

THESIS FOR THE DEGREE OF DOCTOR OF PHILOSOPHY

Passenger kinematics in evasive maneuvers

Advancing Active Human Body Modeling and Understanding Variability in Passenger Kinematics During Evasive Maneuvers

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Cover:
SAFER HBM v10 in a braking maneuver (from Paper B), initial position in transparent, the
simulation with the most slouched initial posture from Paper D.

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ABSTRACT

In situations that might lead to a vehicle crash, drivers often perform an evasive maneuver, such as braking or steering, in an attempt to avoid a crash. If a crash was not avoided, the maneuver could influence the injury outcome by altering the occupant's position. Occupants use their muscles in response to a maneuver, and because the typical accelerations are low during maneuvers, the muscle activity can influence the kinematics. Thus, it is important to include the response to these potential maneuvers before the crash when predicting occupant injuries in a crash. The response to maneuvers could be evaluated by adding active musculature to existing evaluation tools, such as human body models. Furthermore, in volunteer studies, the head and torso displacements during maneuvers vary between occupants, but the cause for this variability remains to be identified. Two aims were defined for this thesis, addressed in two parts. The first aim was to advance the active neck and lumbar muscle controllers in the SAFER HBM to predict average response to maneuvers. The second aim was to further understand why such variability is seen in occupant response to evasive maneuvers.

Three muscle controller concepts were evaluated in this thesis, two of which were aimed at emulating the reflexes responding to input from the vestibular system that control the head position in space, and one controller that emulated reflexes that respond to lengthening of muscles. For the first aim, the active muscle controllers in the SAFER HBM were updated to allow for simulations with large vehicle yaw rotations, and the predictive capabilities were evaluated in braking, steering, and combinations. In a subsequent study, the updated controllers were tuned to volunteer kinematics in braking and steering, and the model performance was evaluated in the same conditions. It was concluded that the SAFER HBM, with the updated and tuned controllers, could predict passenger head kinematics in braking and steering with good to excellent results.

The occupant variability was addressed by statistical analysis of volunteer kinematics in six different vehicle maneuvers. In two subsequent studies, the Active Human Body Model developed within the first aim was used to analyze the model sensitivity to Human Body Model and boundary condition characteristics in braking. From the analysis of volunteer kinematics, it was concluded that the belt system was the most influential predictor for head and torso displacements across all maneuvers, while other characteristics such as sex, stature, age, and body mass index were less influential. In the subsequent studies, the seat forward/rearward position and spinal curvature were found to be most influential in braking.

Keywords: Active Human Body Model, evasive maneuvers, active muscles, pre-crash, occupant variability, volunteer kinematics

To infinity and beyond!

Buzz Lightyear

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Emma

July 2023

LIST OF APPENDED PAPERS

This thesis is based on work summarized in the following publications. Contributions of Larsson E. listed using CRediT.

Paper A	Larsson E., Iraeus J., Fice J., Pipkorn B., Jakobsson L., Brynskog E., Brolin K., Davidsson J., Active Human Body Model Predictions Compared to Volunteer Response in Experiments with Braking, Lane Change, and Combined Manoeuvres, <i>Proceedings of IRCOBI Conference 2019, Florence, Italy</i>. Author contribution: Formal analysis, Methodology, Visualisation, Writing – original draft
Paper B	Larsson E., Ghaffari G, Iraeus J., Davidsson J., Passenger Kinematics Variance in Different Vehicle Manoeuvres – Biomechanical Response Corridors Based on Principal Component Analysis, <i>Proceedings of IRCOBI Conference 2022, Porto, Portugal</i>. Author contribution: Software, Formal analysis, Methodology, Visualisation, Writing – original draft
Paper C	Larsson E., Iraeus J., Pipkorn B., Östh J., Forbes P. A, Davidsson J., Predicting occupant head displacements in evasive maneuvers; tuning and comparison of a rotational based and a translational based neck muscle controller, <i>Draft manuscript</i> Author contribution: Conceptualization, Software, Formal analysis, Methodology, Visualisation, Writing – original draft
Paper D	Larsson E., Iraeus J., Davidsson J., Investigating sources for variability in volunteer kinematics in a braking maneuver, a sensitivity analysis with an active Human Body Model. <i>In review</i> Author contribution: Conceptualization, Formal analysis, Methodology, Visualisation, Writing – original draft
Paper E	Larsson E., Iraeus J., Davidsson J., Synthetic experiments to investigate occupant variability in braking maneuvers, a simulation study using Active Human Body Models., <i>Draft manuscript</i> Author contribution: Conceptualization, Formal analysis, Methodology, Visualisation, Writing – original draft

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ABBREVIATIONS

FE	Finite element
HBM	Human Body Model
AHBM	Active Human Body Model
VCR	Vestibulocollic reflex
CCR	Cervicocollic reflex
CNS	Central nervous system
PCSA	Physical cross-sectional area
BMI	Body mass index
STP	Spatial tuning pattern
MVIC	Maximum voluntary isometric contraction
EMG	Electromyography
SD	Standard deviation
CoV	Coefficient of variation (standard deviation/average)
PCA	Principal Component Analysis
PC	Principal Component
M-DRM	Multiplicative Dimensional Reduction Method
CORA	CORrelation and Analysis (software)
APF	Angular Position Feedback
MLF	Muscle Length Feedback
PID	Proportional Integral Derivative

1 INTRODUCTION

Looking beyond road traffic fatalities, tens of millions of people sustain injuries on the roads every year (World Health Organization, 2018). For instance, in a survey on health in the European Union (eurostat, 2023), 1.5% of the respondents reported that they had been involved in a road traffic accident that resulted in injury (any severity) during the last 12 months. The rate of death from road traffic accidents was 9.3 per 100,000 population (World Health Organization, 2018), and thus, by rough estimate, there were 160 injuries for every fatality on European roads. In Sweden, there were 2.8 fatalities per 100,000 population (World Health Organization, 2018), and 1.3% of respondents reported that they had been involved in a road traffic accident that resulted in injury for the same period (eurostat, 2023). Thus, for every fatality in Sweden, roughly estimated there were 460 injuries. On US roads, there were around 39,000 fatalities, and 2,280,000 injuries in 2020 (Stewart, 2022), or approximately 58 injured for every fatality. Not surprisingly, injuries (any severity) by far outnumber fatalities and that the reduction of injury rates are also an important part of the puzzle of making roads safer.

Improvements are made in traffic safety continuously (Borsos et al., 2012), with reduced risk of fatalities and injuries for occupants in newer vehicles (Ryb et al., 2011, Forman et al., 2019, Høye, 2019, Anderson and Searson, 2015). Some of these improvements can be attributed to different safety technologies such as seat belts and airbags (Kahane, 2015, Blincoe et al., 2023). Consumer tests of vehicles are used to evaluate and rate vehicles that are introduced on the market. Both for the European and US market, a high rating of occupant safety in the consumer test correlate with lower risk of injury or loss of life in real life crashes (Kullgren et al., 2019, Teoh and Arbelaez, 2022). Although high ratings in these tests correlate with improvements to real-world safety, the tests currently cover only a fraction of the potential real-world crash scenarios. For instance, consumer tests with seated anthropometric test devices are limited to well defined passenger or driver postures (for example (Euro NCAP, 2021)). In real life however, initial occupant position varies (Reed et al., 2020). Furthermore, a real-world impact is often preceded by an evasive maneuver, which could alter the occupant position.

The prevalence of evasive maneuvers prior to crashes has been quantified for different scenarios, see Figure 1. Evasive maneuvers prior to impact in cross-centerline head-on collisions on US roads, for encroaching and impacted car, was quantified in (Riexinger et al., 2019). Maneuvering prior to crashes has also been quantified for road departure crashes on US roads (Riexinger and Gabler, 2018), intersection crashes on US roads (Scanlon et al., 2015), single-vehicle crashes on US roads (Kaplan and Prato, 2016), and simulated lead-vehicle braking (Wu et al., 2017). In summary, across all studies, most drivers took some evasive action prior to an impact, but the type of maneuver depended on the situation. Furthermore, many modern vehicles are equipped with Advanced Driver Assistance Systems (Östling et al., 2019, Tan et al., 2020, Seacrist et al., 2020), that are designed to brake or steer if the driver does not react in a critical situation. With these systems many vehicle crashes can be prevented or mitigated (Östling et al., 2019, Leledakis et al., 2021a, Tan et al., 2020, Seacrist et al., 2020).

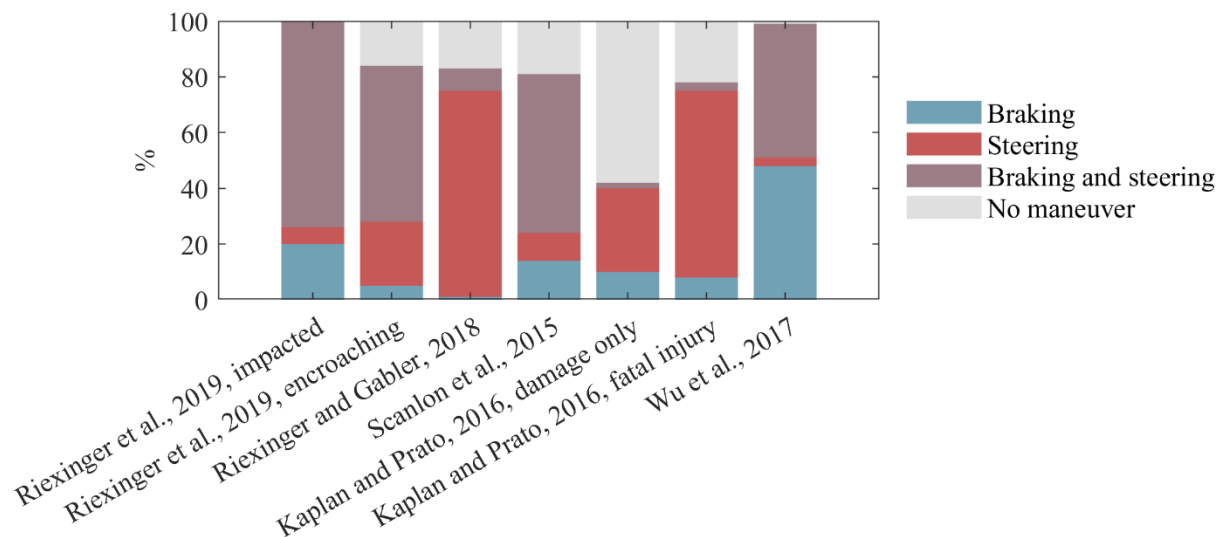


Figure 1. Proportion of crashes preceded by braking, steering, combination of braking and steering, or no maneuver.

Although these maneuvers, performed either by the driver or the driver assistance systems, often avoid or mitigate a crash, if a crash still happens the outcome of the crash can be influenced by the maneuver. Thus, there is a need to understand how the maneuver influences the occupant, and how this might influence the crash outcome. In several studies with volunteers, the occupant response to maneuvers has been quantified (Ghaffari et al., 2018, Huber et al., 2015, Reed et al., 2019, Ólafsdóttir et al., 2013). Typically, the occupants alter their position in the vehicle when exposed to a maneuver. In response to this maneuver the occupants activate their muscles (Ólafsdóttir et al., 2013, Ghaffari et al., 2019, Ghaffari and Davidsson, 2021, Huber et al., 2013), and muscle activity in turn can affect the occupant position (Beeman et al., 2011, Kirschbichler et al., 2014, Chan et al., 2022). Thus, in response to a maneuver, occupant position, posture, and muscle activity are altered compared to in normal riding.

Occupant position and posture can be influential in crash outcome. A posture where the occupant is leaning out of the seat belt has been shown to induce larger crash kinematics in frontal impacts (Donlon et al., 2020), which (through simulations) has been shown to increase injury risk (Bose et al., 2010). A trend of higher risk for serious injuries for out-of-position occupants was seen in a database study (McMurry et al., 2018). Specifically, a higher rate of injuries to thorax, abdomen, cervical spine, and lower extremities were found, although the sample size was too small to determine if this trend was statistically significant. These results agree with findings from the Volvo statistical accident database (Jakobsson et al., 2004), where a slightly higher, but not statistically significant rate of AIS 1 neck injuries in frontal impacts was seen for female occupants with rotated head postures. Occupant posture was also identified as important for crash outcomes in run-off-road crashes (Jakobsson et al., 2014). In a simulation study (Leledakis et al., 2021b), torso postures (leaning forward, inboard, outboard, or reclined) influenced crash kinematics and kinetics in frontal, near-side and far-side impacts. For frontal impacts, both forward and reclined postures increased the forward occupant displacements. For near-side and far-side impacts, a posture leaning to the opposite side of the crash (inboard in near-side and outboard in far-side) increased lateral displacements during the crash, and this effect was increased when combined with a forward-leaning posture. In another simulation study of frontal impacts (Boyle et al., 2020), a forward leaning posture resulted in decreased head and neck injury risks, and increased chest deflection.

