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Volunteer Muscle Activity in Dynamic Events

Input Data for Human Body Models

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Cover:
Raw EMG signal from a sternocleidomastoid muscle during autonomous braking (Paper A)

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Abstract

Human body models (HBMs) are virtual human surrogates used to predict kinematic and injury responses during motor vehicle crashes. In recent years, active musculature has been incorporated into HBMs for enhanced biofidelity in simulated emergency scenarios, in particular low-severity crashes and pre-crash situations, where occupant responses are influenced by muscle tension. Development and validation of HBMs that simulate neuromuscular control requires information on muscle activation patterns and contraction levels for different loading levels and directions. This information can be acquired by measuring muscle activity in volunteers with electromyography in replicated pre-crash events. This thesis investigates occupant responses in various pre-crash type braking scenarios and multidirectional perturbations.

Muscle activity was measured in volunteers in the following scenarios; maximum voluntary braking, autonomous braking with standard seatbelt, autonomous braking with reversible pre-tensioner activated 200 ms before braking, and seated perturbation in multiple directions without restraint. Muscle activity and forward displacement during autonomous braking was influenced by type of restraint system and role (passenger vs. driver). Pre-tensioning the seatbelt caused decreased forward displacement as well as increased startle like muscle activity in some volunteers. Active HBMs that model the startle reflex can elucidate its effect on injury risk in the crash phase. The difference in posture between drivers and passengers resulted in decreased upper extremity and increased lower back muscle activity for passengers and more forward displacement. Active HBMs validated against the data presented here can be used to further assess the difference between the two occupant roles and to aid the optimisation of safety systems for each group. The spatial tuning patterns generated from multidirectional perturbation showed variable activation amplitudes and preferred directions for the neck muscles. Implementing muscle and direction specific activation schemes in active HBMs might result in better prediction of the head and neck responses.

The research outcomes provide data sets for active HBM validation in pre-crash braking events and the development and validation of omnidirectional models. Further studies that identify occupant muscle responses are needed. Measuring muscle activity during a pre-crash steering manoeuvre or during a realistic visual threat to identify the muscle responses following a startle reflex would support the advancement of future omnidirectional models and startle reflex control methods.

Keywords: EMG, occupant kinematics, reflex, seatbelt pre-tension, emergency braking, multidirectional perturbations, active human body model, validation data.

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Well folks, enjoy the rest of the show!

Jóna Marín Ólafsdóttir
September 2014
Gothenburg, Sweden

Thesis

This thesis is based on the work contained in the following publications:

Paper A

J. M. Ólafsdóttir, J. K. H. Östh, J. Davidsson and K. B. Brolin. "Passenger kinematics and muscle responses in autonomous braking events with standard and reversible pre-tensioned restraints." *IRCOBI Conference Proceedings.* 2013, pp. 602-617.

Division of work: 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 and was reviewed by all authors.

Paper B

J. Östh, J. M. Ólafsdóttir, J. Davidsson and K. Brolin. "Driver kinematic and muscle responses in braking events with reversibly pre-tensioned and standard restraints – Validation data for active human body models." *Stapp Car Crash Journal* 57 (2013), pp. 1–41.

Division of work: 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 and was reviewed by all authors.

Paper C

J. M. Ólafsdóttir, K. Brolin, J.-S. Blouin and G. P. Siegmund. "Dynamic Spatial Tuning of Cervical Muscle Reflexes to Multi-Directional Seated Perturbations." *Submitted for publication.*

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

Contents

I am so clever that sometimes I don't understand a single word of what I am saying

-Oscar Wilde

Part I

Overview

1 Introduction

Human body models (HBMs) are computer models of the human body increasingly used in automotive safety research. They have biofidelic anthropometry and material properties and are designed to predict human kinematic and injury responses under external loading. HBMs can be used in crash simulations for the evaluation of safety systems performance. The repeatability and flexibility of HBMs allow for the optimisation of safety systems through extensive simulations of different crash scenarios.

Historically, active musculature has not been incorporated in HBMs. However, several authors have pointed out that driver initiated evasive manoeuvres such as braking and/or steering prior to a crash are common [1]-[4] leading to postural changes and muscle reactions in occupants. Furthermore, with recent trends in automotive safety systems development such as autonomous braking and steering, pre-crash events can be expected to occur even more frequently in the future. With the growing recognition of the influence of muscle activity on dynamic response and injury risk of vehicle occupants [5]-[19], active muscles have increasingly been included in HBMs for improved biofidelity in pre-crash simulations [3, 9], [20]-[22].

