes

(Winter, 2009).

Because EMG signals can vary widely across individuals and muscles, even with comparable muscle forces, the EMG signal is usually normalised. Normalisation diminishes the effect of extrinsic and intrinsic factors that affect the EMG signal. For instance, the distance from the contracting muscle fibres, and the conductivity of the tissues between them and the electrodes varies widely, and can make subject comparison of EMG levels problematic. Although, these issues are reduced for indwelling electrodes, normalisation is still needed to facilitate intrasubject comparisons and can help to compensate for different levels of strength across individuals when that is desired.

The range of normalisation methods currently used indicate that different methods may be applicable depending on application (Burden, 2010), and that the selection of a method used in related studies can facilitate comparison. In vehicle occupant studies two methods have been used predominantly: normalisation to EMG levels from maximum voluntary contractions (MVCs) under isometric and posture specific conditions and normalisation to maximum EMG levels recorded during the experimental task. The latter method is sometimes preferred as the MVC method requires additional resources and can be time consuming and fatiguing. Although normalising to the maximum EMG observed in experimental trials can result in less intersubject variability (Chapman et al., 2010) comparisons of muscle activation levels between muscles, tasks or individuals, are more meaningful with the MVC method.

Insight into generated muscle forces is often of interest in biomechanical models. Since direct muscle force measurements is a complex and invasive procedure, muscle forces are often estimated from EMG measurements. However, the relationships between EMG and force are not straightforward. Both linear and nonlinear relationships between surface EMG and force have been found (Staudenmann et al., 2010), reflecting the complexities of both the measurement methodology and the underlying physiology. Nonlinear relationships could arise if the contribution of synergistic muscles in load sharing vary unequally with contraction levels (Staudenmann et al., 2010), or if the arrangement of motor units within a muscle places smaller motor units deeper in the muscle and consequently farther from the electrodes. Because smaller motor units will generally be recruited first (Marieb & Hoehn, 2010), the slope of the relationship between force and EMG could vary with force level (de Luca, 1997; Roeleveld et al., 1997; Staudenmann et al., 2010), However, depending on the muscle (de Luca, 1997) and the application, linear approximation of the force – EMG relationship can give reasonable predictions (Staudenmann et al., 2010)

The Hill muscle model (or Hill-type model) is widely used to represent muscle contraction dynamics in computational biomechanical models. Hill-type models (see for instance (Winters, 1995; Zajac, 1989) describe the resultant force generated by the contracting muscle (contractile element) and its passive stiffness. The contractile properties are modelled based on the relationship between force, muscle length and shortening velocity (Figure 3.2). A muscle generates its maximum active force at its optimal length, $l_{opt}$, the length at which most binding sites for myosin and actin are available. At longer and shorter lengths the force generating capacity of the sarcomeres decreases as a result of fewer available binding sites. When a muscle shortens (concentric contraction) with increased speed the muscle force decreases as tension cannot be sustained due to rate limiting factors at the binding sites. The force decreases until the maximum shortening velocity, $V_{max}$, is reached. In contrast, during lengthening (eccentric contraction) muscle force increases above its peak isometric force as a greater force is needed to break the bond at the binding sites than is generated during the isometric contraction. In the Hill model, the net effect of these force-length-velocity dependencies is then scaled with the current activation level bounded between 0 and 1.

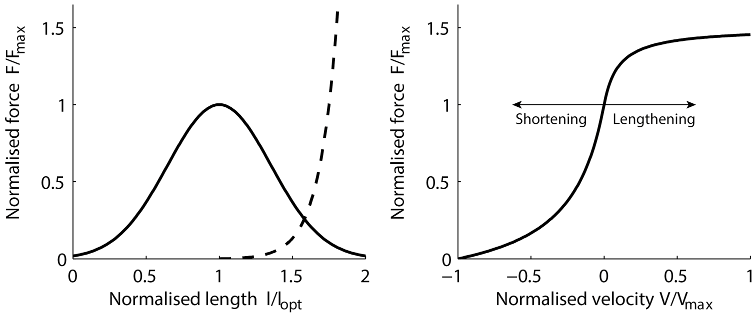


Figure 3.2: *Force, length and velocity relationships in a Hill-type muscle model. Left: Normalised active (solid line) and passive (dotted line) force as a function of normalised muscle length. Right: Normalised force as a function of normalised shortening/lengthening velocity.*

# 4 Human Body Models for Safety

Computational models of the human body have been used to study the occupant response in crashes since the 1960s (Yang, 2015). Since then, human body models (HBMs) have become widely used in safety research within the automotive industry and academia. Following the immense increases in computing power in recent decades, HBMs have grown tremendously in detail and functionality, with current models incorporating element based representation of different body parts ranging from skeletal structures to individual ligaments.

Two main approaches have been applied to model the human for occupant simulations, multibody (MB) and finite element (FE) methods. With MB, the model is constructed from multiple rigid and flexible bodies of simplified geometry, interconnected with spring-damper formulations to represent joints. The MB analysis consists of solving the equations of motions for the MB system. The biggest advantage of the MB approach is that the computational run times are short. However, as material formulations are not included, injury prediction capabilities are limited and constrained to injury criteria at a global level. In contrast, the FE approach, divides each entity to be simulated into a set of small, interconnected elements allowing highly detailed representation of complex geometry. Moreover, each entity is assigned specific material properties through constitutive laws, enabling simulation of local deformation as the equations of motion are solved for each element. FE models therefore have the potential to represent various body entities in fine detail and predict injuries at the tissue level. The compromise of having high level of detail in FE models is the substantially increased computational time.

Numerous HBMs have been developed in the past, both as component models of isolated body parts and as whole body models (for a comprehensive review see Yang et al. (2006) and Yang (2015)). Recent whole body HBMs include the MB MADYMO Human Model (Happee & Ridella, 2000), and a variety of FE models, including the HUMOS2 (Vezin & Verriest, 2005), the Ford Human Body Model (Ruan et al., 2003), the WSU Human Model (Shah et al., 2001), the GHBMC series of models (Gayzik et al., 2011), and the THUMS family of models (Iwamoto et al., 2002; Iwamoto & Nakahira, 2015). Simulations with these FE models can be performed with one or more of the commercial FE solvers, LS-DYNA (LSTC, Livermore, CA, USA), PAM-CRASH (ESI Group, Paris, France), and Radioss (Altair, Troy, MI, USA). The MADYMO Human Model is implemented in MADYMO, a combined MB-FE solver (Tass International, Helmond, The Netherlands). Among the FE models the THUMS and GHBMC series are the most widely used HBMs.[1] Both models are commercially available in sizes identified as $50^{th}$ percentile male, $95^{th}$ percentile male, and $5^{th}$ percentile female.

[1] In a survey conducted by the PIPER consortium in 2014, 156 industrial and academic users of FE HBMs answered that 43% used THUMS v3.0 or v4.0 and 24% used the GHBMC model. 18% answered “other model” and 15% did not use FE HBM (PIPER, 2017).

All of these models were originally developed to study occupant responses during impact and did not include active muscles initially. The development of modern safety systems that activate before an impact, such as electrical seatbelts and autonomous braking/steering, has increased the demand on the biofidelity level of HBMs as this process requires prediction of occupant pre-crash kinematics. Pre-crash events, which commonly are of longer duration than crash events, can trigger a response in the occupant that is influenced by muscle tension. To capture occupant pre-crash kinematics, HBMs need to account for the influence of muscle contraction.