Muscle activity has also been shown to alter forces and injury risk predictions of lower extremities and spine, without altering kinematics, in impact simulations. Specifically, leg muscle activity has

been shown in several studies to affect forces and subsequently injury risk predictions in the tibia and femur. Leg muscle activity was shown to increase the risk of tibia and femur fractures in (Nie et al., 2018), and (Li et al., 2019) found femur forces and moments to increase with muscle bracing. The same trend was identified a previous study, where tibia and femur loads were found to increase with bracing levels (Bose and Crandall, 2008). Furthermore, muscle activity has been shown to alter spinal forces and predicted injury risks. Specifically, for high-velocity impacts (>60 km/h), increasing neck muscle activity increased neck injury criteria (Nij) values (Deng et al., 2021), while for lower velocity impacts an increased neck muscle activity decreased neck injury criteria (Nij) values. In other simulation environments, a decrease in neck injury criteria (Nij) together with a slight increase in lumbar spine forces was seen when including muscle activity (Östh et al., 2020, Östh et al., 2022). Muscle activity was also found to reduce the predicted risk for rib fractures and brain injuries in far-side impacts (González-García et al., 2021).

The risk of injury is not equal among occupants. Female occupants, older occupants, obese occupants have been found to have higher risk in similar crashes. In (Forman et al., 2019), females of all ages were found to have higher risk of injury in frontal impacts compared to males of same age, stature and BMI. In a study on younger occupants, (Abrams and Bass, 2022), females were found to have increased risk of sustaining fatal or severe head and abdominal injuries in crashes. (Klinich et al., 2016) found similar injury trends for females and males, with a slight skewness for some regions. Females had a higher risk of injury to the lower extremities, while males had a higher risk of injury to the head, face, and spine. In the same study, the risk of injury in frontal impacts was found to increase with increasing BMI, and the risk of thorax injuries was higher for older occupants. In contrast to the three other studies referred above, (Klug et al., 2022) found no clear trends of injury risk based on occupant characteristics. In summary, occupants with different characteristics can be at different risk in different configurations, indicating a need to account for occupant diversity in vehicle safety evaluation.

Summary: Many crashes are preceded by some evasive maneuver such as braking or steering. If a crash cannot be not avoided, these maneuvers can affect the occupant injury risk by altering the posture and position, as well as from the altered muscle activity.

1.1 PREVIOUS WORK

1.1.1 Active Human Body Models for simulations of evasive maneuvers

Several human body models (HBMs) that can respond to evasive maneuvers have been developed. Typically, these models deploy feedback control to activate the muscles. There have been two main modelling approaches, relating to two different physiological reflexes. The angular position feedback (APF) controllers use joint angle deviations to activate the muscles, and a single controller is typically used to activate several muscles. APF controllers are primarily used to emulate the vestibulocollic reflexes (VCR) (Keshner, 2009, Binder et al., 2009b) that respond to rotational and translational acceleration of the head, and maintain the head orientation in space, but have been extrapolated to other body segments as well. Muscle length feedback (MLF) controllers emulate stretch reflexes present in muscle spindles, which respond to lengthening of individual muscles (Binder et al., 2009a, Keshner, 2009), and attempt to counter this by increasing activation. MLF controllers are usually implemented with one or more PID controllers per muscle.

The MADYMO multi-body HBM representing an average sized male (Meijer et al., 2012, Meijer et al., 2013), available through the MADYMO software, has active muscles in the neck, upper and lower extremities, and joint actuators in the thoracic and lumbar vertebral joints. The models muscle

activation is controlled by APF controllers, and intermuscular load sharing is based on the muscle function in the model (Nemirovsky and Van Rooij, 2010). The reference posture is determined in a global or local coordinate system, depending on user input. The model has been validated in braking and lane change (Diederich et al., 2021), vibrational loading, and slalom maneuvers (Mirakhorlo et al., 2022).

The active Total Human Model for Safety (THUMS) v6 model (Kato et al., 2017, Kato et al., 2018), available as a small sized female, average sized male and large sized male has active muscles in the neck, lumbar, upper and lower extremities. Muscles in 17 regions are controlled by 36 PID controllers, with intermuscular load sharing determined by anatomical descriptions from textbooks. The reference posture in THUMS is determined in local coordinate systems. For example, head/neck joint angles are determined in a coordinate system attached to the torso. The THUMS model also includes additional PID controllers aimed at producing bracing forces in the hands and feet. The model was validated in the driver position using frontal sled decelerations, at 2.5 g and 5 g. In addition, a previous version in the THUMS model (v5) has been validated in a passenger position in lateral loading (Iwamoto and Nakahira, 2015).

The active Global Human Body Models Consortium (GHBMC) model (occupant simplified v2.3) (Devane et al., 2022), available as a small female and average sized male, has active muscles implemented in neck, lumbar, upper and lower extremities. The neck muscles are controlled by both APF and MLF control, using 210 MLF controllers and three APF controllers. The lumbar, upper and lower extremities use APF controllers, 29 APF controllers were used to control the scapula-thoracic, glenohumeral, elbow, wrist, thorax-pelvis, hip, knee and ankle joints. Thus, in total, the model has 32 APF controllers and 210 MLF controllers. For the APF controllers, intermuscular load sharing was determined by studying the line of action in the model. The controller gains have been tuned to match responses of relaxed volunteers in low-speed frontal impacts at 1 g and 2.5 g acceleration levels. A previous version of the model has been validated in the driver position using frontal sled decelerations, at 2.5 g and 5 g (Devane et al., 2019).

In another neck muscle controller implementation used in the GHBMC (Correia et al., 2021), PID controllers were used to emulate vestibular and muscle stretch reflexes. Head rotations were used as input to the APF controller and stretch of trapezius and sternocleidomastoid muscles were used as input to the MLF controller. APF control was only activated if the head rotated more than 5°. Flexors were activated with a 5 times higher muscle activity than extensors. The model was evaluated in frontal, lateral and rear impacts.

In the SAFER HBM (Pipkorn et al., 2019, Iraeus and Pipkorn, 2019), lumbar, neck, and arm muscles are controlled by APF controllers (Ólafsdóttir et al., 2019, Östh et al., 2014). Muscles in six body regions are controlled by six PID controllers. Leg muscles can be activated with pre-recorded muscle activity data. In addition, MLF can be used to control the neck and lumbar muscles. In both neck and lumbar muscle controllers, the PID controller responds to angular displacement between two defined anatomical points, and intermuscular load sharing is based on directionally dependent muscle activations recorded from volunteers. Neck and lumbar controllers aim at maintaining the posture in the global coordinate system. The model can be used in simulations of braking for both occupants of both driver and passenger positions and a previous version of the model has been validated in 1.1 g braking for both driver and passenger positions (Östh et al., 2015, Östh et al., 2014, Östh et al., 2012).

THUMS-D, based on THUMS (v3), has a hybrid muscle control system with both feedback and open-loop feedforward control (Martynenko et al., 2019), implemented in the thoracic region, neck, and

upper extremities. The feedback portion uses MLF control, while the open-loop feedforward has a pre-defined level of activation. Although a MLF controller does not have an explicitly defined reference posture, setting the reference posture with defined muscle lengths means that the posture maintained will be a local posture. The model was validated in the passenger position in lane change and braking (Martynenko et al., 2019).

In addition to these full-body models, there are several models that represent selected body segments, such as a head-neck model with separate APF and MLF controllers that developed and validated against perturbation anterior-posterior loading at different frequencies (Happee et al., 2017).

1.1.2 AHBM controller tuning

Active muscle controllers have been tuned in several previous studies.

Putra et al. (Putra et al., 2021) tuned the active muscle controllers in the ViVA model to kinematics in rear-end impacts. An isolated head-neck model was used, six parameters (P and D gains, neural delay and three activation dynamics time constants) of the model were tuned to minimize difference between model and volunteer head translation and rotation time histories, and/or vertebral rotation time histories. The LS-OPT curve mapping algorithm was used to compare models and volunteers. A metamodel-based optimization, sequential response surface method (SRS) with domain reduction was used.

Devane et al. (Devane et al., 2022) tuned the GHBM small female and average male models to volunteer kinematics in low-speed frontal impacts. Eleven controller parameters, for four joint angle (neck (PD), lumbar (PID), upper (PID) and lower extremity (PID)) and 210 muscle length controllers (PD), were tuned using a single stage iteration with space filling design. The models were rated based on an overall score that combined relative error between peak forward displacements, and CORA scores for force time-histories. The best models were manually selected based on the overall score. In a later publication (Devane and Gayzik, 2023), the gains (PID) of the upper extremities were tuned to minimize mean square error between the model and volunteer hand displacements in weight drop experiments. A metamodel-based optimization, SRS with domain reduction was used.

Öst et al. (Öst et al., 2015) tuned the SAFER HBM to volunteer kinematics in braking maneuvers. Ten controller parameters (P and D gains of head, neck, lumbar, shoulder and elbow controllers) were tuned to minimize difference between model and volunteer head and torso displacement, seat belt and steering column force time-histories. The weighted integrated factor (WIF) method (Hovenga et al., 2005) was used to compare simulations and volunteers. A single stage iteration with space filling design with meta-modelling was used to tune the gains.

Correia et al. (Correia et al., 2020) tuned the GHBM model open-loop muscle activation in frontal impacts at different acceleration levels (2-15 G) to match average resultant head kinematics in frontal impacts. A metamodel-based optimization, sequential response surface method (SRS) with domain reduction was used. In a subsequent study (Correia et al., 2021), the gains of the APF and MLF PID controllers were tuned to minimize the difference between muscle activation from the tuned open-loop muscle activations in the first study, and muscle activity produced by the PID controller. The input to the PID controllers were based on head kinematics from volunteers (APF controller) and simulated muscle stretches (MLF controller, from the optimized model in first study), and the output was muscle activity.

1.1.3 Application of AHBM to evaluate crash safety

One of the intended uses with active SAFER HBM is to evaluate the occupant injury risk in an impact following a maneuver. There are several different methods to use HBMs to evaluate the occupant response to a subsequent crash.

The SAFER HBM employs a “Whole-Sequence” approach, where the maneuver and crash are evaluated in the same simulation. The Whole-Sequence simulation with the SAFER HBM was first presented in 2016. In (Östmann and Jakobsson, 2016), the difference between a frontal impact at original driving speed, and braking (two configurations, inertia reel and pretensioned) followed by a frontal impact at reduced impact speed was investigated. In that study, it was possible to use the Whole-Sequence simulation approach to develop collision mitigation systems with respect to occupant protection. In (Saito et al., 2016), a pre-pretensioned belt was shown to reduce chest compressions and increase HIC and BrIC (Takhounts et al., 2013), in simulations of braking followed by a frontal impact. For reclined occupants, a braking maneuver prior to impact successfully repositioned the occupant from the reclined to an upright position, although due to the flexibility of the lumbar spine the pelvis was not rotated to a fully upright position (Östh et al., 2020). Extrapolating muscle control from the maneuver to impact showed that including active muscles during the impact could alter the predicted injury risk (Östh et al., 2022). Including belt pre-tensioning in the braking maneuver reduced occupant head lateral displacements in the subsequent far-side impact (Wass et al., 2022).

The same Whole-Sequence approach was used in a study with THUMS TUC-VW AHBM (active) and THUMS TUC (VPS) HBM (passive) to study the influence of in-crash muscle activations in a far-side impact (González-García et al., 2021). The results show that larger lateral displacements in the crash were found for the active models with no or low muscle activity, compared to the models with higher levels of muscle activity. Using the active model reduced the predicted risk of rib fractures compared to using the passive model. Larger lateral displacements were also found for simulations that included braking prior to the crash, compared to simulations with constant velocity prior to impact.