Various techniques have been used to simulate muscle recruitment strategies in HBMs. A common objective to the different techniques is to apply a reasonable amount of muscle force around each joint at each time step to capture the dynamic response of the occupant under low loading conditions. To support the development process of simulated activation schemes a sound understanding of muscle and kinematic responses in actual pre-crash situations is essential. A feasible and ethical way to gain such knowledge is to perform volunteer experiments in replicated pre-crash events. A replicated pre-crash event consists of a controlled test environment where a non-injurious but representative acceleration pulse is applied to seated volunteers. During these events the volunteers' muscle activity can be measured with electromyography and the kinematics recorded with a camera or a motion capturing system. These types of data, along with measured boundary conditions, are again essential for proper model validation.

1.1 Active HBMs and Validation

Several active HBMs, i.e. HBMs with active musculature, have been developed to study occupant kinematics under pre-crash and low g loading [3, 8, 9, 12, 13], [21]-[24]. Most of these models are either whole body models [3, 21, 22] or models of the head and neck complex [8, 9, 12, 13, 24]. Various activation schemes have been proposed with earlier techniques consisting of arbitrary pre-defined activation curves applied to various muscles [10, 12, 13]. The resulting kinematic response from various load cases was then compared to experimental data for validation. For a more generic and biofidelic approach recent models have included forms of neuromuscular control models by simulating muscle regulation with feedback control [21, 22] and reinforcement learning [3] approaches. With these methodologies the models become more adaptive and capable of simulating a range of load cases possibly without any redefinition of model parameters. These models not only require experimental data on occupant kinematic response and boundary conditions but also muscle response data which is important for two reasons:

- *Model development*: To understand human motor control strategies specific to pre-crash events to aid the unravelling of methods of simulated recruitment strategies and for model tuning.
- *Model validation*: To allow thorough comparisons of model outputs, i.e. predicted muscle activity in conjunction with kinematics and interaction forces, to evaluate the model performance.

Volunteer experiments that include muscle activity acquisition are thus an integral part of active HBM model advancement. Measuring volunteer muscle activity under different replicated pre-crash conditions with varied loading levels and loading directions is crucial for the development and validation of models intended for simulation of various pre-crash scenarios.

1.2 Measuring Muscle Activity

Skeletal muscles consist of bundles of muscle fibres, elongated muscle cells which contain the contractile elements of the muscle, the sarcomeres (Figure 1.1). Muscles are innervated by motor neurons that transfer nerve impulses, initiated by the central nervous system, 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, so called end plate potential ignites an action potential (flux of ions through the membrane) that propagates from the neuromuscular junction across the whole muscle fibre. The propagation of action potential leads 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.

Electromyography is a technique for measuring muscle activity. Electromyography measures the action potential propagation in the muscle fibres either on the surface of the skin with surface electrodes or inside the muscle with wire (indwelling) electrodes. The measured signal, the electromyogram (EMG), is composed of the superposition of all action potentials detectable by the electrode recording the signal [25]. 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 element [25].

Estimating muscle force from EMG measurements is not a simple task and no golden formula describing the force – EMG relationship exists. Perhaps intuitively one would reason that a linear relationship exists between EMG and muscle force. However, both linear and nonlinear relationships between surface EMG and force have been found [26]. Suggested reason for a nonlinear relationship is that the perceived nonlinearity might