## 4.1 Modelling of Muscle Recruitment in Active HBMs

Simulating muscle contractions of occupants with active HBMs requires a definition of the muscle recruitment, that is, which muscle elements are activated and their level of activation over time. In early models, muscle recruitment patterns were designed as pre-defined muscle activation curves, prescribed to each muscle group based on its presumed function and the direction of loading (Brolin et al., 2005, 2008; Stemper et al., 2006; van der Horst, 2002; Wittek et al., 2000). Pre-defined activation curves guided by or directly constructed from experimental EMG data have also been applied (Choi et al., 2005; Cronin, 2014; Fice & Cronin, 2012; Fice et al., 2011). Optimised activation curves have been proposed either as constant level of activation that represents a relaxed or tensed state prior to loading or as time-varying activation optimised to match the models kinematic response to a volunteer response during loading (Bose et al., 2010; Chancey et al., 2003; Dibb et al., 2013). A more extensive optimisation approach was presented in Iwamoto et al. (2012). In a pre-simulation, reinforcement learning techniques were used to generate surfaces of activation curves as functions of joint angle and angular velocities. During a subsequent simulation, activation was then applied as prescribed by the activation surface maps, based on the current joint kinematics.

These various approaches represent a set of open loop strategies in which the activation curves for different muscles, or groups of muscles, are generated offline prior to simulation of the intended load case. Inherently, open loop strategies do not consider the current model state or varying conditions during the simulation. While the main contribution of these models has been to demonstrate, in particular for the head and neck, how muscle tension affects kinematics and potential neck injuries in certain load cases, the open loop approach is less appropriate for models intended for simulating a wide range of pre-crash scenarios.

Adaptive, active HBMs are needed for general safety system development and assessment. Occupants are exposed to diverse loading conditions during pre-crash vehicle manoeuvres. The loads vary in direction and magnitude and originate from both vehicle manoeuvring and from restraints or other dynamic systems inside the vehicle. The CNS receives sensory information about any changes in the external environment and the internal state of the body, and consequently triggers and modulates muscle recruitment based on these signals. To capture the eventual responses from such a vast variety of conditions, active HBMs need a closed loop recruitment strategy that adjusts muscle activation levels based on similar inputs during the simulation.

To achieve closed loop control in active HBMs, most previous studies have applied proportional-integral-derivative (PID) controllers to determine activation levels for various muscle groups based on continuously acquired information about the dynamic state of the respective body part. In an MB HBM with Hill-type line muscle elements in the neck region, PID controllers were used to generate muscle activation levels to counteract induced head motion by sensing head rotation in the sagittal and frontal planes (Fraga, 2009). Although this approach to some extent approximates vestibular feedback, the results were limited by the assumption of zero neural delays, no contraction dynamics, and a simple load sharing strategy where each neck muscle acted to counteract only flexion, extension, left or right lateral bending. This model was later extended to further counteract axial head rotation and with a refined load sharing strategy (Nemirovsky & Rooij, 2010). The updated model consisted of three PID controllers, one for each rotational degree of freedom, with feedback based on a head rotation metric (head angle relative to T1). Load sharing among muscles was defined through optimization. A pre-defined co-contraction level was added to the activation level determined by the controller. Although the concept was designed to predict three dimensional head motion it was only tested in sagittal plane perturbations. Meijer et al. (2013, 2012) adopted the same concept and in addition to the head-neck control implemented PID controllers on the shoulder, elbow and hip joints. Individual thoracic and lumbar spine joints were stabilised with torque actuators controlled by PID controllers regulating the applied torque at each joint based on rotational displacement relative to the sacrum. A co-contraction level of 50% was used. This whole body MB active HBM (based on the MADYMO Human Model) was evaluated relative to human subject data in emergency braking events.

PID control strategies with joint rotation feedback have been developed for whole body FE HBMs (various versions of THUMS) with multiple Hill-type line elements (Iwamoto & Nakahira, 2015; Östh et al., 2015, 2012). By regulating muscle activity in the neck, trunk, and arms, and incorporating neural delays and muscle contraction dynamics, Östh et al. (2015, 2012) predicted passenger and driver responses in autonomous emergency braking events with a standard seatbelt and a seatbelt incorporating a reversible pre-tensioner. Employing a single PID controller per body part, a simple load sharing strategy was adopted grouping muscles as either flexors or extensors. Although effective for simulating motions in the sagittal plane, this simplifying assumption becomes less tenable when considering other loading directions. Iwamoto and Nakahira (2015) used a similar approach, excluding neural delay and contraction dynamics, to predict responses in a low-g lateral load case. Although a load sharing strategy determining individual contributions of muscles was applied, the specified constants were not empirically derived. Based on an objective rating method the performance of the model in this load case was deemed "marginal" (score below 0.4 out of 1).

A detailed neural control model for head-neck stabilisation in anterior-posterior perturbations was presented in Happee et al. (2017). This MB head-neck model with Hill-type muscle elements (adopted from Meijer et al. (2013) and updated) included both vestibular and muscle spindle feedback. The vestibular reflex loop consisted of head rotational and translational feedback and specific models of the semicircular canal and otolith organ sensor dynamics. The spindle loop consisted of muscle length and lengthening velocity feedback of individual muscle elements with sensor dynamics represented as constant gains. Load sharing that stabilised individual neck joints in isometric maximum voluntary contraction was determined through optimisation. Although the model application was confined to sagittal plane perturbations, the method could be extended to control three dimensional motions at higher load severities. Feller et al. (2016) applied muscle spindle feedback in the neck muscles of an FE HBM and simulated a supine head fall. No vestibular feedback was included.

In summary, recruitment strategies of varying complexity and relevance for three dimensional motion prediction in pre-crash scenarios have been applied in previous models. Previous open-loop strategies with pre-defined muscle activation curves are limited by the fact that they are specific to a particular load case. A more general approach independent of the load case characteristics, e.g., the loading direction and level of acceleration, is needed so that various scenarios can be simulated, preferably without redefinition of model parameters. Proposed feedback control methods, inspired by the closed loop attributes of the CNS, present a promising basis for the development of general purpose FE HBMs for pre-crash simulations. However, no omnidirectional FE HBM is currently available that combines vestibular and muscle spindle feedback to regulate muscle activity.

## 4.2 Experimental Data for Active HBMs

The conventional way of validating HBMs for crash simulations is to compare the model responses to experimental data from impact testing with post mortem human subjects (PMHSs), both on component and whole body level. This comparison is essential to validate passive responses of the model and evaluate its ability to predict injuries or injury risk during impact. In contrast, validation of active responses of HBMs requires data from living subjects obtained by measuring responses of volunteers while exposing them to non-injurious, low-speed impacts or replicated pre-crash events in a controlled environment. EMG data from such experiments can provide an approximation of the muscle recruitment schemes adopted by occupants in actual pre-crash scenarios, supporting the development and validation of strategies to simulate muscle activity in active HBMs. To ensure that HBMs generate relevant kinematic predictions, volunteer experiments must include kinematic and boundary condition measurements (for instance seatbelt forces). Initial volunteer postures and geometry of the experimental setup should be quantified for positioning of HBMs.

Low-speed impact studies have generally been performed in a lab environment using a sled system to perturb the volunteer. Most of the studies that have included EMG measurements have focused on neck muscle activity in rear-end impacts (Blouin et al., 2003, 2006, 2007; Brault et al., 2000; Kumar et al., 2000; Magnusson et al., 1999; Mang et al., 2015, 2012; Siegmund et al., 2007; Siegmund, Blouin, Carpenter, et al., 2008; Siegmund et al., 2002, 2003a, 2003b; Szabo & Welcher, 1996). Beeman et al. (2011, 2012) and Arbogast et al. (2009, 2012) performed frontal and lateral/oblique (Arbogast et al., 2012) sled tests, reporting kinematic trajectories with an impact-like pulse (2.5 – 5.0 g), but did not report EMG.