An alternative approach to the “Whole-Sequence” is to use an active human body model to simulate the maneuver, and then transfer the results from the last state of the HBM after the maneuver to a passive HBM and use the passive HBM for the impact simulation. This approach has been used for instance with THUMS, to simulate braking with THUMS v5, followed by frontal impact simulated with THUMS v4 (Yamada et al., 2016). Trajectories and belt forces, but not muscle forces, were transferred from the braking to the frontal impact. The same approach was used to simulate braking and lane change (THUMV v5) followed by frontal and side impacts (THUMS v4) (Matsuda et al., 2018). This approach was generalized to allow for the combination of any maneuver model and impact model. First, in a study with THUMS-D, to simulate a combined turn and brake maneuver with active THUMS-D, followed by a side impact with passive THUMS-D (Öztürk et al., 2019). Occupant position and velocities were transferred from the final state of the maneuver to the first state of the crash. This transfer approach was further generalized to allow for any combination of finite element or multi-body maneuver model and impact model (typically FE) (Dominik Breitfuß, 2022).

1.1.4 Occupant variability

The effect of human and environmental characteristics on kinematics in evasive maneuvers has been investigated in several studies, Table 1. Some studies were conducted in a simplified environment, such as sled tests and laboratory tests, while some tests were performed in-vehicle. Both driver and

passenger postures have been included, and longitudinal (braking, frontal impacts), lateral (lane change, lateral impacts) and combinations have been included in the publications. Different belt systems have been included, such as standard, inertia-reel belts (2-point, 3-point, 4-point), and 3-point belts with electrical reversable retractors (ERR) that tensioned the belt prior to the maneuver (pre-pretension).

Table 1. Previous work investigating effects of occupant and environmental characteristics on kinematics in evasive maneuvers. "Inertia reel" belts refer to the standard 3-point inertia reel belts.

Reference	Environment	Occupant posture	Belt	Maneuver	Acceleration [m/s ²]
Arbogast et al. (Arbogast et al., 2012)	Sled	Passenger	inertia reel	Lateral impact	18.7
			ERR		
			inertia reel	Lateral oblique	18.5
			ERR		
Beeman, Kemper et al. (Beeman et al., 2016, Beeman et al., 2011, Beeman et al., 2012, Kemper et al., 2014)	Sled	Driver	inertia reel	Frontal impact	24.5-49.1
Chan et al. (Chan et al., 2022, Chan et al., 2021)	Sled	Driver	inertia reel	Frontal impact	9.8-24.5
				Frontal oblique	9.8-24.5
Ghaffari et al. (Ghaffari et al., 2018, Ghaffari et al., 2019, Ghaffari and Davidsson, 2021, Ghaffari, 2021)	In vehicle	Passenger	inertia reel	Lane change	5.8
			ERR		
Graci et al. 2019 (Graci et al., 2019)	In vehicle	Passenger	inertia reel	Braking	7.2-9.2
Graci et al. 2020 (Graci et al., 2020)	In vehicle	Passenger	inertia reel	Lane change	7.3
Graci et al. 2022 (Graci et al., 2022)	Sled	Passenger	inertia reel	Braking	9.8
			ERR		
Holt et al. (Holt et al., 2020)	Laboratory	Passenger	inertia reel	Lateral cyclic loading	5.2-7.2
			ERR		
Huber, Kirschbichler et al. (Kirschbichler et al., 2014, Huber et al., 2013, Huber et al., 2015, Huber et al., 2014)	In vehicle	Passenger	2-point	Braking	10
				Lane change	10
			inertia reel	Braking	10-11
				Braking lane change left	8-10
				Braking lane change right	

				Lane change left	10
				Lane change right	
Kempter et al. (Kempter et al., 2018, Kempter et al., 2022)	Simulator	Passenger		Braking	2.2
Reed et al. 2018 (Reed et al., 2018)	In vehicle	Passenger	inertia reel	Braking	9.8
				Lane change	6.8
Reed et al. 2021 (Reed et al., 2021)	In vehicle	Passenger	inertia reel	Braking	8.8
				Lane change right	6.8
				Lane change left	6.8
				Turn and brake	6.8
Van Rooij et al. (Van Rooij et al., 2013a, Van Rooij et al., 2013b)	Laboratory	Driver	4-point	Lane change	5
		Passenger			
Ólafsdóttir, Östh et al. (Ólafsdóttir et al., 2013, Östh et al., 2013)	In vehicle	Passenger	inertia reel	Braking	11
			ERR		
		Driver	inertia reel		
			ERR		

The volunteer characteristics from the included studies are shown in Figure 2. Typically, less than 20 volunteers were included, and six of the previous studies focused exclusively on the male population. Weight was reported more often than BMI. The average age was typically below 40.

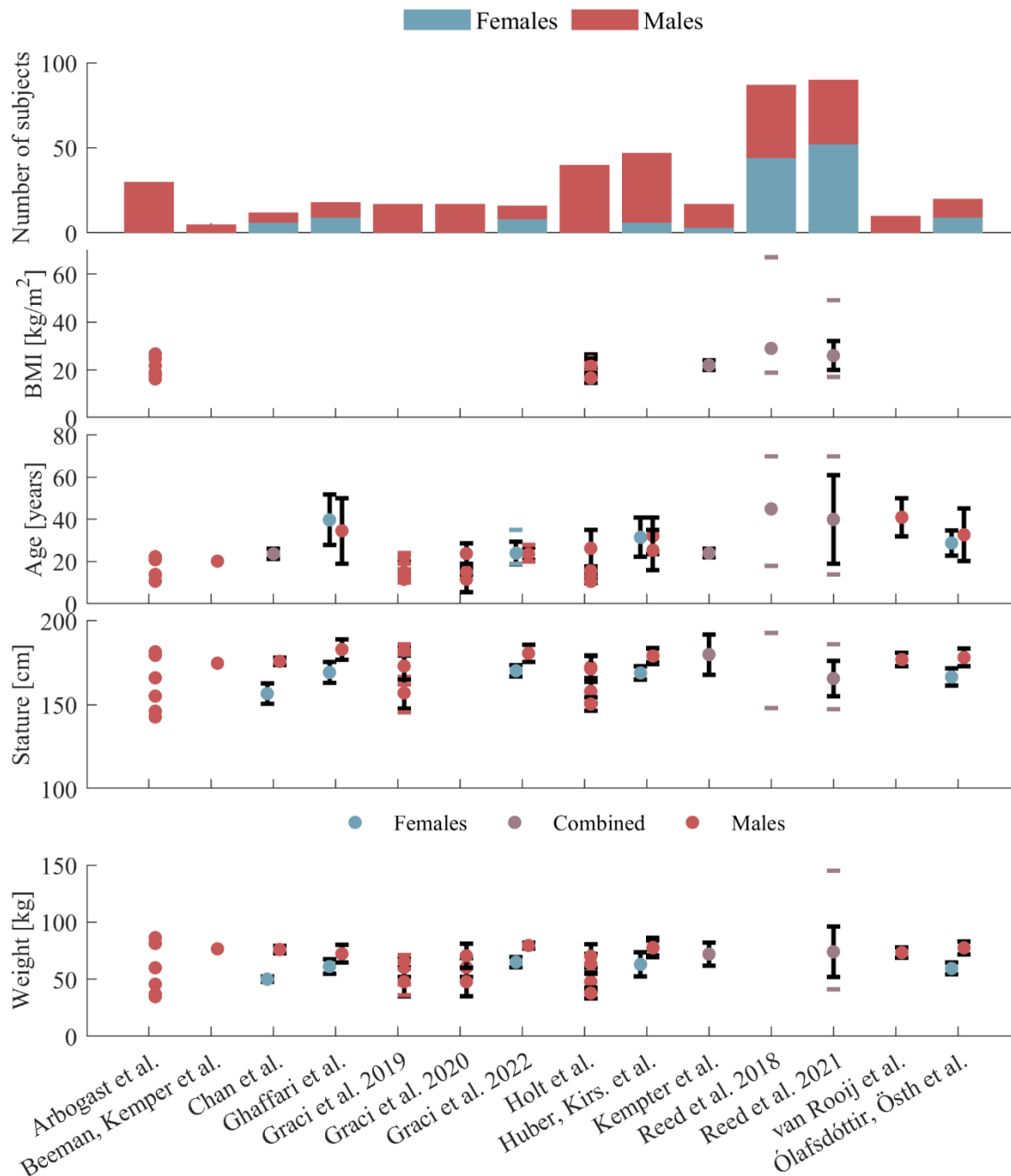


Figure 2. Number of subjects, BMI, age, stature and weight in different publications. Average values are represented with dots, standard deviation with error bars and minimum and maximum with horizontal, colored lines. Females in blue and males in red. For some publications, BMI, age, stature and weight was presented for females and males together, these are shown in purple. In some publications, such as Arbogast (Arbogast et al., 2012), BMI, age, stature and weight for several groups were presented, and for those publications, each group was plotted. Full references are presented in Table 1.

Differences in head displacements have been correlated to many characteristics of the occupant, vehicle, or maneuvers. A common trend was the influence from the belt system, exposure and bracing. Belt pretension reduced displacements and bracing reduced displacements in all studies where they were included, Figure 3. For all studies except Ólafsdóttir, Östh et al. (Ólafsdóttir et al., 2013, Östh et al., 2013) and Graci et al. 2022 (Graci et al., 2022) exposure reduced displacements or displacement velocity. In some studies, males exhibited larger displacements than females, while in other studies, males and females had similar displacements. Age increase was associated either with reduction in displacements, or not associated with any change in displacement. Typically, BMI was not associated with changes in head displacement.

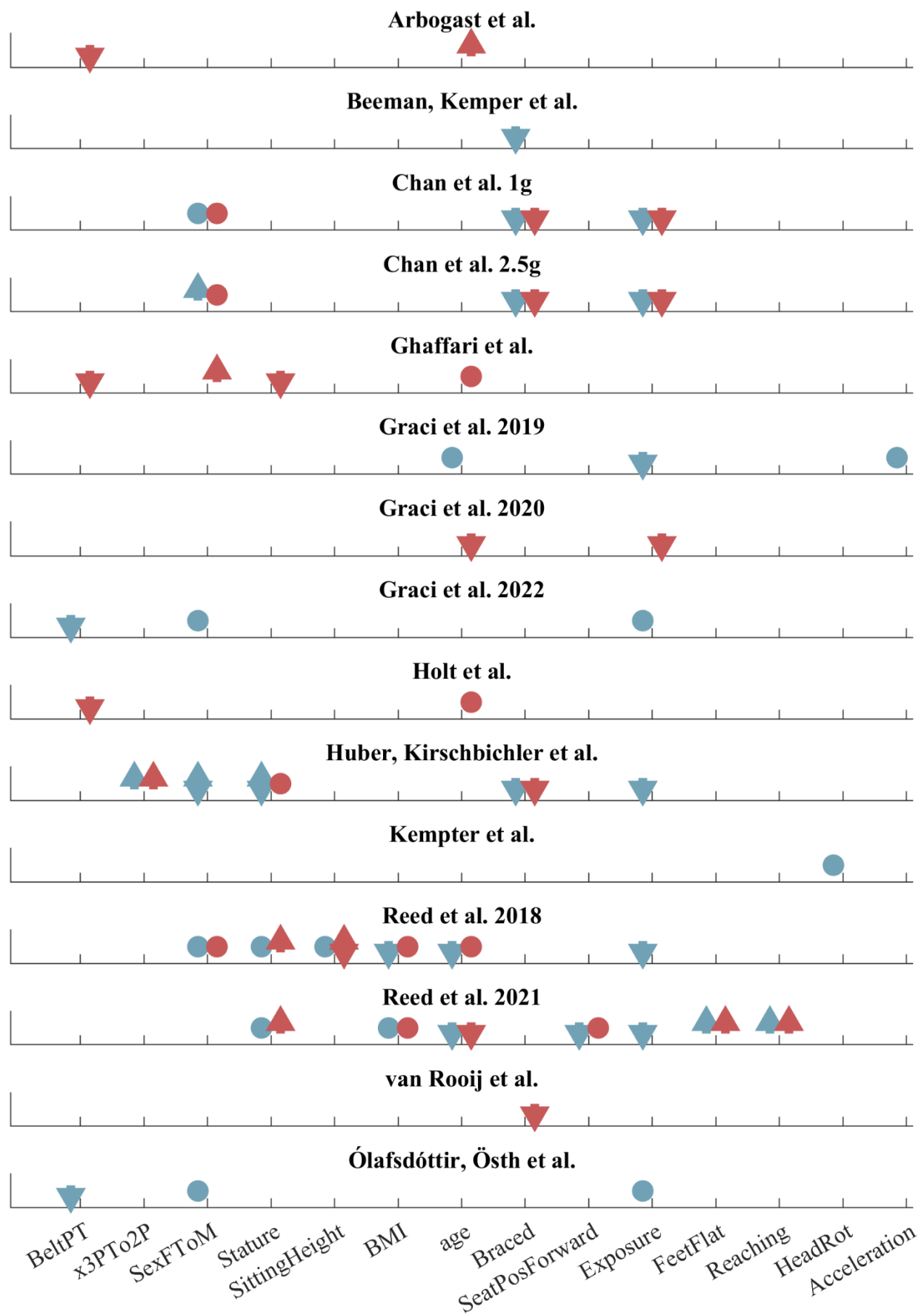


Figure 3. Effect from change in characteristic on head displacement (torso displacement for Arbogast study). Red markers are tests with longitudinal acceleration, blue markers are tests with lateral or oblique accelerations. Arrows indicate the

direction of head displacement difference for that specific characteristics change (upwards indicates larger displacement if change to binary characteristic or increase in numerical characteristic). Overlaid arrows upwards and downwards indicates that there was a difference between conditions, but the direction of change was not specified. Dots show characteristics that were included in the analysis but found not to be influential. Full references are presented in *Table 1*.