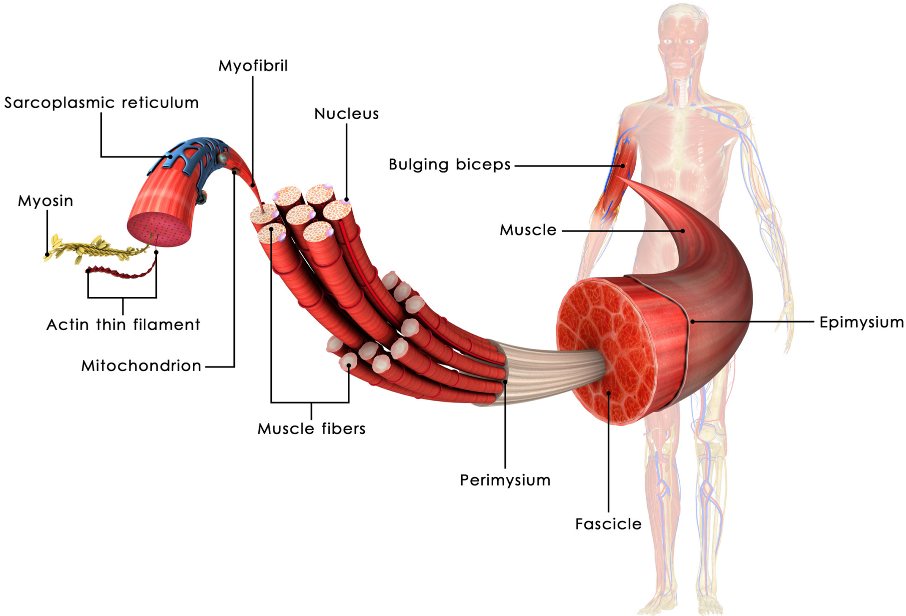

Figure 1.1: *Skeletal muscle structure.*

be due to that contributions of synergistic muscles in load sharing vary unequally with contraction level [26]. Another reason may be due to the arrangement of motor units within a muscle where smaller units are located deep in the muscle and larger units in the more superficial part. According to the size principle smaller motor units will be recruited first [27] so at lower force levels, due to a longer distance between the small units and the electrode [28], a disproportionate increase in force with increased EMG amplitude will be noticed [26, 29]. However, depending on the muscle [29] and the application, linear approximations of the force – EMG relationship can give reasonable predictions [26].
To express EMG levels in terms of muscle activation, the EMG signal is generally normalised. Normalisation also 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 between individuals and prohibits subject comparisons of EMG levels if normalisation is not included. While these issues are less problematic for indwelling electrodes, without normalising the wire EMG signals, intrasubject comparison cannot be made. Normalisation is also important due to various levels of muscle strengths of individuals. Several normalisation methods are currently used and there seems to be no consensus on which normalisation method is the most appropriate [30]. Which would be the most suitable method can of course depend on the application of a given study, however, to make comparisons between studies it is important to establish a consistent normalisation method. 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 certain resources and can be very time consuming where subjects need to perform repeated contractions for each tested muscle with resting time between contractions. Although this method has shown less intersubject variability [31] it does not allow comparisons of muscle activation levels between muscles, tasks or individuals.

1.3 Previous Experiments

Studying the human muscle responses in loading events representative of pre-crash situations provides an approximation of the activation schemes adopted by occupants in actual pre-crash events, facilitating the development and validation of strategies to simulate muscle activity. Several studies have presented volunteer EMG activity recorded during low velocity impacts or perturbations. Ejima et al. [6, 7] measured EMG, kinematics, and boundary conditions in volunteers in frontal loading conditions with a sled configuration and accelerations ranging from 0.2 – 1.0 g. The results showed that pre-tensing the muscles prior to impact resulted in smaller maximum neck flexion angles and less head excursions for both passengers and drivers. The thorough kinematic and EMG measurements clearly indicated the influence of muscle tension on kinematic response in low g frontal impacts, however, EMG levels were not normalised to MVC. Mathews et al. [32] tested children and young adults (age groups 8-14 and 18-30 years) in 3.8 g frontal sled perturbations. Although kinematic data were not presented an important finding for active HBM development was that peak MVC-normalised EMG amplitudes and activation timings were not age specific. Activation timings and peak EMG amplitudes were also presented in [33]. Although the data was not intended for active HBMs in particular they define what maximum normalised EMG levels could be expected for a range of low frontal loading levels and that the level of awareness affected the EMG response.
Neck muscle activity in low velocity rear-end impacts has been thoroughly studied [11], [15]-[18], [34]-[43]. Although rear-end impacts are not classified as pre-crash events, these studies have indicated that for low velocity cases, neck muscle activity might influence the risk of injury. Most studies provide valuable information on activation timings and maximum EMG levels during the impact relevant for active HBM development and one study provides detailed time histories of kinematic and MVC-normalised EMG data suitable for active HBM validation [39]. The data set of this latter study was limited to three subjects.
Compared to the numerous studies focusing on muscle activity during sagittal plane loading, lateral and oblique load cases have received little attention. Recent findings showed that muscle tension significantly affected volunteer kinematic responses during pure lateral and lane change type loading of 0.4 – 0.7 g [1, 2]. The presented EMG levels were not normalised to MVCs. Huber et al. [44] presented activation timings and absolute EMG levels during 1 g lane change manoeuvres. Kumar et al. [45] studied the influence of perturbation direction in three neck muscles. Only peak MVC-normalised EMG levels were presented, nevertheless the results showed that the amplitudes were affected by the

direction of the applied load.