Comprehensive data sets, suitable for validation of active HBMs, are limited because fewer studies have focused on exposures similar to pre-crash events (<1.5 g and long duration). Kinematics, EMG, and boundary conditions have been measured in low-g frontal and lateral conditions using a sled setup (Ejima et al., 2012, 2007, 2008) and a custom made experimental vehicle (Rooij et al., 2013). These studies presented extensive data sets on volunteer responses in a pre-crash lab environment. However, EMG data were not normalised to MVC and the volunteers were instructed to either relax or tense their muscles before and during the event.

Volunteer studies performed with passenger vehicles in replicated pre-crash conditions are scarce. Few studies have subjected volunteers to emergency braking (Carlsson & Davidsson, 2011; Huber et al., 2015), lane change and combined steering and braking manoeuvres (Huber et al., 2013, 2015; Muggenthaler et al., 2005), presenting valuable kinematic data suitable for evaluating active HBMs. However, no MVC-normalized EMG data were presented.

Previous volunteer studies have demonstrated that muscle tension affects the dynamic response in low-speed impacts and replicated pre-crash events, establishing the need for including active muscles in HBMs. Detailed kinematic data have been presented in these studies, enabling validation of kinematic predictions with HBMs. However, kinematic comparisons of model responses can only verify the simulated recruitment schemes indirectly. For inclusive validation, comparison of muscle responses and predicted muscle activation levels is needed. For such a comparison, appropriately normalised EMG data are required.

# 5 Aim

The general aim of this thesis is to advance the development of human body models capable of simulating occupant responses in a wide range of pre-crash scenarios. This aim was pursued both through volunteer experiments recording occupant muscle and kinematic responses, and modelling of reflexive muscle recruitment in a FE HBM in multidirectional loading. This aim was achieved through the following objectives:

1. to quantify occupant responses in replicated pre-crash braking scenarios,
2. to quantify the spatial tuning of neck muscle recruitment in dynamic loading generated by seated perturbations, and
3. to develop a method for simulating neck muscle recruitment in finite element human body models for improved head and neck kinematics prediction in omnidirectional loading.

# 6 Research Approach

The current research has been carried out as part of the *Active human body models for virtual occupant response* project (referred to as the SAFER A-HBM project), a research collaboration between Chalmers University of Technology, Volvo Car Corporation, and Autoliv Research in Sweden. The aim of the project is to develop and validate an active HBM, the SAFER A-HBM, to provide an enhanced tool for future development, virtual testing, and verification of safety systems during combined pre- and in-crash events.

## 6.1 Volunteer Experiments

The dynamic behaviour of vehicle occupants in pre-crash conditions can be studied through volunteer experiments by exposing subjects to replicated pre-crash events. A replicated pre-crash event consists of a controlled test environment in which a non-injurious but representative acceleration pulse is applied to seated volunteers. These experiments are generally performed using a sled, or other custom made vehicle, in a laboratory setting or with a passenger vehicle driven on a test track. In the context of active HBM development the primary objectives of pre-crash volunteer experiments are:

a. *To gather data on human posture control strategies specific to pre-crash events to guide the development of methods for simulating recruitment strategies.*

b. *To provide data for tuning model parameters and evaluation of model performance.*

This can be partly achieved by measuring the volunteer kinematic response and interaction forces. However, for the development and validation of simulated muscle recruitment, EMG data are also needed. In this thesis, a combination of passenger vehicle and sled test experiments were used. A vehicle experiment was conducted primarily to provide validation data (objective b.) for pre-crash braking scenarios. Sled experiments were used to quantify the spatial tuning of neck muscles, providing essential data for the development of omnidirectional muscle recruitment strategies (objective a.).

Neck muscle spatial tuning can be studied in isometric conditions (Gabriel et al., 2004; Keshner et al., 1989; Siegmund et al., 2007; Vasavada et al., 2002), which allows for a more controlled protocol that maintains constant head position and exerted head forces through visual force feedback. However, isometric contractions represent voluntary recruitment. Dynamic experiments, such as sled perturbations, are needed to evoke reflexive feedback mechanisms that reflect conditions occupants encounter during sudden vehicle pre-crash manoeuvring.

## 6.2 FE Model With Neural Control

Within the framework of the SAFER A-HBM project the overall goal is to develop an HBM that can simulate combined pre-crash and subsequent crash events in a wide range of scenarios and predict injuries from crash loads based on tissue criteria. To support this goal, and to broaden the impact of the methods developed in this thesis, an FE HBM that is widely used in industry and academia, the commercially available THUMS v3.0 (Toyota Central R&D Labs., Inc., Nagakute, Aichi, Japan), was chosen as the baseline model and LS-DYNA (LSTC, Livermore, CA, USA) as the explicit FE solver. The THUMS model represents the body dimensions of a mid-size adult male and is modelled in a driver posture (Figure 6.1 left). Several model modifications implemented in previous work within the SAFER A-HBM project, incorporated to improve the compliance of the model's passive response in pre-crash loading, were adopted in the current work (see Östh et al. (2012)). Furthermore, the neck muscle elements reported in Östh et al. (2012) were used. Modifications to the skin material properties and several additional muscle elements were included (see Paper D). Figure 6.1 (right) illustrates the neck muscle elements

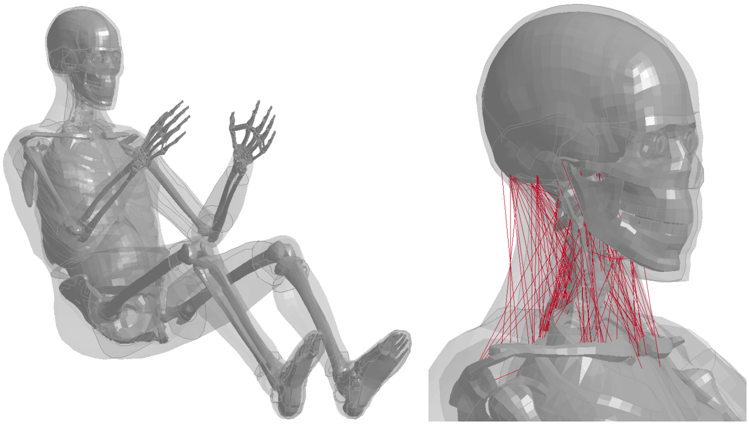

Figure 6.1: *The THUMS v3.0 baseline model (left) with the incorporated neck muscle elements (right).*

Applying feedback control to regulate muscle recruitment in HBMs has been identified as a plausible method for predicting pre-crash kinematics (Iwamoto & Nakahira, 2015; Meijer et al., 2012; Östh et al., 2012). For the advancement of a general purpose HBM with omnidirectional capabilities a method representing contribution from two primary sensory feedback mechanisms involved in human head-neck stabilisation was developed to regulate muscle activity in individual neck muscle elements. The method consisted of muscle length feedback of each muscle element, representing contribution from muscle spindles, and head kinematics feedback, representing vestibular feedback, with a load sharing strategy based on the spatial tuning identified from the sled experiment in Paper D.

The methods developed are not HBM specific and can be adopted by other HBMs alternative to THUMS. However, the implementation is LS-DYNA specific (standard binary, MPP v971 R8.0.0).

# 7 Summary of Papers

## 7.1 Paper A and B

**A: Passenger kinematics and muscle responses in autonomous braking events with standard and reversible pre-tensioned restraints.**

**B: Driver kinematic and muscle responses in braking events with reversibly pre-tensioned and standard restraints – Validation data for active human body models.**

**Aim:** To quantify the muscle and kinematic response of passengers and drivers in various emergency braking scenarios with different restraint configurations. The goal was to generate detailed data sets suited for the development and validation of active HBMs for pre-crash braking simulations.