For the previous volunteer experiments, the average coefficient of variation (CoV, standard deviation divided by average) for head peak displacement was around 40%, Figure 4.

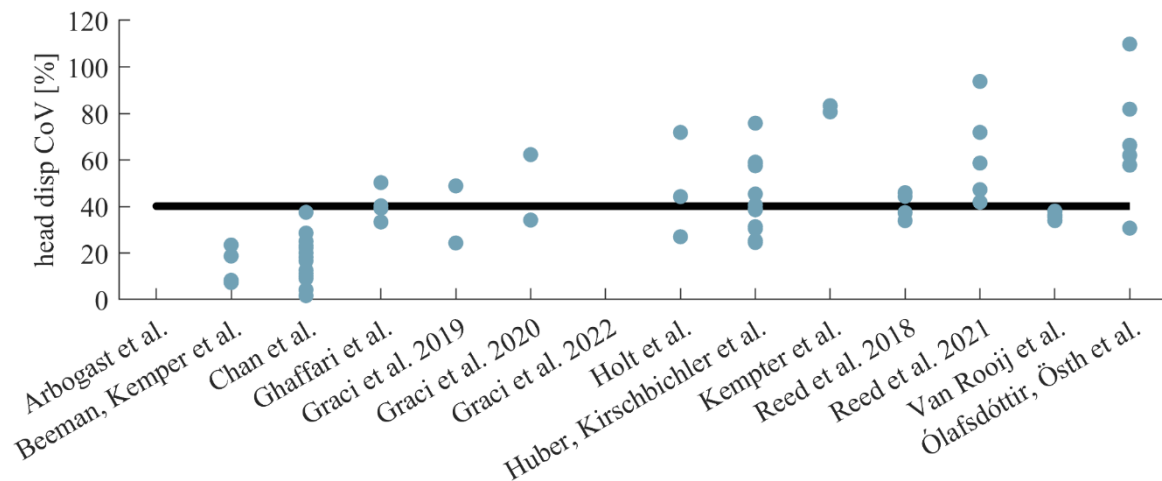


Figure 4. Coefficient of variation (CoV) for head displacement in different studies. The black line shows average CoV. Full references are presented in *Table 1*.

In a more controlled environment (sled), the coefficient of variation was smaller than in a less controlled environment (in vehicle).

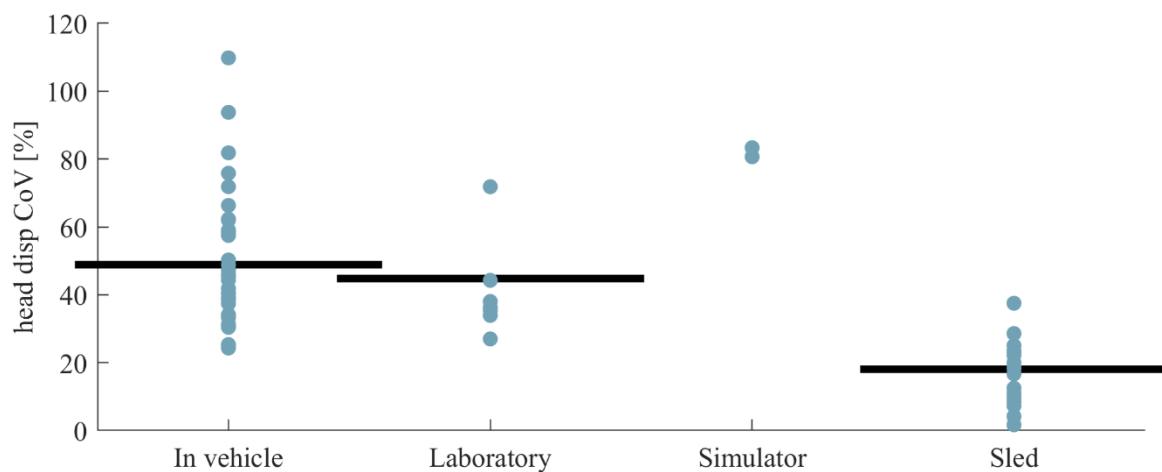


Figure 5. Coefficient of variation (CoV) for head displacement sorted by environment. The black line shows average CoV per environment. No average was calculated for the simulator environment because only one study (Kempter et al. (Kempter et al., 2018, Kempter et al., 2022)) was classified as a simulator study. Full references are presented in *Table 1*.

2 AIMS AND OBJECTIVES

The motivation for this work was to aid the design of safer vehicles in a way that includes occupant diversity. More specifically, targeting the variability in occupant response to maneuvers that occurs moments prior to a crash. Two aims and objectives were defined for this thesis.

1. The first aim was to further develop the active muscle controllers of SAFER HBM to more accurately model passenger response to evasive maneuvers. The objective was to increase the objective bio-fidelity rating score of SAFER HBM in braking and lane change.
2. The second aim was to identify what drives the variability in passenger kinematics in evasive maneuvers. Specifically, the objective was to identify important human and vehicle characteristics that influence kinematics in evasive maneuvers, and to quantify their expected effect on passenger kinematics.

3 IMPORTANT METHODS

For the first objective, the controllers were updated in two simulation studies, Paper A and Paper C, Figure 6. In Paper A, the neck and lumbar controllers of SAFER HBM v9 were updated and performance of the model was compared to a model without active muscles, in braking, lane change and combinations of both maneuvers. In Paper C, a new neck muscle controller was developed, and both the existing and new controllers were tuned to minimize difference between HBM and volunteer displacements in braking and lane change maneuvers through multi-objective optimization. SAFER HBM v10 was used in Paper C. Three studies were addressed the variability of kinematic responses. In Paper B, results from volunteer tests were analyzed and presented as regression models with belt system, sex, stature, age, and BMI as predictors. Papers D and E were simulation studies using SAFER HBM v10. In Paper D, a sensitivity study was performed to identify important characteristics of the HBM. In Paper E, a sensitivity study was performed to identify important boundary conditions. After that, a synthetic experiment was performed by varying the three most important HBM characteristics from Paper D and the three most important boundary conditions from Paper E.

The controllers from Paper A were used in Papers C and D, while the updates from Paper C were used in Paper E. The results from Paper B were used in Papers C-D. The results from Paper D were used in Paper E.

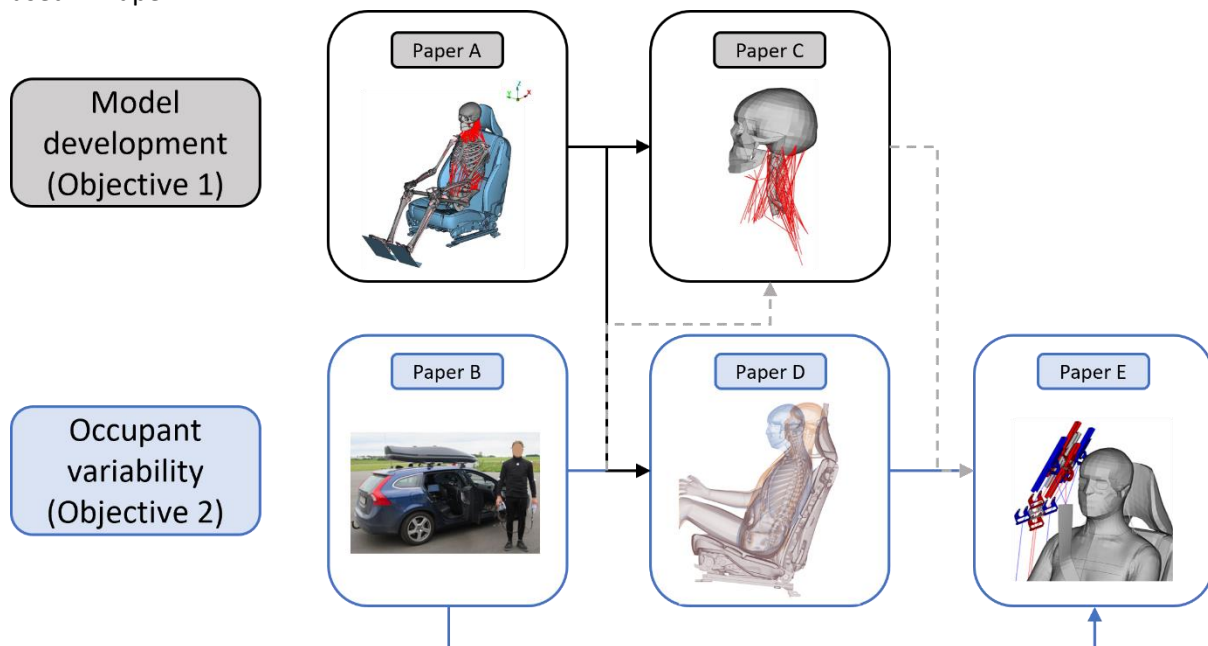


Figure 6. Included papers, how they relate to the objectives, and to each other.

In the following subsections, an overview of established methods that were used in several of the studies are presented.

3.1 PRINCIPAL COMPONENT ANALYSIS (PCA)

For a high-dimensional data set, the data can be transform from a high-dimensional data set to a low-dimensional data set while preserving the most important variability by using Principal Component Analysis (PCA) (Jolliffe and Cadima, 2016, Eriksson et al., 2013, James et al., 2021). It does this by identifying a set of orthogonal vectors that are uncorrelated, sorted such that most variability is in the first PC direction, the second-most in the second PC direction, and so on. This is

typically done around the average. For each component, each sample has a PC score, to indicate how far away from the average the sample is in that direction. If all PCs are retained, the amount of data needed to describe each test is slightly larger than the initial data. However, since the components are sorted according to importance, it is usually enough to retain only a few of the first components to get a fair description of the original data, if there is enough similarity in the data. This is where the data goes from high-dimension to low-dimension. Applied on a two-dimensional data set, Figure 7, for a data set where there is more variability in one direction (upper row), PCA identifies and aligns the first PC with this direction. For data where there is no trend (lower row), PCA identifies a direction with most of the variability, but since there is no clear direction of most variability, the PC transformation is less powerful, since more principal components are needed to describe the data.