1.4 Aims

The aims of this thesis were to:

- Investigate both passenger and driver responses in pre-crash braking scenarios with different restraint configurations to provide detailed validation data for active HBMs
- Analyse neck muscle reflex activity during multi-directional perturbations to identify the recruitment strategies in various loading directions

2 Summary of Papers

Three publications on volunteer experiments are included in this thesis (Part 2). They are briefly summarised below. Table 2.1 summarises the various test cases presented in the three publications, Papers A, B and C.

Table 2.1: *Experimental test cases presented in the thesis.*

	Occupant Role	Restraint Type	Load Case	Test Rig
Paper A	Passenger	Standard seatbelt Rev. pre-tensioner[1]	Auto braking	Vehicle
Paper B	Driver	Standard seatbelt Rev. pre-tensioner[1]	Auto braking	Vehicle
		Standard seatbelt	Voluntary braking	
Paper C	Passenger	None	Multidirectional perturbations	Sled

1.The reversible pre-tensioner applied 170 N tension 200 ms before the initiation of the autonomous braking (noted auto braking).

2.1 Summary of Papers A and B

The aim of the two papers was to investigate the muscle and kinematic response of passengers and drivers in various emergency braking scenarios in different restraint configurations. This was done to generate detailed data sets suited for the validation and initialisation of active HBMs for pre-crash braking simulations.

The experiment included twenty 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 2.1) with standard and reversible pre-tensioned seat belts and maximum voluntary driver braking with a standard seat belt (Paper B). With the pre-tensioner a 170 N tension was applied 200 ms before the initiation of the autonomous braking. 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 seat belt positions and seat indentions (driver only) to facilitate the validation of the model boundary conditions.

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 pre-tension 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 barking. 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. Drivers experienced the shortest head displacements during voluntary braking, 38 and 35 mm compared to 116 and 98 mm for females and males respectively, as a result of being prepared and bracing themselves against the steering wheel. Female and male kinematic responses were similar, especially for passengers and during voluntary driver braking.

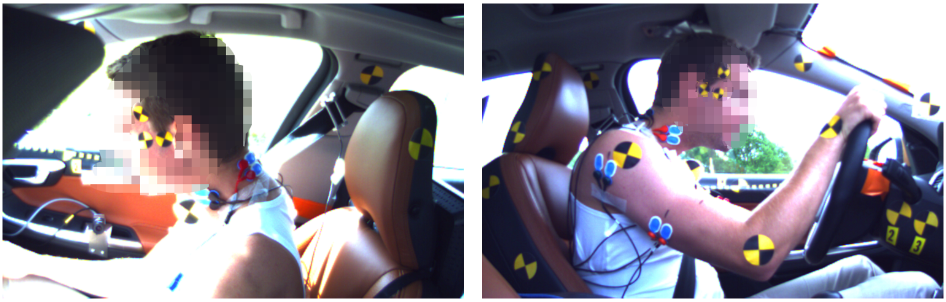

Figure 2.1: *Volunteer subjected to autonomous braking as passenger (left) and driver (right).*

2.2 Summary of Paper C

The activation patterns of neck muscles vary with the direction of the intended or imposed head motion. The quantification of these patterns is necessary for the advancement of omnidirectional active HBMs. Thus the aims of this study were to determine the activation patterns and spatial tuning of neck muscles during dynamic reflexive actions from perturbations.

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 varied with 45° intervals from forward, see Figure 2.2. 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 2.2. All wires were inserted at 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 for normalisation to provide normalisation constants. The dynamic spatial tuning patterns of each muscle were analysed at 90, 110 and 130 ms after perturbation onset.

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 directions. The findings indicated 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.