**Method:** The experiment included 20 female and male volunteers driving in a passenger car on rural roads subjected to five different test cases; autonomous braking as both passenger (Paper A) and driver (Paper B, see Figure 7.1) with standard and reversible pre-tensioned seatbelts and maximum voluntary driver braking with a standard seatbelt (Paper B). With the pre-tensioner a 170 N tension was applied 200 ms before the initiation of the autonomous braking (1.1 g). EMG was measured with surface electrodes placed bilaterally on selected cervical, lumbar and upper extremity muscles. Subjects performed maximum voluntary contractions in isometric conditions for each muscle tested to provide normalisation constants for the EMG signals. EMG was recorded 1.5 s before and during the braking events. Kinematic responses of the head, neck and upper extremities were acquired by film analysis. Interaction forces between the vehicle and occupant were measured, as well as seatbelt positions and seat indentions (driver only).

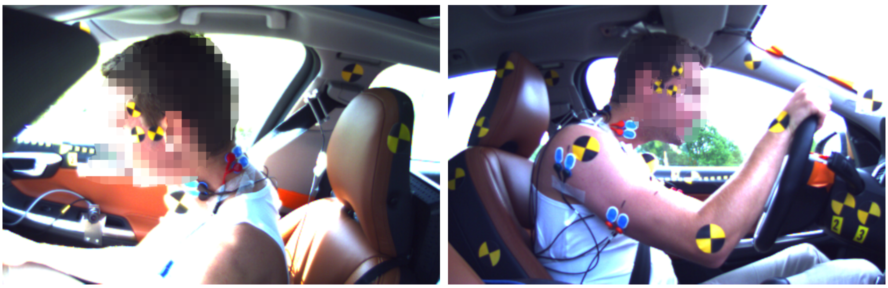

Figure 7.1: *Volunteer subjected to autonomous braking as passenger (left) and driver (right).*

**Results:** The cervical and lumbar extensors displayed the highest contraction levels during steady state braking along with shoulder and elbow extensors for drivers. Females generally displayed higher muscle activity compared to males. Belt pretension affected the muscle and kinematic response of both passengers and drivers. Tensioning the belt invoked a reflexive response, particularly in females, where the muscle activity increased once the retractor began to pull. For the other braking conditions muscle activity increased once acceleration started to ramp up or when the volunteer prepared to press the brake pedal during voluntary braking. Pre-tension resulted in statistically significant reduction of head forward displacement compared to the standard belt configuration for passengers, 81 mm for females and 66 mm for males. For drivers, the reduction was 65 mm for females and 21 mm for males. As a result of being prepared and bracing themselves against the steering wheel, drivers experienced approximately one third of the forward displacement of the head compared to autonomous braking with a standard seatbelt. Female and male kinematic responses were similar, especially for passengers and during voluntary driver braking. However, reduction in head displacement from seatbelt pre-tension was more pronounced for females.

**Conclusion:** The two studies provide a detailed data set for tuning and validation of active HBMs for simulation of emergency braking scenarios. Pre-tensioning the seatbelt prior to braking reduced forward head displacement of drivers and passengers during braking. However, it did induce a muscle reaction in some occupants. Drivers modified their response during voluntary braking compared to autonomous braking which further reduced their forward head displacement.

**Division of work:** (**A**) The experiment was planned by Östh with advice from Davidsson and Brolin. Ólafsdóttir and Östh performed the experiment. Ólafsdóttir analysed the data with support from Östh. The paper was written by Ólafsdóttir, which was reviewed by all authors.
(**B**) The experiment was planned by Östh with advice from Davidsson and Brolin. Ólafsdóttir and Östh performed the experiment. Östh analysed the data with support from Ólafsdóttir. The paper was written by Östh, which was reviewed by all authors.

## 7.2 Paper C

### Dynamic Spatial Tuning of Cervical Muscle Reflexes to Multi-Directional Seated Perturbations

**Aim:** The activation patterns of neck muscles vary with the direction of the intended or induced head motion. The quantification of these patterns is necessary for the advancement of omnidirectional active HBMs. Thus, the aim of this study was to determine the activation patterns and spatial tuning of reflexively activated neck muscles during seated perturbations.

**Method:** The study included eight volunteers subjected to 1.55 g perturbations in a car seat mounted on an experimental sled. The volunteers were unrestrained and the head restraint was removed. The direction of perturbation was varied in intervals of 45°from forward (see Figure 7.2). EMG activity was measured with wire electrodes inserted into the left sternocleidomastoid, trapezius, levator scapulae, splenius capitis, semispinalis capitis, semispinalis cervicis, and cervical multifidus muscles, see Figure 7.2. All wires were inserted at the C4/C5 level with an additional CM insertion at C6/C7. Left sternohyoid activity was measured with surface electrodes. Before the perturbation tests, the volunteers performed isometric maximum voluntary contractions in the eight corresponding directions to provide normalisation constants. The dynamic spatial tuning patterns of each muscle were analysed at 90, 110 and 130 ms after perturbation onset.

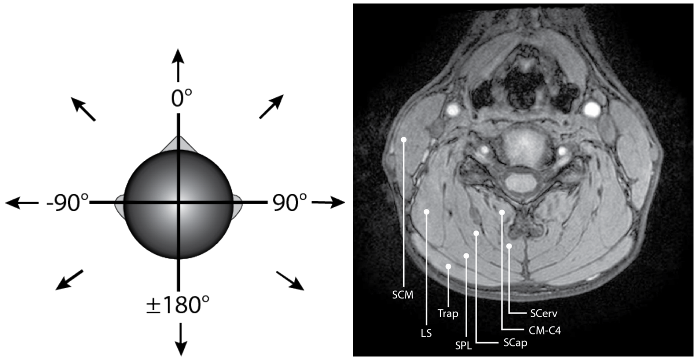


Figure 7.2: *Applied perturbation directions (left) and wire insertion locations (right). SCM: sternocleidomastoid, LS: levator scapulae, SCap: semispinalis capitis, STH: sternohyoid, Trap: trapezius, SCerv: semispinalis cervicis, CM-C4: cervical multifidus C4/C5 level, CM-C6: cervical multifidus C6/C7 level, SPL: splenius capitis.*

**Results:** The spatial tuning patterns at the three time instants were similar and therefore the results presented focused on the 110 ms time point. At this time, only minor head motions had occurred. The anterior muscles were more active during forward and forward oblique perturbations, while most posterior muscles had higher EMG amplitudes in rearward and rearward oblique perturbations. Despite the similarities, the muscles had variable activation amplitudes and preferred direction of activation.

**Conclusions:** The findings indicate that omnidirectional models that include a representation of the neuromuscular control of cervical muscles should implement muscle specific activation schemes that can account for the directional preferences of the neck muscles. The spatial tuning curves derived in this study can be used as input to determine load sharing in omnidirectional neck models or to validate directional tuning of simulated neck muscle recruitment.

**Division of work:** Siegmund and Blouin planned and performed the experiment. Ólafsdóttir made the outline of this study with advice from Brolin and Siegmund. Ólafsdóttir analysed the data and wrote the paper, which was reviewed by all authors.

## 7.3 Paper D

**Modelling reflex recruitment of neck muscles in a finite element human body model for predicting omnidirectional head kinematics**

**Aim:** To develop a neural control scheme for regulating individual neck muscle activity in an FE HBM for simulation of omnidirectional pre-crash manoeuvres. The proposed scheme included a representation of vestibular and muscle spindle feedback, and load sharing based on the spatial tuning curves presented in Paper C.

**Method:** THUMS v3.0 was used as the baseline FE HBM. The neural control scheme (Figure 7.3) used neck link (T1-head vector) rotation and muscle length feedback, representing vestibular and muscle spindle feedback respectively, to determine the activation level in individual muscle elements. To transform a single excitation signal generated by neck link feedback a load sharing strategy was defined using the spatial tuning patterns from Paper C. To explore the capacity of the neural control scheme to generate muscle recruitment patterns that counteract imposed head motion in various directions and to verify the spatial tuning of recruitment, the HBM was exposed to multidirectional horizontal gravity simulations. To study the effect of each feedback loop on the model response, the simulations were run with individual and combined feedback loops.