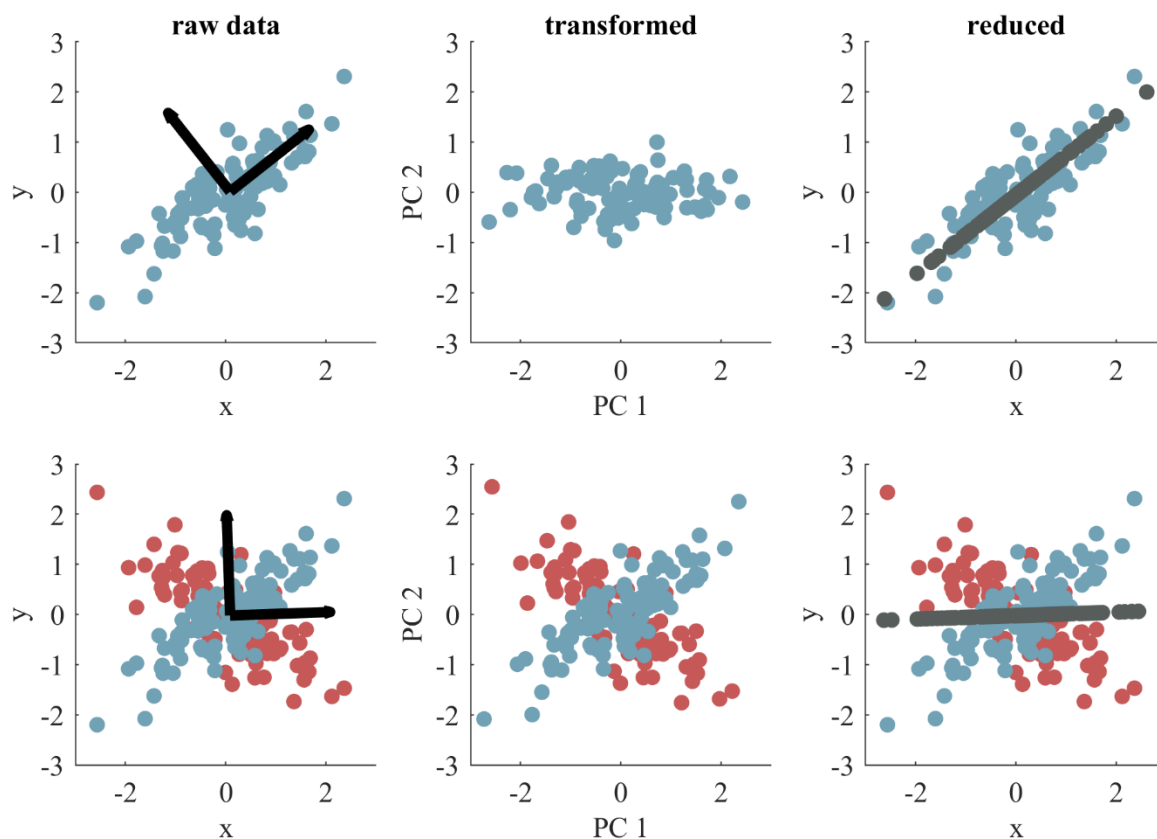


Figure 7. Example of PCA transformation. In the upper row, a set with a linear relation between x and y , with some random variability around this relation was used. In the lower row, two data sets were combined (red and blue), both with a linear relation between x and y , but with opposite direction. To the left, raw data in the Cartesian coordinates is shown, with principal components visualized with black arrows. In the middle, the two data sets have been transformed into their principal components. The data set to the left, in blue. To the right, the original data has been reduced, by using the first principal component only, and omitting the second principal component.

Use of method:

- **Paper B.** Used to parametrize kinematics and belt forces, to allow for creation of regression models for time-series kinematics and belt forces.
- **Paper D.** Used to parametrize the spinal curvatures prior to parameter variation in sensitivity analysis.
- **Paper E.** Used to parameterize acceleration pulses prior to parameter variation in sensitivity analysis.

3.2 MONTE CARLO AND LATIN HYPERCUBE SAMPLING

A common method to approximate solutions to problems with varying inputs is Monte Carlo sampling (Fishman, 1996b). The Monte Carlo sampling is used to generate new samples, based on probabilities from previous data. For each sample, data points (or parameter values) are drawn at random based on the probability distribution for that parameter/data point. Although the method is widely used and accepted, the method requires a large sample size, even for a small number of inputs typically thousands of samples are needed for the method to be reliable (Fishman, 1996a). An alternative to a traditional Monte Carlo sampling is Latin Hypercube sampling. In Latin Hypercube sampling (McKay et al., 1979), the number of samples (N) is set a-priori, and the distribution is divided into bins of equal probability, one per bin is sampled, Figure 8. Compared to traditional Monte Carlo, the samples become representative of the prescribed distribution by design, and thus a smaller sample size is needed to ensure a representative sample. The Latin Hypercube sampling is often referred to as a stratified Monte Carlo sampling.



Figure 8. Monte Carlo and Latin Hypercube simulations in one dimension, with $N=10$. Black lines show the probability distribution. For the Latin Hypercube simulations, the bins with equal probability are indicated by the filled areas.

Use of method:

- **Paper B.** Monte Carlo sampling used to generate corridors, $N=10000$.
- **Paper E.** Latin Hypercube sampling used in second part of study to randomly sample the HBM parameters and boundary conditions, $N=20$.

3.3 SENSITIVITY ANALYSIS

Sensitivity analysis is used to relate uncertainties in model input parameters to proportions of uncertainties in the model output (Saltelli et al., 2008). Sensitivity analysis is typically used to help prioritize among parameters and guide modelling simplifications. There are many methods of performing sensitivity analysis, and fundamentally, these can be categorized as local sensitivity analysis or global sensitivity analysis. In local sensitivity analysis, parameters are varied around one point, and One-At-A-Time approaches are often used, where one parameter at the time is changed, Figure 9. In global sensitivity analysis, the parameters are changed together to investigate the sensitivity to that parameter in the global parameter space, Figure 9.

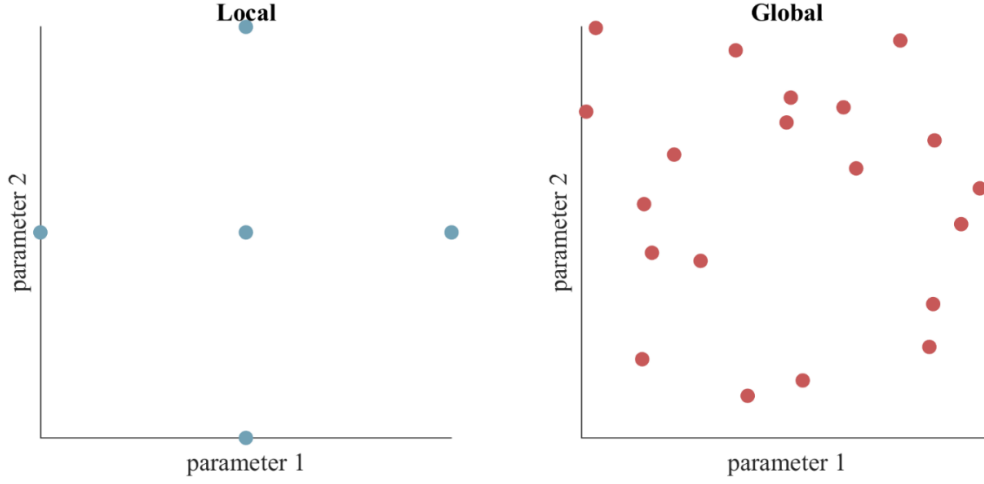


Figure 9. Local and global sensitivity evaluations.

A measure of global sensitivity is the Sobol first order sensitivity index, S_i (i denotes the parameter). For a model with input parameters $\mathbf{x}$ and model output $\mathbf{y}$, S_i is calculated according to Equation 1, where $E_{x_{\sim i}}(\mathbf{y}|\mathbf{x}_i)$ is the expected mean value when one parameter (parameter i) is fixed, $V_{x_i}(E_{x_{\sim i}}(\mathbf{y}|\mathbf{x}_i))$ describes the variance in mean value when varying the point at which the parameter is fixed, and $V(\mathbf{y})$ the unconditional variance (i.e., no parameters fixed). Thus, $S_i=1$, means that all the total variance in the data set can be attributed to variations of that specific (i :th) parameter.

$$S_i = \frac{V_{x_i}(E_{x_{\sim i}}(\mathbf{y}|\mathbf{x}_i))}{V(\mathbf{y})}$$

Equation 1

Or, as explained by (Saltelli et al., 2019), “ S_i is the expected fractional reduction in the variance of $\mathbf{y}$ that would be achieved if factor x_i could be fixed.”. Typically, Monte Carlo methods are used to analyze the global sensitivity, but as discussed in section 3.2, this requires a large (>1000) sample size (Zhang and Pandey, 2014), which is unfeasible for computationally expensive models.

3.3.1.1 M-DRM

The multiplicative dimensional reduction method (M-DRM) presented by (Zhang and Pandey, 2014) is a method to approximate the sensitivity of model response to model parameters, with substantially fewer evaluation points compared to a Monte Carlo based approach. With the method, the first order sensitivity index is approximated through local one-at-a-time variations. In short, the model output $\mathbf{y}$, depending on input parameters $\mathbf{x} = [x_1, \dots, x_n]^T$, can be described through some function, $\mathbf{y} = h(\mathbf{x})$. The function h is approximated with reference to a fixed input point (cut-point) with coordinates $\mathbf{c}$. The function is approximated for one of the parameters at the time, with the other parameters kept at their cut-point, Equation 2.

$$h(\mathbf{x}) \approx h_0^{1-n} \prod_{i=1}^n h(x_i, \mathbf{c}_{-i})$$

Equation 2

The mean and mean square (ρ_i and θ_i) can then be approximated using one-dimensional integrals, computed numerically with Gaussian quadrature, Equation 3. w_{ij} describes the Gauss weight for the i :th parameter and j :th Gauss point.

$$\begin{cases} \rho_i \approx \sum_{j=1}^N w_{ij} h(X_i^j, \mathbf{C}_{-i}) \\ \theta_i \approx \sum_{j=1}^N w_{ij} [h(X_i^j, \mathbf{C}_{-i})]^2 \end{cases}$$

Equation 3

Using the approximative mean and mean squared (ρ_i and θ_i), the first order sensitivity of the model to the selected parameter can be approximated according to Equation 4.

$$S_i \approx \frac{\theta_i / \rho_i^2 - 1}{\left(\prod_{k=1}^n \theta_k / \rho_k^2 \right) - 1}$$

Equation 4

With this approach, the number of simulations needed to evaluate the sensitivity of the model to n parameters, with N Gauss points becomes at most nN . If the nominal model and the middle point for evaluation is the same for all parameters, this is reduced to $n(N-1) + 1$. For five Gauss points, and 5 parameter variations with 5 different assumed distributions, this could be described as in Figure 10.

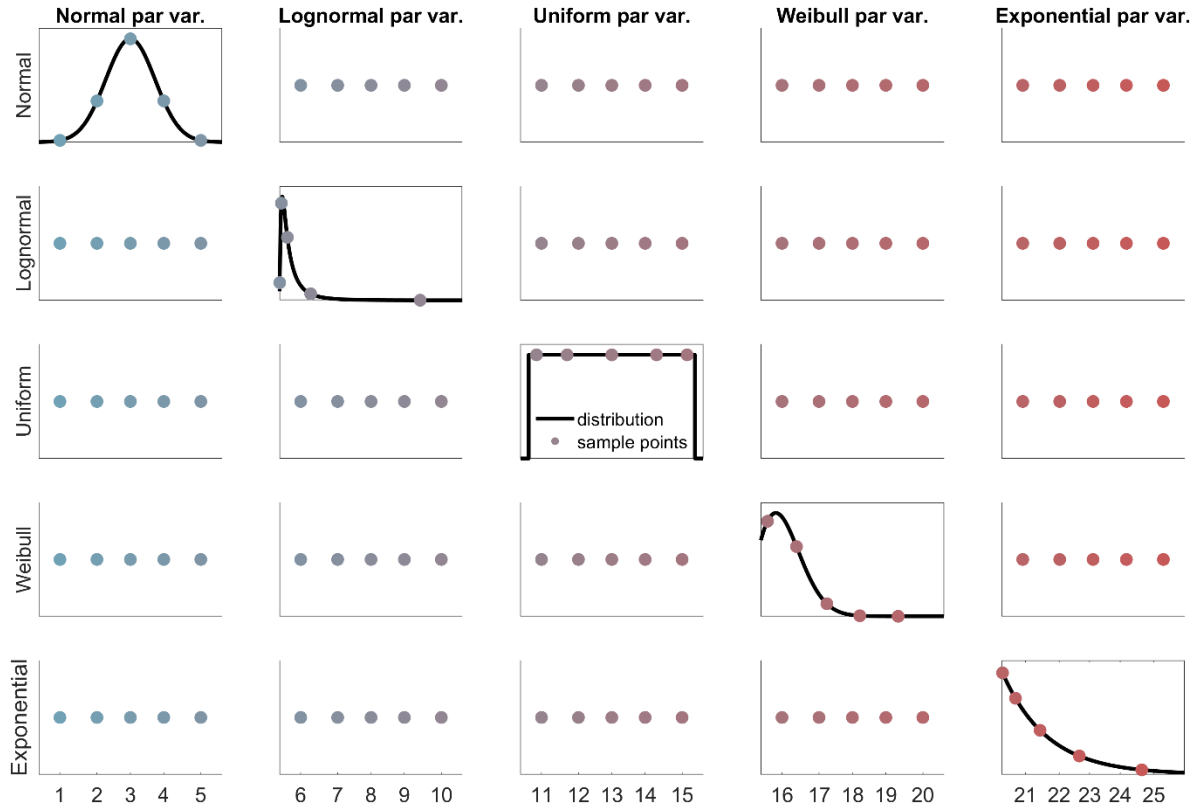


Figure 10. M-DRM evaluations, 5 Gauss points and 5 parameters (Normal, Lognormal, Uniform, Weibull, Exponential) with 5 different distributions (Normal, Lognormal, Uniform, Weibull, Exponential). Rows indicate parameter, and each combination of parameters is indicated by column and color.