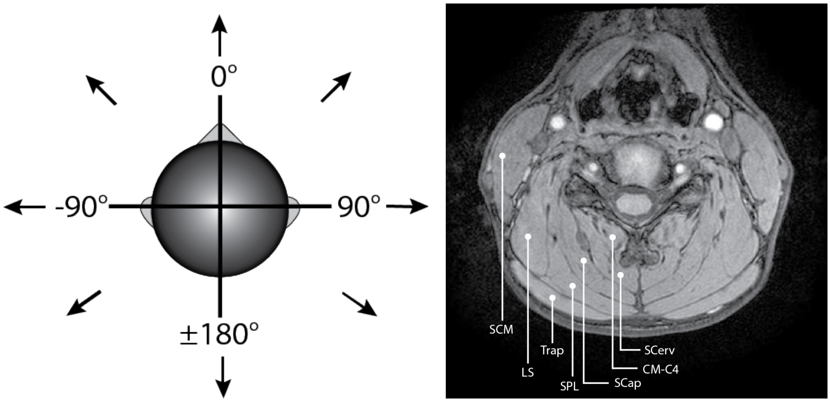

Figure 2.2: *Applied perturbation directions (left) and wire insertion locations (right). SCM: sternocleidomastoid, LS: levator scapulae, SCap: semispinalis capitis, STH: sternohyoid, Trap: trapezius, SCerv: semispinalis cervicis, CM-C4: cervical multifidus C4/C5 level, CM-C6: cervical multifidus C6/C7 level, SPL: splenius capitis.*

3 Discussion

3.1 Initial Muscle Response

During pre-crash, muscle responses can be evoked by visual, auditory, vestibular or somatosensory stimuli. In the experiments reported here no visual or auditory stimuli were present and the initial muscle activation thus most likely triggered by the vestibular and somatosensory systems, which in the case of neck muscles are referred to as the vestibulocollic and cervicocollic reflexes (VCR and CCR). VCR is driven by inertial sensors (vestibular organ) in the head that detect head motion, by sensing head velocity and acceleration, to stabilise the head in space. CCR is driven by proprioception sensory mechanisms such as muscle spindles and skin receptors to stabilise the head on the torso. These reflexes, which relative contribution depends on the direction of motion [46], might thus be triggered through the onset of head acceleration and when the torso starts to move relative to the head. It has also been speculated that in forward perturbations, mechanoreceptors in the trunk and pelvis might trigger neck muscle activity [47]. Similarly these mechanoreceptors along with receptors in the hands of drivers could potentially trigger activation in other muscle groups. From the data presented here, it has not been determined whether the observed reflex activations can be attributed primarily to VCR, CCR or other somatosensory pathways through the various aforementioned mechanoreceptors. However, it is speculated that these particular pathways are not primarily involved in triggering muscle activity for drivers and passengers restrained with a reversible pre-tensioner during autonomous braking events in Paper A and B. In these events both drivers (trials 2-4, i.e. first exposure excluded) and passengers reacted with an initial burst in muscle activity immediately after the belt was pulled before any vehicle deceleration or head motion occurred. Therefore, 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.

A similar burst, or startle, was not detected during autonomous braking with a standard belt or during the multi-directional perturbations (Paper C). Blouin et al. [36] showed that for forward perturbations the subjects' neck activities decreased with repeated trials attributed to an adopted feedforward muscle control strategy. In the experiment in Paper A and B, the first trial was always autonomous braking with pre-tension in the driver seat and in Paper C the volunteers experienced 16 forward perturbations of the same intensity prior to the multi-directional perturbations. The EMG amplitudes detected in the subsequent events were probably attenuated to some extent. Why the startle activity was reintroduced in subsequent autonomous braking trials with the reversible pre-tensioner (trials 2-4 for drivers and passenger trials) can likely be explained with the findings from a later study by Blouin et al. [17]. They found that by superposing a startling tone stimuli and perturbation in repeated trials the habituation-related attenuation of the muscle responses could be reversed to produce the same startle response as recorded in the first trial. These findings further support the speculation that the sensory information about the seatbelt pre-tensioning triggered the apparent muscle reflexes.