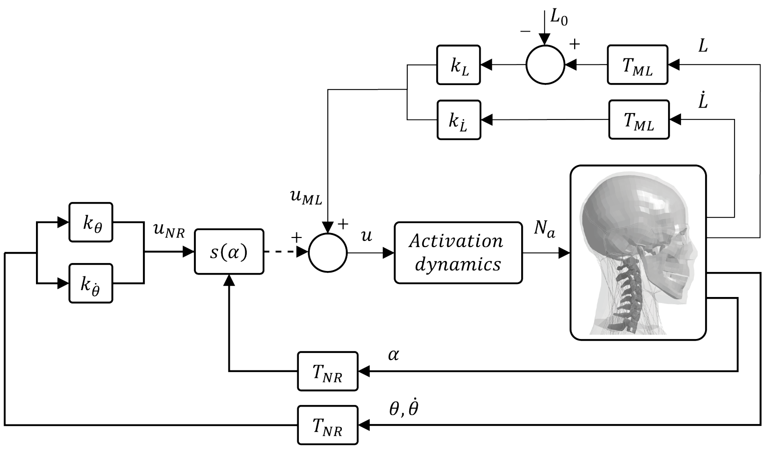


Figure 7.3: *Schematic illustration of the neural control scheme for regulating reflex neck muscle activity based on head kinematics and muscle length feedback.*

**Results:** The muscle activity generated by the neural control scheme effectively reduced head displacement in all directions compared to the passive HBM. The spatial tuning of recruitment with combined feedback loops was similar to experimental data, excluding the sternohyoid and trapezius muscles. Muscle length feedback resulted in substantially higher peak head rotations compared to neck link feedback and combined feedback during induced flexion and right oblique flexion. However, muscle length feedback generally resulted in less rotation of the C7, C4, and C2 vertebra in all loading directions. Integrating both feedback loops therefore had a combined effect with smaller vertebral rotations than neck link feedback alone and smaller head rotation than muscle length feedback in all direction.

**Conclusions:** The results indicate that combining sensory information about head kinematics and individual muscle length to generate activation levels for each muscle influences the predicted head and spinal posture and may improve the prediction of intervertebral kinematics in some loading directions. Further work is needed to verify these findings.

**Division of work:** The concept for the neural control was developed by Ólafsdóttir with the advice of Brolin and Östh. Ólafsdóttir planned the study, implemented the neural control model, and conducted the majority of the simulations. Ólafsdóttir analysed the results and wrote the paper, which was reviewed by all authors.

## 7.4 Paper E

**Neck muscle activation patterns in dynamic conditions**

**Aim:** To assess the applicability of using perturbation data to define load sharing between neck muscles in models that predict pre-crash responses, the objective of this study was to determine neck muscle activation patterns under multi-directional dynamic loads typical of pre-crash events and to compare them to activation patterns from perturbations.

**Method:** To allow a comparison to the perturbation data reported in Paper C the experiment was carried out using similar conditions. The study included four volunteers subjected to an acceleration level of 0.55 g (net $\Delta v$ = 4.0 m/s, $\Delta t$ = 0.7 s) while seated in a car seat mounted on an experimental sled. The head restraint was removed and the volunteers were restrained with a lap belt. The direction of perturbation was varied in interval of 45°from forward, see Figure reff:wire. EMG activity was measured with wire electrodes inserted into the left sternohyoid, sternocleidomastoid, trapezius, levator scapulae, splenius capitis, semispinalis capitis, semispinalis cervicis, and cervical multifidus muscles at the C4/C5 level. Before the perturbation tests, the volunteers performed isometric maximum voluntary contractions in the eight corresponding directions to provide normalisation constants. The dynamic spatial tuning patterns of each muscle were analysed at 110 ms after acceleration onset and compared to the spatial tuning recorded from perturbations ($a_{max}$ = 1.55 g, $\Delta v$ = 0.5 m/s, $\Delta t$ = 0.060 s)

**Results:** The pre-crash like pulse (0.55 g) evoked higher activity compared to the perturbation pulse (1.55 g) for most muscles. The activation levels were, however, often within the interquartile range of the levels from the perturbations. Similar median spatial tuning was induced by both acceleration pulses, except for Trap and SPL.

**Conclusions:** The preliminary findings from this study suggest that spatial tuning of muscle activity in perturbations can be an indicator of the preferred direction of muscles during pre-crash like loading, however this could not be statistically verified due to the limited number of subjects.

**Division of work:** The experiment was planned by Ólafsdóttir with advice from Davidsson and Siegmund. Ólafsdóttir, Fice, and Mang performed the experiment. Ólafsdóttir analysed the data and wrote the paper, which was reviewed by all authors.

# 8 Discussion

## 8.1 Influence of Autonomous and Driver Maximum Braking on Posture: Implications for Models

As described in Paper B, volunteer drivers applying maximum braking tended to adopt an anticipatory response in which they modified their posture to prepare for the applied deceleration, actively compensating for the induced motion due to the inertial loading. This increased muscle contraction and resulted in reduced head excursion relative to autonomous braking. For instance, mean forward head excursion for female drivers with a standard seatbelt during steady state driver braking was 38 mm compared with 116 mm during autonomous braking. With a reversible pre-tensioner tensing the seatbelt prior to autonomous braking, forward excursion was 51 mm. Average tricep and bicep activity increased from approximately 21 to 42 %MVC and 7 to 13 %MVC, respectively. Even though the mean forward head displacement when restrained with the reversible pre-tensioner was less than half of that with a standard belt, the voluntary braking resulted in the least forward displacement. Simulation studies have shown that more forward postures can lead to early contact with the airbag causing increased head injury criteria (HIC) values (Bose et al., 2010) and higher seatbelt loads (Antona et al., 2011) during a frontal crash compared to a nominal posture. However, a rearward posture with the head close to the head restraint can also lead to increased HIC values (Bose et al., 2010). To assess the change in injury outcome with posture, models need to capture the adaptive responses of drivers. This has been successfully done in an active HBM where anticipatory responses were modelled and the driver maximum braking in Paper B was simulated (Östh, Eliasson, Happee, & Brolin, 2014). Including an anticipatory control strategy with a previously developed postural feedback control (Östh et al., 2015, 2012) provided outputs that were closer to the volunteer response than with the postural control alone.

With the same belt configuration as the drivers in Paper B, the head excursion of female passengers (Paper A) was 186 mm with a standard seatbelt and 105 mm with a reversible pre-tensioner. Muscle activity was similar for drivers and passengers except for the upper extremities and lower back muscles. Without the steering wheel to brace against, passengers cannot restrict the induced forward motion to the same extent as drivers, resulting in approximately twice as much head excursion and increased back muscle activity. This indicates that the design requirements for restraint systems are different for drivers and passengers in pre-crash braking situations. By introducing models that have been thoroughly validated against data, differences can be further assessed and the systems optimised for each occupant group.

## 8.2 Startle Response Triggered by Pre-Collision Seatbelt Tensioning

The primary goal of reversible pre-tensioned seatbelts is to remove belt slack so that the occupant is better coupled to the seat, thereby reducing occupant movement during a crash. Reversible pre-tensioners may also be capable of repositioning an occupant to a more centred position if the tension is high enough and applied for a long enough duration. The data presented in Paper A and B indicated that tensioning the seatbelt prior to braking may trigger a startle response. In these experiments, some drivers (trials 2-4, i.e., first exposure excluded) and passengers reacted with an initial burst in muscle activity after the seatbelt was pulled and before any vehicle deceleration occurred. It is likely that the tactile stimulus of the sudden pressure from the seatbelt pre-tensioning triggered mechanoreceptors in the chest or shoulder area which initiated the startle-like burst of muscle activity.