Use of method (M-DRM):

- **Paper D.** Used to analyse sensitivity of model to variations in HBM parameters ($n=7$, $N=5$).
- **Paper E.** Used in first part of study to analyse sensitivity of model to variations in environment parameters ($n=8$, $N=3$).

4 ACTIVE HBM MODELLING BACKGROUND

4.1 PHYSIOLOGY

4.1.1 Sensors in humans

Humans can sense changes in the surrounding environment through several sensory organs (Marieb and Hoehn, 2019). Some of the possible sensory inputs a human could use to respond to a vehicle maneuver are presented in the subsections below.

4.1.1.1 *Semicircular canals (vestibular system) – rotational acceleration*

The semicircular canals, located in the inner ear, consist of three orthogonal arced canals (Marieb and Hoehn, 2019). The canals are filled with fluid and gel, and equilibrium receptors are located at one end of the canal. These receptors sense changes in equilibrium that arise during inertia loading during head rotational accelerations. Since the canals are orthogonal, rotations around all axes can be detected.

4.1.1.2 *Otoliths (vestibular system) – linear acceleration, head position relative gravity*

The otoliths, located in the inner ear, consist of one membrane layer with tiny stones on the surface, and one layer of hair cells that sense hair cell deflection (Marieb and Hoehn, 2019). When the head is upright, the hair cells are vertical. If the head is accelerated, the inertia of the stones will cause hair cell deflection, and acceleration is sensed. Because of their alignment, when the head is upright, mainly horizontal plane accelerations can be sensed through the otoliths. Head orientation in the gravity field can be sensed by the otoliths because the gravitational accelerations can be sensed.

4.1.1.3 *Muscle spindles – muscle stretch*

Muscle spindles are located inside skeletal muscles and consist of sensory fibers that wrap around the center of modified (intrafusal) muscle fibers and sensory fibers in the ends of the modified (intrafusal) muscle fibers. The former sense stretching and rate of stretching of the muscle while the latter sense stretching only of the muscle (Marieb and Hoehn, 2019).

4.1.1.4 *Golgi tendon organs – tendon tension*

Golgi tendon organs, located inside tendons, consist of small bundles of tendon fibers with sensors between and in the fibers and sense tendon tension force (Marieb and Hoehn, 2019).

4.1.1.5 *Joint kinesthetic receptors – joint position*

Joint kinesthetic receptors, located inside joint capsules of synovial joints, consist of several receptor types, and sense joint position and joint motion (Marieb and Hoehn, 2019).

4.1.1.6 *Tactile, Lamellar and Bulbous corpuscles – pressure on skin*

Pressure applied on the skin can be sensed through several different types of sensors, which allows for sensation of both lighter and deeper applied pressures (Marieb and Hoehn, 2019). Some of the sensors sense change in pressure while others sense the pressure continuously.

4.1.1.7 *Vision*

In addition to the sensors described above, humans use their vision to navigate and respond to changes in the environment. Visual processing is a complex task which involves several parts of the brain (Marieb and Hoehn, 2019).

4.1.2 Muscle activation

To maintain posture and to control movement, humans use active contraction of skeletal muscles (Marieb and Hoehn, 2019). The central nervous system (CNS) activates skeletal muscles either by voluntary or reflexive contraction. In response to a stimulus, the early response would be dominated by reflexes, while later, the response could be attributed to voluntary control (Scott, 2012, Macefield, 2009, Kurtzer, 2015).

Voluntary control can be used to activate muscles, which is when the muscles are consciously activated to produce a movement or a force (Betts et al., 2013). For instance, in the vehicle environment, this could be when reaching to close a vehicle door or correcting the position of the rear-view mirrors.

Reflexes can be either intrinsic or acquired, and the difference between the two is not always clear (Marieb and Hoehn, 2019). Reflexes are fast responses that do not require conscious control to (for instance) activate muscles (Marieb and Hoehn, 2019). An example of an intrinsic reflex is patellar reflex (knee jerk reflex) that can be tested by striking the patellar tendon with a small hammer, which causes muscle activation to extend the knee (Betts et al., 2013). An acquired reflex in the vehicle environment could be to press the brake pedal suddenly and forcefully in response to a threatening situation, such as the vehicle in front suddenly stopping.

Two reflexes that could be important for how an occupant would recruit their muscles in response to external loading, such as the accelerations in a vehicle maneuver, include the vestibulocollic reflex and stretch reflex. The vestibulocollic reflex (VCR) (Binder et al., 2009b) activates neck muscles upon sensing linear or rotational accelerations of the head and aims at maintaining the head position in space, and/or dampen oscillations during head movement (Goldberg and Cullen, 2011). The stretch reflex (cervicocollic reflex for neck muscles (CCR)) contracts a muscle if a change in length of that muscle is sensed (Binder et al., 2009a) and aims to maintain the original length of the muscle.

4.1.3 Muscle physiology

To activate a muscle, the CNS sends an electrical pulse (action potential) through the motor neurons (Marieb and Hoehn, 2019). When the action potential reaches the neuromuscular junction, it triggers a chemical reaction, which in turn triggers an action potential that travels through the sarcolemma, surrounding the muscle fibers. This second action potential triggers another chemical reaction, which in turn contracts the muscle fibers within a motor unit (a group of muscle fibers innervated by one motor neuron). In response to a single action potential, the tension in the muscle fiber is quickly increased and then more slowly relaxed, Figure 11.

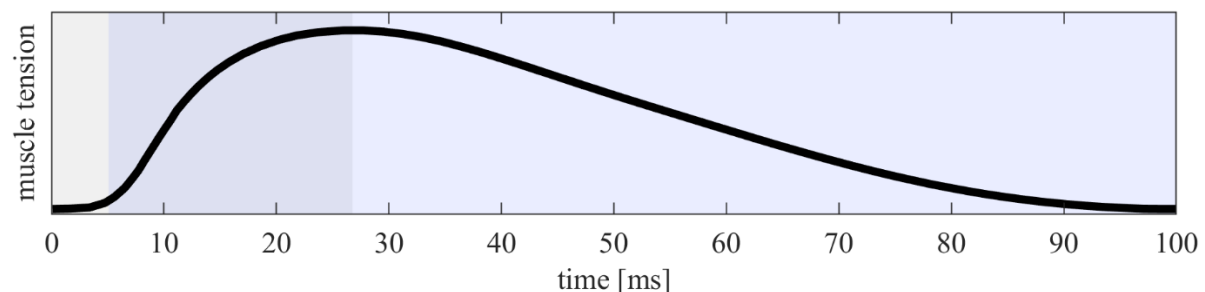


Figure 11. Muscle tension response to a single stimulus (at $t=0$). The process starts with a latent period (grey), before any contraction in the muscle fibre occurs, followed by the contraction period (grey/blue), where tension in the muscle fibre is built. Lastly, in the relaxation period (blue), the tension in the muscle fibre is relaxed back to no tension. Image recreated from Anatomy and Physiology, Betts et al. (Betts et al., 2013) <https://openstax.org/details/books/anatomy-and-physiology>

If another action potential is received before the fibers are relaxed, the muscle fibers contract again increasing fiber tension force (wave summation), Figure 12. If the frequency is high enough, no relaxation occurs (tetanus), and maximum tension within the muscle fibers is achieved, Figure 12.

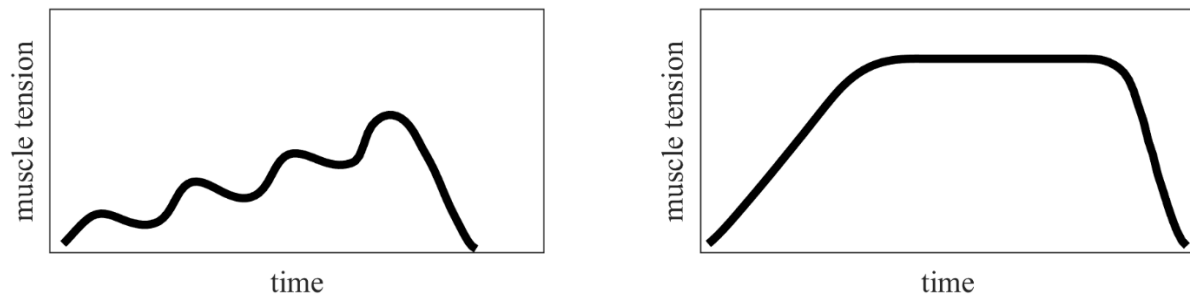


Figure 12. Muscle tension in response to several stimuli. Increasing tension in the motor unit occurs if additional pulses are sent to the muscle before complete relaxation (left figure). No relaxation occurs if the frequency of stimuli is high enough (right figure). Image recreated from *Anatomy and Physiology*, Betts et al. (Betts et al., 2013) <https://openstax.org/details/books/anatomy-and-physiology>

Besides the increased tension in muscle fibers from increased action potential frequency, the strength of a muscle contraction can be increased by activating more motor units. The smaller motor units are activated first, leading to a non-linear relationship between the number of motor units activated and tension force produced (Marieb and Hoehn, 2019).

The electrical activity related to muscle activation can be measured using electromyography (EMG) (Marieb and Hoehn, 2019), either by placing electrodes on the skin or by placing needle electrodes inside the muscle. When measuring activity using EMG, the voltage of the action potentials that propagate in the muscle (and potentially surrounding muscles) are measured. EMG signals are often normalized to a maximum value, either by asking the subject to maximally contract that muscle (maximum voluntary isometric contraction (MVIC)) or by using the maximum recorded activity in the test that is being investigated.

Skeletal muscles produce movement by applying tension forces between bones (or, in some cases, skin) (Betts et al., 2013). In many joints, movement is induced by rotation around the joints, and the muscles produce a moment around the joint, with a magnitude depending on the force in the muscles and the lever arms. The muscle that causes the movement is called an agonist. Synergists are surrounding muscles that assist the agonist, while antagonists are the muscles opposing the movement (Betts et al., 2013). During movement, the antagonists might also co-contract with the agonist and synergists to stabilize the joint (Latash, 2018).

4.2 CONTROL THEORY

Controllers are typically classified as either open-loop or closed-loop (feedback) controllers (Åström and Murray, 2021), Figure 13. Closed-loop controllers use information regarding the state of the system it is controlling, while open-loop controllers do not.

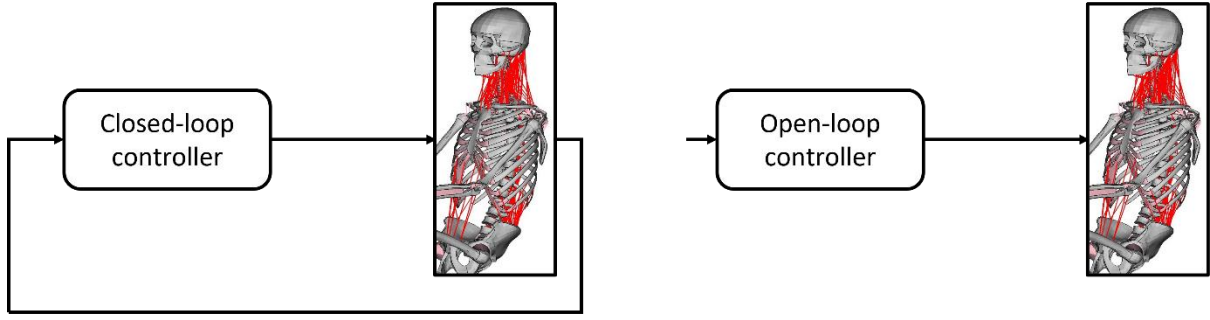


Figure 13. Closed-loop (feedback) controller to the left, and open-loop controller to the right.