The duration of the startle response initiated by the belt pre-tensioning was about 200–300

ms. It is not clear whether the relatively high increase in muscle activity due to startle would reduce or aggregate the risk of injury during crash. 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 [10]. On the other hand, it has been hypothesised that in low velocity rear-end impacts increased cervical multifidus activity during a startle response, in the presence of increased loading due to abnormal vertebral motions, might strain the facet capsule to an injurious level [38, 48]. In the light of this hypothesis Mang et al.[43] have suggested a novel pre-crash safety system where a loud startling tone be presented 250 ms prior to a rear-end impact to reduce cervical multifidus activity during the impact and thereby the risk of whiplash injury [42, 43]. Omnidirectional active HBMs that incorporate 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 with different loading directions.
From the current studies it is not fully clear what kind of muscle response would follow the startle reflex in real-life pre-crash events, i.e. whether the increased muscle activity levels would be sustained or increased even further due to voluntary actions such as bracing or execution of an evasive manoeuvre. Active HBMs simulating the startle response 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 a visual and auditory stimulus, 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.

3.2 Influence of Autonomous and Driver Maximum Braking on Posture

During driver maximum braking in Paper B volunteers seemed to adapt an anticipatory response where they modified their posture to prepare for the applied deceleration and actively compensated for the imposed motion due to inertial loads instead of relying on the seatbelt. This elicited reduced head excursion and increased co-contraction. For instance, for female drivers the head excursion during steady state driver braking was 38 mm compared to 116 mm during autonomous braking with a standard belt and 51 mm with a reversible pre-tensioner tensing the belt prior to braking. Biceps activity increased from approximately 7 %MVC to 13 %MVC. Even if the forward head displacement when restrained with the reversible pre-tensioner was less than double 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 [23] and higher belt loads [49] during a frontal crash compared to a nominal posture. However, a rearward posture with the head close to the headrest can also lead to increased HIC values [23]. 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 simulated [50]. Including an anticipatory control strategy with a previously developed postural control [21, 51] provided outputs that were closer to the volunteer response than with the postural control alone. The model was only tested for a braking event but the author foresaw that the same method could be applied to simulate other driver voluntary actions such as evasive steering. Pre-crash steering manoeuvres may cause postural changes that are different from those occurring during braking. To provide validation data for omnidirectional models, occupant kinematic and muscle responses during voluntary and autonomous steering need to be quantified in future experiments.

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 belt 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 imposed forward motion to the same extent as drivers, resulting in close to double head excursion and increased back muscle activity. This indicates that the design requirements for the restraint systems are different for drivers and passengers in pre-crash braking situations. Introducing models that have been thoroughly validated against data, such as that presented here, this difference can be further assessed and the systems optimised for each group. For example, Östh et al. [21] validated an active HBM against the data from the four autonomous braking events in Papers A and B and then performed a parameter study varying the force onset and force level of a reversible pre-tensioner to investigate the effect on occupant (driver and passenger) forward displacement during autonomous braking. Although the topic of the study was not to optimise the restraint system for either group, it clearly displayed the possibility of performing such studies with validated active HBMs.

3.3 Directional Dependency of Muscle Recruitment

Muscle activity in various neck muscles was measured with indwelling electrodes in Paper C. Although surface electrodes are more commonly used in vehicle occupant studies they can only capture the activity of certain muscles, mainly muscles that are superficial to the body. Surface electrodes cannot be used to capture the deep layered muscles in the neck, for instance. Consequently, little is known about how the more than 20 neck muscle pairs are activated to control the different degrees of freedom during dynamic events. In particular, reports of the role of deeper musculature and the spatial tuning of different muscles during imposed motion and dynamic reflexive actions are largely lacking. The spatial tuning of various muscles is crucial for the development and validation of omnidirectional active HBMs. The dynamic spatial tuning patterns reported in Paper C were both muscle and direction specific, indicating that models cannot apply the same recruitment strategy during sagittal plane loading and loading in other motion planes. Furthermore, future omnidirectional active HBMs might need to implement neuromuscular control strategies that can regulate individual muscle activity to account for the distribution of activity between the different neck muscles. With such muscle control, the models ability to better predict the head-neck motion and the influence of muscle tension on injury mechanism in various loading directions can be reached. Activity was recorded in nine neck muscles for a typical passenger posture in Paper C. More studies are needed to identify the dynamic spatial tuning of deep muscles at other spinal levels (lumbar and thoracic), as well as deep musculature in other body parts. Also, the influence of different postures, such as driving posture on the spatial tuning of neck muscles remains unknown.