The duration of the increased muscle activity from the response initiated by the belt pre-tensioning was about 200–300 ms. The potential influence of this relatively high increase in muscle activity due to startle on risk of injury during a crash is unknown. A simulation study with a head-neck model where startle-like activations (increased activity for 150 – 250 ms) were applied to the neck muscles resulted in reduced loading of the cervical ligaments for both frontal and lateral impacts (Brolin et al., 2005). On the other hand, startle responses in low-speed rear-end impacts increase cervical multifidus activity; in the presence of increased loading due to abnormal vertebral motions, this increased activity might strain the facet capsule to an injurious level (Siegmund, Blouin, Carpenter, et al., 2008; Winkelstein et al., 2001). In the light of this hypothesis Mang et al. (2015) have suggested a novel pre-crash safety system in which a loud startling tone is presented 250 ms prior to a rear-end impact to advance the startle reflex so that multifidus activity will have faded during the impact and thereby reducing the risk of whiplash injury (Mang et al., 2015, 2012). Active HBMs incorporating detailed neuromuscular control models that can simulate the startle reflex, could shed further light on the influence of muscle activity on cervical injury mechanisms during crashes in different loading directions.

However, more data are needed to determine the type of muscle response that would follow an initial startle reflex in real-life pre-crash events, e.g., voluntary actions such as bracing. Active HBMs simulating the startle responses need to include strategies that allow the transitioning from startle reflex to voluntary response. Furthermore, although a startle response was neither detected in passengers and drivers during autonomous braking with a standard seatbelt nor in drivers during maximum voluntary braking, an actual pre-crash event under these conditions could possibly evoke a startle reflex triggered by visual and auditory stimuli, such as an oncoming vehicle or other suddenly appearing obstacles. To verify startle reflex models, it would be beneficial if future volunteer studies were focused on identifying startle responses due to realistic visual threats and the transition from startle to voluntary bracing in pre-crash events.

## 8.3 Directional Dependency of Muscle Recruitment

Quantification of the spatial tuning of various muscles is fundamental for the development and validation of omnidirectional active HBMs. The spatial tuning of several neck muscles has previously been reported in isometric conditions (Gabriel et al., 2004; Keshner et al., 1989; Siegmund et al., 2007; Vasavada et al., 2002). Data from these studies can be used to verify the directionality of isometric contractions in models and provide an estimate of the dynamic spatial tuning. However, the relative inter-muscle contribution to dynamic stabilisation cannot be determined from isometric data. The information gained from dynamic experiments is thus twofold: the spatial tuning specific to reflexive contractions and the relative differences in activation between different muscles in various directions. The tuning curves presented in Paper C and E can thus be used to verify the directionality of neck muscle recruitment of models in dynamic conditions or provide direct input to models for defining load sharing.

Verifying the directional activity of neck muscles is essential to ensure that the applied recruitment strategies are effective in multiple directions resulting in improved omnidirectional head kinematics predictions. Such comparison is particularly important for detailed neck models intended for injury prediction where accurate prediction of neck loads is needed. Due to the high degree of redundancy of muscles in the neck it is possible that global head kinematics can be accurately predicted with non-representative muscle recruitment patterns. For instance, in a simulation of a 0.6g side sled test an active HBM predicted volunteer head kinematics with a "good" and "fair" CORA[2] score while predicted neck muscle activation was rated "marginal" or "unacceptable" (Iwamoto & Nakahira, 2015). In particular the predicted shape and phase had a poor fit to the experimental EMG data. Therefore the onset and load pattern of muscle tension over time will be inaccurate and skew the estimation of the contribution of muscle force to neck loads. Taking into account experimental data on muscle recruitment contributes to improved predictions of both kinematics and muscle force, ultimately leading to improved prediction of injuries and injury mechanisms.

Muscle activity in various neck muscles was measured with indwelling electrodes in Paper C and E. Although surface electrodes are more commonly used in vehicle occupant studies they can only capture the activity of certain muscles, mainly muscles superficial to the body. Surface electrodes do not accurately isolate the activities of deep layered muscles in the neck. Consequently, little is known about how the

[2] CORA is an objective rating method used to quantify how well simulated human responses fit experimental data (Gehre et al., 2009). On a scale from 0-1 where 1 represents a perfect fit "good" rating ranges from 0.65-0.86 and "unacceptable" from 0.0-0.26.

more than 20 neck muscle pairs are activated to reflexively stabilise the head during impulse dynamic events. In particular, reports of the role of deeper musculature and the spatial tuning of different muscles during externally induced head motion are largely lacking. This gap was addressed in Paper C and E where spatial tuning patterns from eight muscles were presented from two load cases.

Activity was recorded in eight neck muscles in Paper C and E. Although the measured muscles comprise a large proportion of the muscle tissue at the C4-C5 level, the activity of other muscles as well as potential diverse sub-compartments of the measured muscles remains unknown. More studies are needed to identify the dynamic spatial tuning of these muscles and muscles at other spinal levels (lumbar and thoracic). Volunteers were tested in a typical passenger posture with the head in a neutral forward facing position. Different postures such as driving posture or non-neutral head positions may influence the spatial tuning of some neck muscles.

## 8.4 Validation Data for Active HBMs

To facilitate the prediction of injuries or the influence of muscle tension on injury with active HBMs the predicted muscle forces should be realistic. The predicted muscle forces must be validated against experimental data. Since direct muscle force measurements is a complex and invasive procedure, muscle forces are often estimated from EMG measurements. As mentioned in the section Chapter 3 *Measuring and modelling muscle activity*, linear approximations of the force – EMG relationship can often give reasonable predictions of muscle force from EMG measurements (Staudenmann et al., 2010). This is mainly true for isometric contractions. For dynamic contractions force will vary nonlinearly with muscle length and shortening velocity, as well as have length-dependent contributions from the passive properties of the muscle. Therefore, depending on the muscle length and length change, different levels of force will be present for the same neural drive, i.e. the same EMG level. Another issue is that the relative position of the electrode to the detectable and contracting muscle fibres can change during dynamic contractions (de Luca, 1997; Farina, 2006). This makes validation of predicted forces by active HBMs for injury risk estimation problematic. A more feasible validation procedure could be to compare the measured EMG levels to the predicted activation levels by the muscle model and muscle controller in the active HBM as was done in (Iwamoto & Nakahira, 2015; Östh et al., 2014) or comparing the produced moments in the model and joint moments predicted by inverse dynamics (Staudenmann et al., 2010). A limitation of the latter procedure is that only a comparison of the net moment produced and not the contribution of individual muscle forces can be established.

Three types of data are required for thorough validation of active HBMs in pre-crash events: EMG data normalised to MVC, kinematics and boundary conditions. All EMG data presented in the current work was normalised with MVC. As described in Chapter 3 *Measuring and modelling muscle activity*, an alternative method for EMG normalisation is to normalise to the maximum EMG amplitudes recorded during the experimental task. This method is sometimes used to save time or reduce intersubject variability and has been applied in several pre-crash studies (Ejima et al., 2012, 2007, 2008; Rooij et al., 2013). The interpretation of these signals cannot be directly replicated with active HBMs, however, and their applicability for validation is thus limited. MVC normalised EMG signals are more appropriate for active HBM development and validation where the signals are represented as a percent of maximum activation, which can more easily be defined in an active HBM. Nonetheless, a limitation of the MVC method is that it relies on the assumption that a true maximum contraction is achieved during the MVC execution which can be difficult or even impossible to verify.