A commonly used and relatively simple feedback controller is a Proportional-Integral-Derivative (PID) controller (Åström and Murray, 2021). Given a reference point, the controller responds to the deviation from the reference point, called error ($e(t)$), the accumulation of error, and rate of change in error. The gains of the controller (K_p , K_i , K_d) determine how the controller responds. The proportional part gives a larger response for a larger error, the integral part gives a larger response if the error has been present for longer, and the derivative part gives a larger response if the error is increasing faster. The proportional part can reduce large errors but cannot alone reduce the error down to zero, because as the error decreases, the response also decreases. The integral part can reduce the error down to zero, but will respond slower than proportional control. The derivative part mainly acts to stabilize the system.

$$u(t) = K_p e(t) + K_i \int_0^t e(\tau) d\tau + K_d \frac{de(t)}{dt}$$

4.3 SAFER HBM

The SAFER HBM (Pipkorn et al., 2023), Figure 14, was used throughout this thesis. In Paper A, v9 was used, and in papers C-E, v10 was used. The SAFER HBM, based on THUMS v3, is a finite element HBM developed for simulations of vehicle crashes. The model has been thoroughly validated for impact simulations (Pipkorn et al., 2023), and v9, including selected updates intended for v10, was validated for low-speed frontal impacts (Larsson, 2020). In addition to the passive properties, the model contains active musculature in neck, lumbar, arms and legs, and a previous version of the model was successfully used to simulate drivers in braking (Östh et al., 2015). A detailed description of version updates from v9 to v10 was presented in (Pipkorn et al., 2021), and some of the version updates are presented here. In v9, the torso, arm and forearm were connected only in the joints, while in v10 the upper extremity flesh is modelled with a continuous mesh over the shoulder and

elbow. The pelvis, and subsequently lumbar muscle attachment points were updated between the models.

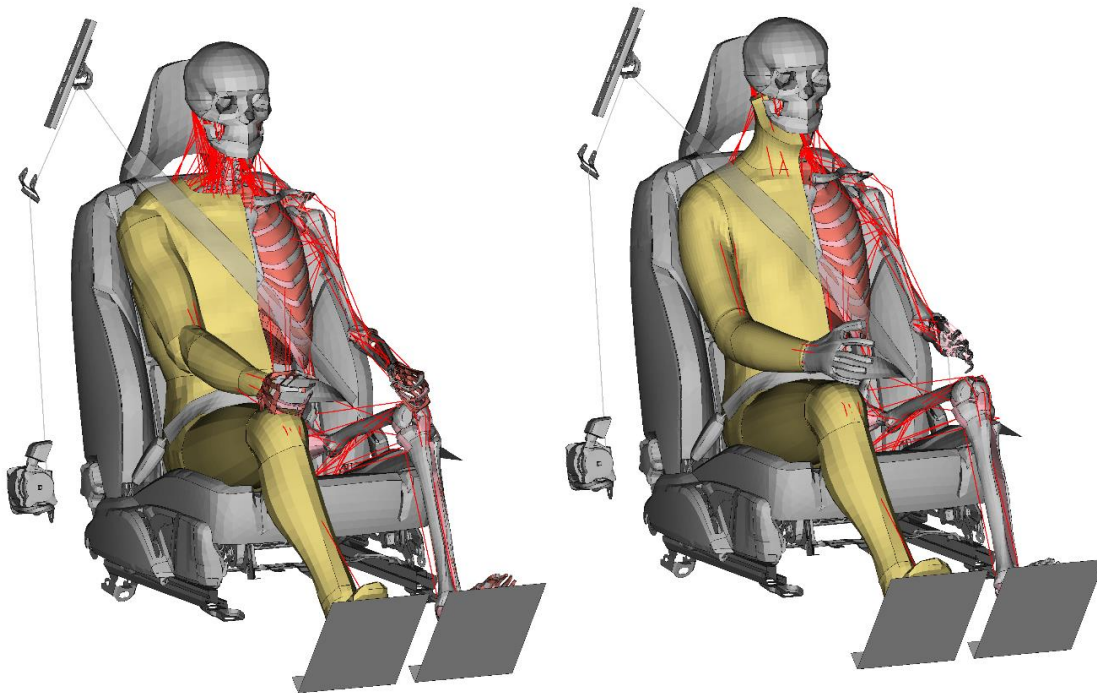


Figure 14. SAFER HBM, v9 (left) and v10 (right), seated as a passenger in a model of a Volvo V60 seat, the same model as used in all simulation papers (A, C-E), corresponding to the seat used in Paper B. Flesh and skin on right side hidden for visibility. Shoulder belt transparent for visibility.

The skin and flesh properties were updated between v9 and v10. In v9, the torso and upper extremity subcutaneous adipose tissue and muscle (flesh) were modelled as a combined entity. In v10, the adipose tissue and muscle were separated, and the material properties updated, Figure 15. Furthermore, the skin modelling has been updated from a linear isotropic material, to a non-linear anisotropic material (Manschot and Brakkee, 1986), with material directions based on skin tension lines (Langer's lines) (McIntosh and Fyfe, 2013). For skin and adipose tissue, the v10 material properties were softer than those in v9, while for the muscle tissue, v10 was slightly stiffer in lower

deformations and substantially stiffer in higher deformation, Figure 15.

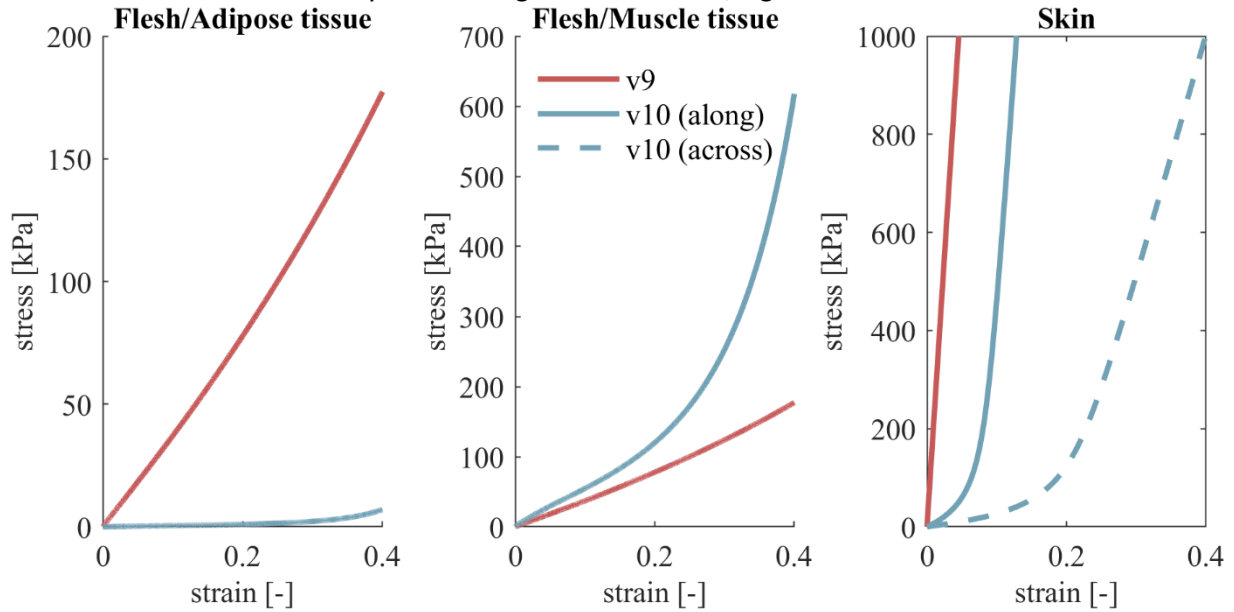


Figure 15. Compression stiffness (compression rate 0.1/s) of adipose and muscle tissue in SAFER HBM v9 (red) and v10 (blue), and skin tensile stiffness in SAFER v9 (red) and v10 (blue). The same stiffness was used for adipose tissue and muscle tissue in v9. The skin was modelled using an anisotropic material model in v10, with different stiffnesses along (solid) and across (dashed) skin tension lines (Langer's lines).

SAFER HBM v9 (Pipkorn et al., 2019, Iraeus and Pipkorn, 2019) included active muscle control in neck, lumbar, upper and lower extremities. For neck, lumbar and upper extremities, muscles were controlled with feedback controllers that respond to angle deviations of defined links in the body (Ólafsdóttir et al., 2019, Östh et al., 2014). For the neck, this link was defined between T1 and head center of gravity, for lumbar between sacrum and T10, shoulder between proximal and distal humerus, and for elbow between proximal humerus, elbow, and wrist, Figure 16. Lower extremity muscles could be activated with pre-recorded muscle activity data. In addition, muscle length feedback could be used to control the neck and lumbar muscles.

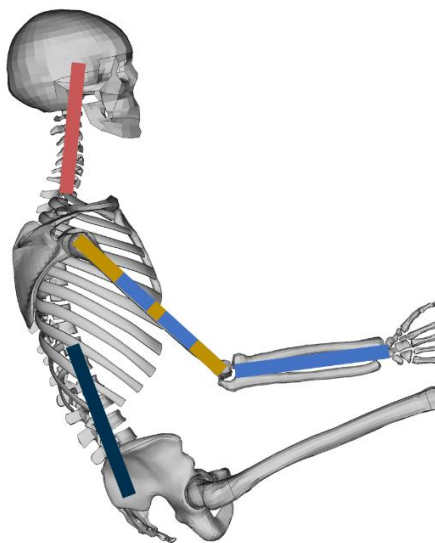


Figure 16. Active muscle controller links defined in SAFER HBM v9. The red line shows the neck link, dark blue line shows the lumbar link, gold dashed line shows shoulder link and light blue lines (one of them behind the gold dashed line) show the two links that define the elbow.

The feedback loop structure for muscle controllers was common for all muscle feedback loops. The angular deviation, with delay, was fed through a PID controller. The PID response was scaled to the individual muscle activities based on defined muscle load sharing, and filtered in activation dynamics, that represents the delay from muscle activation signal to force generation in humans. Finally, the signal was saturated, and baseline activity was added, before applied on the material card of the model. In both neck and lumbar muscle controllers, the PID controller responded to angular displacement of a link between two defined anatomical points, in 3d, and intermuscular load sharing was based on directionally dependent muscle activation from volunteers (spatial tuning) (Ólafsdóttir et al., 2019). Neck and lumbar controllers aimed at maintaining the posture in the global coordinate system. The shoulder controller, used to model drivers, responded to sagittal plane link angular deviations, and muscle load sharing was pre-defined to set values for flexion and extension. The elbow controller, used to model drivers, responded to angular deviations of the elbow joint (2d joint), and muscle load sharing was pre-defined to set values for flexion and extension. The model could predict full-body kinematics in braking for both driver and passenger position, and a previous version of the model has been validated in 1.1 g braking in driver and passenger position (Östh et al., 2015, Östh et al., 2014, Östh et al., 2012).

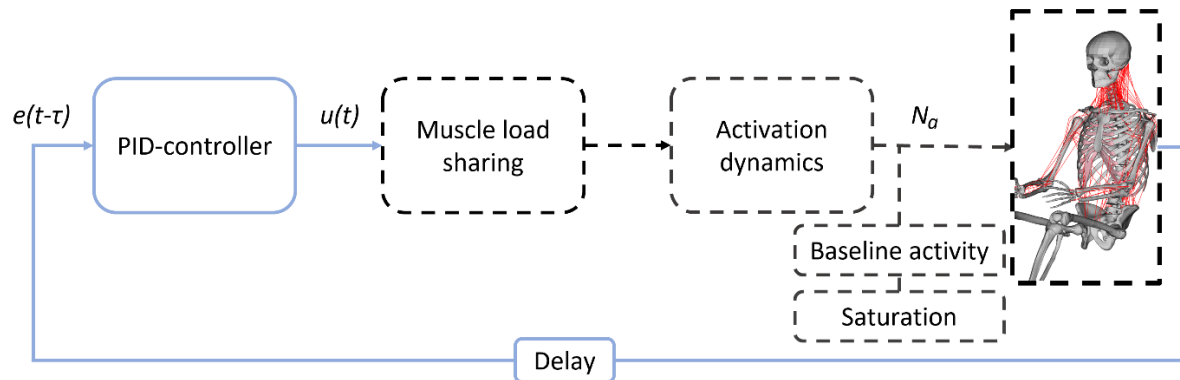


Figure 17. Muscle activation feedback controller. The logic is used for neck, lumbar, and upper extremity controllers.