3.4 EMG Data for Active HBMs

To facilitate the prediction of injuries or the influence of muscle tension on injuries with active HBMs the predicted muscle forces should be realistic. The predicted muscle forces need to be validated against experimental data. Since muscle force measurements are a rather complex and invasive procedure, muscle forces are sometimes estimated from EMG measurements. As mentioned in the Introduction, linear approximations of the force – EMG relationship can often give reasonable predictions of muscle force from EMG measurements [26]. 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 during dynamic contractions is that the relative position of the electrode to the detectable and contracting muscle fibres can change [29, 52]. 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 done in [3, 50] or comparing the produced moments in the model and joint moments predicted by inverse dynamics [26]. A limitation to the latter procedure is that only a comparison of the net moment produced and not the contribution of individual muscle forces can be constituted.
For a thorough validation of active HBMs in pre-crash events three types of data are required: MVC normalised EMG data, kinematics and boundary conditions. All EMG data presented in the current work was normalised with MVC. As described in the Introduction, 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 vehicle occupant studies [1, 2, 6, 7]. 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 possible 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 both of the current experiments, although this data was

not presented for the multidirectional perturbations (Paper C). 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 2.1. The vehicle, however, 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 that closely resembles 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 this might be less important. In the Paper C experiment the volunteers were seated in a regular car seat without a seatbelt and 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 vary depending on the intended use of the respective data set.

4 Conclusions

The thesis presented data sets for development and validation of active HBMs. Papers A and B presented detailed validation data for pre-crash braking events and in Paper C activation schemes in multidirectional perturbations were analysed to identify the spatial tuning of neck muscles. Passengers and drivers in Papers A and B displayed different postural responses to autonomous braking. Drivers, as they were holding the steering wheel, were able to restrict the forward motion better by actively using their arms resulting in less forward displacement and decreased lower back muscle activity. The compensation was even more substantial during maximum voluntary braking. Pre-tensioning the belt prior to braking induced a startle response in some volunteers, both for drivers and passengers. The multidirectional perturbations in Paper C resulted in muscle reflexes that showed variable activation amplitudes and directional preference.

Several requirements for the development of future active HBMs have been recognised from the results of the current studies. Active HBMs that predict occupant response need to include strategies that can account for different occupant roles (passenger vs. driver) and that differentiate between driver initiated braking and autonomous braking. To investigate what influence startle response has on injury risk, active HBMs should be able to simulate the startle reflex. Furthermore, the muscle and direction specific activation schemes reported in Paper C imply that modelling individual muscle control would be a step forward in reaching better predictions of head-neck response to omnidirectional loading.

For future model advancements further studies on occupant responses are needed. Occupant responses during evasive steering and combined steering and braking should be identified where MVC normalised EMG, kinematics, and boundary conditions are measured. The occupant startle response to a possible visual threat should be analysed and more focus on individual muscle response measurements in future occupant experiments would be advisable.

5 Future Directions

From the current work three properties that could advance future active HBMs have been identified:

- Startle response. Pre-tensioning the seatbelt before autonomous braking induced a startle response in some passengers and drivers in Papers A and B. AHMS that simulate the startle response can be used to study the influence of startle reflex on pre-crash kinematics and injury risk during crash.
- Anticipatory response. Drivers in Paper B modified their response to the deceleration load during maximum voluntary braking. Active HBMs need to model anticipatory responses to capture the driver response during driver initiated braking and steering emergency manoeuvres.
- Individual muscle regulation. The neck muscles investigated in Paper C showed muscle and direction specific activation patterns. Implementing directionally dependent control of individual neck muscles could possibly improve kinematic and injury prediction of omnidirectional active HBMs.

To develop and validate these future models, additional volunteer studies are required. Subjecting volunteers to replicated pre-crash events with startle inducing stimuli of various kinds (tactile, visual and auditory) are needed to validate startle reflex models. It is important for active HBM development to identify muscle reactions following the startle reflex due to a realistic threat in traffic. Due to the dearth of volunteer studies into lateral or omnidirectional loading, further experiments with a lateral loading component are necessary. Data from replicated pre-crash steering and combined steering and braking manoeuvers would be useful to improve the omnidirectional response of active HBMs. To validate models of anticipatory response, studies on driver initiated emergency steering manoeuvres would be beneficial as well. Volunteer experiments should include measurements of EMG, kinematics and the appropriate boundary conditions. EMG should be normalised to MVCs and preferably measured from individual muscles.

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