Kinematic data were recorded in the experiments reported here, although these data have not been presented for the multidirectional perturbations (Paper C and E). Using a passenger vehicle in the experiment in Paper A and B posed some restrictions on what kinematic data could be acquired as some areas were obscured from the camera view (see Figure 7.1). However, the vehicle provided an environment that closer resembled traffic conditions where actual pre-crash events occur. Furthermore, this allowed testing of driver responses during actual maximum voluntary braking. The most common method in vehicle occupant studies is to use an experimental sled in a laboratory environment. The potential benefits of this approach are that more detailed kinematics can be measured and the influence of boundary conditions can be reduced for simplification. The appropriate level of simplification depends on the application. For the purpose of thorough validation a test environment closely resembling the actual environment, such as that in Paper A and B, is preferable if boundary condition measurements are included. For the purpose of model development such as the mapping of reflex activation patterns in Paper C and E this might be less important. In the experiment described in Paper C, the volunteers were seated in a regular car seat without a seatbelt or head restraint. Boundary conditions such as seat and floor plate interactions were not measured as they were deemed insignificant due to the low severity of the pulse. For active HBM development and validation the level of detail in kinematic and boundary condition measurements can thus vary depending on the intended use of the respective data set.

## 8.5 Adapting HBMs for Pre-Crash Simulations

Simulating occupants in pre-crash events involves modelling the response of individual muscles that are recruited to maintain upright posture and adjust for postural disturbances due to external loading. For the development of realistic muscle recruitment strategies, the response of the underlying passive model should not account for the stiffness from muscle tension. Traditionally, HBMs have been developed to predict occupant responses in crash conditions. The validation of these models involves comparison to PMHS data as living humans cannot be subjected to injurious crash loads. PMHSs lack the influence of the effective increase in stiffness introduced by muscle tension. However, some HBMs, while exposed to gravity, are capable of maintaining an upright seated posture in their passive state, seemingly addressing the shortcoming of the reduced stiffness. This effect of muscle force has generally not been explicitly modelled but rather included as increased stiffness in material models of other soft tissues such as the skin and intervertebral discs or through constraints between various parts. Existing HBMs have therefore been modified to produce a more compliant response before adding neural control and/or muscle elements for simulating pre-crash responses (Feller et al., 2016; Östh et al., 2012).

Similar model modifications were necessary during the development of the neural control methods presented in Paper D. The model changes introduced to THUMS v3.0 in Östh et al. (2012), aimed at improving the model response to low-g loads such as gravity or pre-crash loads, were adopted in this work. Also, to further reduce excessive stiffness, materials of soft tissues representing the flesh were updated based on Östh et al. (2017). These changes might impose a certain limitation to the usability of the model as the original model properties were chosen to accommodate high strain rates associated with loads experienced in impacts. It is possible that the model changes are less suitable for simulation of crash loads. Investigating how the incorporated changes influence the model's performance in crash simulations was not within the scope of this thesis. However, future modelling efforts should consider implementing the nonlinear and rate dependent properties of biological materials in more detail so that responses in both pre-crash and crash can be predicted with greater accuracy using the same model.

## 8.6 Strengths and Limitations of the Reflex Recruitment Model

The neural control scheme developed in Paper D implemented individual muscle control to reduce head and neck motion responses to external loading in multiple directions. Muscle recruitment of each muscle was based on sensory information about head kinematics (neck link) and the length of each muscle. A load sharing definition was needed to transform a single control signal generated from head kinematics feedback to a set of activation patterns determining the distribution of muscle activation in various loading directions. The experimental spatial tuning curves developed in Paper C were used for this purpose ($s(\alpha)$ in Figure 7.3). Using this strategy, combined with contribution from muscle length feedback resulted in spatial tuning that closely resembled the experimental patterns for most muscles during multidirectional 1g loading.

The reflexive recruitment model studied in Paper D consisted of the aforementioned neural control scheme and the head-neck complex of the THUMS v3.0 model updated with 188 Hill-type muscle elements. The THUMS model is widely used in industry and academia (PIPER, 2017; Yasuki, 2006) which encouraged its application in the present work. Implementing the methods developed in Paper D with THUMS specifically will allow rapid adoption of the current methods by THUMS users, ultimately leading to faster and wider application of this work. The recruitment model was furthermore implemented using native LS-DYNA keywords exclusively and therefore requires no additional compilers (for user defined routines) or coupling to external software. The neural control scheme can thus be easily adopted by other head-neck models implemented in LS-DYNA.

The neural control scheme was constructed on the basis of the major reflex loops involved in head stabilisation, the vestibular (vestibulocollic) and muscle spindle (cervicocollic) reflexes. Although the control scheme does not attempt to model the dynamics of these reflexes they nevertheless motivated the particular approach of the two included feedback loops, i.e., neck link rotation and muscle length feedback and the associated gains. Vestibular feedback was simplified by using absolute rotational displacement and velocity feedback of the neck link. This generated a control signal that was scaled according to the experimental spatial tuning curves based on the position of head relative to T1 in the horizontal plane before it was transferred as activation level to each muscle. No axial head rotation component was incorporated. Axial head rotation was therefore only compensated through the muscle spindle loop. Although it has been suggested that stabilisation of head axial rotation during torso perturbation is dominated by vestibular reflexes (Peng et al., 1999) the necessary EMG data from induced head rotation experiments could not be identified in the literature to determine the relevant load sharing for this degree of freedom.

All muscle activation levels are generated based on model states and experimental data on mean population responses. As such, the neural control scheme is not model specific and may be readily adapted to other FE HBMs. Tuning of feedback gains might be required as the dynamics of the closed loop system are largely dominated by the head-neck inertial and other intrinsic properties, which are generally model specific. The experimental data used was based on a sample size of only eight subjects. Although the subject selection was not targeted towards a specific anthropometry, gender, or age, the small sample size inherently resulted in a small spread in characteristics related to these attributes. A larger, more diverse sample of subjects would allow assessment of the extent to which subject characteristics are related to activation patterns.

Although the neural control scheme does not include any model specific assumptions or inputs in terms of implementation, it models a specific reflex response type. For instance, neck reflexes have been shown to modulate with the mental set (Keshner, 2000). In Paper A and B female subjects displayed more prominent startle reflex than males to the reversible retractor. The neural control scheme does not consider these aspects. However, it is possible that such responses could be captured by varying the feedback gains or adding a feedforward term. Elderly subjects have been found to adopt co-contraction strategies to a greater extent than younger age groups during head-neck stabilisation (Keshner, 2000). To account for increased co-contraction a feedforward term could be added or modification made to the experimental spatial tuning curves defining load sharing ($s(\alpha)$).

The spatial tuning curves used to scale the central signal generated by the vestibular loop were collected during perturbations that were of higher peak acceleration and shorter duration than a typical pre-crash event. Furthermore, the curves were extracted from a single time point during the event (20 ms RMS EMG extracted at 110 ms after acceleration onset). The load sharing definition in the neural control scheme therefore adopts two major assumptions. First, it assumes that the spatial tuning is independent of the loading characteristics and that spatial tuning during perturbation predicts spatial tuning during pre-crash loading. To assess the validity of this assumption, in Paper E, the experimental protocol from Paper C was replicated with a pre-crash like pulse and the spatial tuning from the two load cases compared. The salient similarities in the results, while preliminary, suggest that this assumption was reasonable. Second, the load sharing is assumed to be constant over time. The spatial tuning curves at the 90, 110, and 130 ms time points were similar (Paper C). However, it is likely that later in the event, when voluntary responses become more pronounced, and in particular for sustained loading, the spatial tuning will vary. If the temporal variation in spatial tuning could be estimated, its inclusion in the neural control scheme might be advantageous. In Paper E, EMG data were collected during 700 ms of constant acceleration. These data have unique potential for examination of how spatial tuning varies over time during extended loading. Further study is needed to investigate this aspect of the data.

The study in Paper D considered the head and neck only. However, a similar approach may be applied to trunk muscle recruitment, using trunk motion feedback and muscle length feedback. Although trunk motion feedback does not have a direct neurophysiological parallel this method has been shown effective in models simulating pre-crash braking events (Östh et al., 2015, 2012). As for the neck, this strategy would require determination of trunk muscle spatial tuning in volunteers. Spatial tuning of several trunk muscles has been reported for unsupported seated perturbations (Preuss & Fung, 2008), although preferred directions were not determined in this study. Nevertheless, some muscles seemingly appeared to be directionally tuned. Testing subjects on a car seat may have revealed a different result, as the support from the seatback and side bolsters might affect the directional tuning of trunk muscle recruitment. Further research should be undertaken to investigate the trunk muscle recruitment patterns while seated supported in replicated pre-crash events.

The model with verified spatial tuning of muscle recruitment showed promising results. In particular, the neural control scheme with combined neck link and muscle length feedback was capable of a range of plausible kinematic responses, demonstrating the potential for tuning to match empirical results. However, detailed validation of kinematics using realistic boundary conditions typical of pre-crash exposures is needed.

## 8.7 Population Heterogeneity

The studies contained in this thesis considered a limited range of subject characteristics. The experiments reported in Paper A, B, C and E comprised volunteers close to the midsize anthropometry of males and females and the HBM used in Paper D represented a midsize male. Certainly, male and female occupants with a wide range of body size and age are exposed to motor vehicle crashes. Numerous studies have shown that certain factors related to these characteristics are associated with increased risk of injury (see for instance Bose et al. (2011); Jakobsson et al. (2004); Kent et al. (2009); Newgard et al. (2008); Ridella et al. (2012); Rupp et al. (2013) and a comprehensive review in Hu et al. (2012)). Consequently, HBMs and other assessment tools must represent the variability in occupant characteristics so that safety systems can be designed to protect a larger percentage of the population. Recent advancements in parametric modelling and morphing techniques for rapid development of subject-specific whole body HBMs provide a promising basis for addressing this issue by enabling large-scale population based simulations (Hu et al., 2012; Hwang, Hallman, et al., 2016; Hwang, Hu, et al., 2016). Although the methods developed in this thesis to model occupant muscle recruitment was applied to an HBM of a specific size, the general framework is applicable to male and female models representing other human sizes and ages. Although the geometry of the muscle elements would need to be parametrised the neural control scheme is independent of body size.

One set of feedback gains and scaling factors was used to determine the level of neural excitation and the relative contribution of various muscles in terms of directional tuning and co-contraction (i.e., single load sharing strategy). However, the spread in EMG corridors in Paper A, B, and C shows that muscle recruitment varies among individuals. In general, it can be expected that the variance in reflex muscle recruitment would to some extent be influenced by body characteristics such as sex and age but also cognitive attributes like learning, awareness level, and secondary task engagement (Kumar et al., 2000; Siegmund et al., 2003a; Stenlund et al., 2015). Behavioural aspects might further influence muscle recruitment, in particular when considering voluntary motor commands. To model observed variability in muscle recruitment, stochastic modelling methods similar to those presented by Martelli et al. (2015) could be applied to generate a spectrum of plausible scaling factors, informed by available EMG data, in combination to varying feedback gains and neural delays.

Future studies could include a larger number of volunteers sampled to span a wide range of age and body size. A diverse population will facilitate the examination of potential associations between occupant attributes and responses. Both the kinematics and patterns of muscle activation are of interest. Although the studies in this thesis have focused on braking, data for a wider range of pre-crash manoeuvres are also needed, to determine if occupant characteristics influence the responses differently for different pre-crash modes.

## 8.8 Implications for Safety

The research presented in this thesis provides a rich set of volunteer data from replicated pre-crash braking and multi-directional dynamic loads typical of pre-crash events. The data demonstrate that responses differ for drivers who are actively braking compared to those experiencing autonomous braking. Responses for passengers are also different from those of drivers. A reversible pre-tensioning seatbelt is capable of reducing head excursions during emergency braking. The presented data sets are suitable for tuning and validating active HBMs that are capable of simulating pre-crash responses.

A modelling approach for regulating individual muscle activity showed promise for simulating the patterns of activation of neck muscles under inertial loading conditions and can lead to improved predictions of omnidirectional head and neck kinematics in active HBMs. More work is needed to integrate these findings into tuned and validated HBMs that can be used for optimising restraint systems for improved safety.

# 9 Conclusions

The overall goal of this thesis was to advance the development of human body models that simulate occupant pre-crash responses. Three objectives were defined, spanning volunteer experiments and modelling as presented in Paper A-E and summarised above. Based on the findings reported in Paper A-E, the conclusions related to the objectives are:

1. **Quantifying occupant responses in replicated pre-crash braking scenarios.**

   Detailed data sets were presented on passenger and driver responses during autonomous braking, both with a standard seatbelt and with a seatbelt equipped with a reversible pre-tensioner. Responses during driver voluntary braking were also presented. The data, which included EMG, kinematic, and boundary condition information, are useful for the development of active HBMs and can be used for model tuning or validation.

   The results from these studies provide further insights into the differences in postural responses between passengers and drivers and how drivers modify their responses when initiating emergency braking voluntarily. Pre-tensioning the seatbelt prior to braking can reduce occupant forward displacement during braking. However, this may evoke a muscular response characteristic of a startle reflex. Active HBMs that predict occupant responses must incorporate strategies that can account for different occupant roles (passenger vs. driver) and differentiate between driver initiated braking and autonomous braking. To study the influence of the startle reflex on pre-crash kinematics and injury risk in crashes, active HBMs that model the startle response are required.

2. **Quantifying the spatial tuning of neck muscle recruitment in dynamic loading generated by seated perturbations.**

   Spatial tuning of reflexively recruited superficial and deep neck muscles was presented for seated perturbations with two types of acceleration pulses. The muscles exhibited contraction levels that varied in amplitude and directional preference, highlighting the importance of implementing muscle and direction specific recruitment strategies in neck models. Spatial tuning was similar for both acceleration pulses. However, more subjects are needed to statistically assess this finding. The spatial tuning curves presented in the current work can be used to define load sharing or verify spatial tuning of neck muscle recruitment in models intended to predict omnidirectional head and neck kinematics and to assess the influence of muscle force on injury.

3. **Developing a method for simulating neck muscle recruitment in finite element human body models for improved head and neck kinematics prediction in omnidirectional loading.**

   A method for recruiting individual neck muscles using feedback control was developed and implemented in a widely used FE HBM. Contribution from two reflex pathways representing vestibular and muscle spindles were included by using neck link (T1-head vector) angular deviation and muscle length feedback. The experimental spatial tuning curves from Paper C were used to determine muscle load sharing in response to a central vestibular excitation signal. The predicted spatial tuning using the two feedback loops was verified in multi-directional horizontal gravity simulations. The system was stable and generated plausible head kinematics in all directions.

   This work demonstrates that combining sensory information about head motion and individual muscle length to generate activation levels for each muscle influences spinal alignment and may improve the prediction of intervertebral kinematics in some loading directions. Further work is required to verify these findings.

   The preliminary findings from Paper E suggest that using data from perturbations to determine neck muscle load sharing in models for pre-crash simulations is justifiable. The method presented here has the potential to improve the omnidirectional head-neck response of active HBMs.

Overall, this research has increased the knowledge about occupant muscle responses in dynamic events. The findings highlight important aspects that must be considered to enable active HBMs to capture a wide range of occupant responses. The data presented supports the advancement of current and future HBMs, which in the long term will contribute to the development of improved safety systems that will reduce the number of fatalities and injuries in motor vehicle crashes.

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