5 SUMMARY OF PAPERS

Five studies have been performed to meet the two objectives, Figure 18. Paper A and Paper C both relate to objective one, and Papers B, D and E relate to objective two. Papers A, C, D and E were simulation studies, while in Paper B, an analysis of data collected in a volunteer study was conducted.

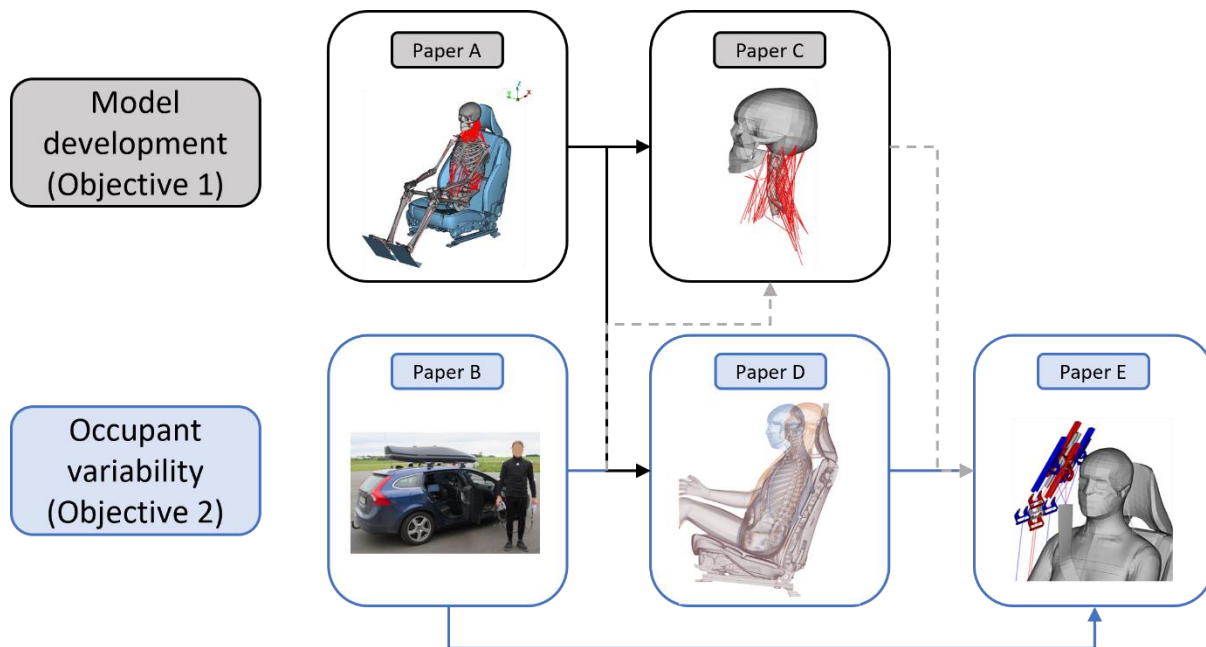


Figure 18. Included papers, how they relate to the objective, and to each other.

5.1 MODEL DEVELOPMENT

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

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

Whereas the original implementation aimed at maintaining the posture in the global reference system, the enhanced models aimed at maintaining a set posture in either 1) a completely local reference system (T1 for neck controller, sacrum for lumbar controller), 2) the vehicle coordinate system, or 3) rotating with the HBM in the horizontal plane.

The three different APF controllers were evaluated in a combined lane change and braking load case. One of the APF configurations was compared to volunteers in braking, lane change, and combined lane change and braking. All three directions were evaluated using two different seatbelt configurations: a regular seatbelt and a belt with an electrical pre-pretensioner, yielding a total of six load cases. The kinematic predictions and muscle activation signals were objectively evaluated using

CORA. The kinematic CORA results ranged from 0.78 to 0.88 for the active models and 0.70 to 0.82 for the passive configuration.

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

In Paper A, it was noted that the model did not predict rotations as well as translations. It was hypothesized that it was because the model responded to translations but not rotations. Thus, in Paper C, a controller that responded to rotations was developed.

In Paper C, a new controller was developed, following the same logic as the APF controller, using a 3D axis-angle representation of head rotations as input to the muscle controller. Intermuscular load sharing was based on rotational direction in the simulation and muscle activity recorded in three volunteer experiments. The gains of the both the new and the APF controller from Paper A tuned to minimize differences between head displacements of the HBM and volunteers in braking and lane change maneuvers (from Paper B) using multi-objective optimizations. Bio-fidelity of the model with tuned controllers was evaluated objectively using CORA. The results indicated comparable performance for rotational and APF controllers after tuning, with somewhat higher bio-fidelity for rotational kinematics with the APF controller. After tuning, good to excellent bio-fidelity was indicated for both controllers, in forward direction in braking, and lateral direction in lane change. For rotational displacements, and translational displacements in the other directions, bio-fidelity ranged from poor to excellent, with somewhat higher overall CORA scores for the HBM with the APF controller in both braking and lane change. The overall CORA scores for head displacements were 0.79-0.85 and 0.90-0.94 in braking and 0.68-0.77 and 0.82 for rotational and APF controllers, respectively, compared to 0.93 and 0.82 for the original model. Overall, the results showed that when tuned, both the rotational and APF controllers can be used to predict the occupant response to an evasive maneuver, with slightly better performance with the APF controller, allowing for the inclusion of evasive maneuvers prior to a crash in evaluation of vehicle safety.

5.2 OCCUPANT VARIABILITY

The first objective of Paper B was to investigate predictors for vehicle passenger kinematics, such as belt configuration, sex, age, stature, and BMI, for different types of vehicle maneuvers. The second objective of Paper B was to create and report kinematic corridors for selected combinations of predictors, to be made available for validation of active HBMs. Principal component analysis and linear mixed models were used on selected data to create predictive models for kinematics and belt time histories, using belt configuration, sex, age, stature, and BMI as co-variates. Monte Carlo simulations of resulting models were used to generate upper and lower response corridor limits around the predicted responses. For translational and rotational displacements of the head and the torso, the first three principal components together captured 91%-99% of the variance in the responses. Belt configuration, sex, age, stature, BMI, and their interaction effects were found statistically significant ($p < 0.05$) in the linear mixed model analysis in lane changes, braking and U-turns at 40 km/h but not in U-turns at 30 km/h or when aware of turn. Response corridors for average sex, stature and BMI, were provided. In conclusion, the models and data provided can be used for validation of human body models with a range of anthropometries and in different maneuvers and belt configurations potentially occurring in pre-crash maneuvers. Although all

occupant characteristics were predictors of at least one of the PC scores, the explanatory power was small.

Since the occupant characteristics (sex, age, stature, BM) explained very little of the variation in Paper B, other characteristics were investigated through simulations in Paper D, and subsequently in Paper E. In Paper D, sensitivity of the HBM in braking to human characteristics (spinal alignment (PC1 and PC2), muscle physical cross-sectional area (PCSA), neural delay, fat stiffness, muscle stiffness, and skin stiffness) were investigated through M-DMR sensitivity analysis. The results indicated that the most influential of these characteristics were spinal alignment (PC1 and PC2), and muscle PCSA. In the first step in Paper E, M-DRM sensitivity analysis was used to identify the most important characteristics of the test environment. Belt position at belt locking, belt stiffness, acceleration shape (PC 1), velocity change, seat position, arm to thigh constraint force, seat to HBM friction and D-ring vertical position were varied, and results indicated that seat position, velocity change and belt stiffness were the most influential test vehicle characteristics. In the second step in Paper E, a synthetic experiment was conducted, where the three most influential characteristics for the HBM (Paper D) and test environment (first step, Paper E) were randomly varied using Latin Hypercube sampling. Based on the results from the second part in Paper E, the most important characteristics were seat position and occupant posture. A more forward seat position reduced the peak forward displacements and the time to peak, as did a more curved posture. In the synthetic experiments, these two parameters accounted for 70%-80% of variance in peak displacements, and 60% in time to peak.

6 DISCUSSION

The motivation for this work was to aid the design of safer vehicles in a way that includes occupant diversity. AHBM can aid the design of safer vehicles by predicting the occupant posture, kinematics and muscle forces when transitioning from an evasive maneuver into a crash. This can be used to either develop safety systems that prevent the occupant from moving from the optimal position during a maneuver or by enabling evaluation of passive safety systems for an occupant in a realistic non-standard position. The understanding of occupant variability could be used to guide the developers in deciding which occupant characteristics to include in the development work. For instance, while occupant diversity typically focuses on sex, body size and age, these characteristics predict very little of kinematics in a maneuver, Figure 3, and thus occupant diversity in the maneuver context could include other characteristics instead. Based on the results from this thesis, a possible candidate for such a characteristic is spinal curvature.

A posture where the occupant is leaning out of the seat belt prior to a frontal impact has been shown to induce larger crash kinematics (Donlon et al., 2020). For near-side and far-side impacts, a posture towards the opposite side of the crash induced larger displacements during crash (Leledakis et al., 2021b). In (Leledakis et al., 2021b), a forward leaning posture, similar to a posture after braking, was found to induce larger displacement in lateral impacts, due to less engagement with the side bolsters. A forward posture was shown in (Leledakis et al., 2021b) to increase the crash kinematics in frontal impacts, while in (Boyle et al., 2020), a forward posture was found to reduce the risk of neck and cervical spine injuries, but increase chest deflection. This information, together with AHBM can be used to tune future restraint systems such that an appropriate level of belt pre-tensioning is applied depending on type of maneuver, and potentially also the type of expected crash. For instance, possibly a higher degree of pre-tensioning is required for braking followed by a lateral impact to retain the occupant closer to the side bolsters, while prior to a frontal impact the pre-tensioning force during braking could be slightly lower, since a more forward posture was not found to exclusively increase predicted injury risks (Boyle et al., 2020), however, it increased crash kinematics and kinetics in another study (Leledakis et al., 2021b). Further, for maneuvers that include a lateral component, restraints could be tuned such that they reduce both forward and lateral displacements during maneuvering.

Since the method used in Paper E managed to identify the influence from geometrical changes to the vehicle environment, opportunities to utilize the method with active SAFER HBM to understand the influence on occupant kinematics from geometrical changes. Physical tests have the advantage that they typically capture what happens in the physical world, and this makes the results from physical tests the benchmark. However, the physical tests have drawbacks as well. For instance, boundary conditions are not so easily defined or controlled, and measurement equipment can often influence the results. For example, the instrumentation and measurement systems in Paper B likely made the experiments less naturalistic for the volunteers. The subjects were also exposed to multiple tests, which have been shown to reduce their displacements, Figure 3. In physical testing, the number parameters that can be measured are limited, since each additional parameter requires additional measurement equipment. Furthermore, all measurements include some uncertainty. The limitations and uncertainties in simulations are different than those in physical testing. In simulations, boundary conditions can be precisely controlled, measurements do not influence results, and no additional equipment is needed for additional measurements. Identical muscle activation strategies can be replicated across multiple simulations and maneuvers. The precision of measurements is high, and simulations can be parallelized easily. The drawback of the models is

instead the accuracy and uncertainty of the model itself, requiring validation of the model from detailed sub-system validation to full-scale validation prior to use. By combining both simulations and physical tests, the impact of the respective limitations can be understood and often reduced, allowing further insight into the real-world physical behavior.

The AHBM is intended to be used to simulate an evasive maneuver that is followed by a crash. In the volunteer tests used in this study, no threat of collision was present, and thus the scenario might not be representative of what it is intended to simulate. Although there are ethical concerns of adding a crash element to a volunteer study, there are some ways of introducing a perceived threat to make the scenario more realistic. For instance, in a recent in-vehicle test, a balloon car was used to simulate a crash situation (Roka et al., 2023). The participants were riding as passengers in a vehicle that was following another vehicle (lead vehicle) around a test track. The lead vehicle suddenly changed lane to reveal a stationary balloon car, and the test vehicle braked to avoid a crash with the balloon car. Another approach could be to introduce a crash in driving simulators, as has been done previously (Hault-Dubrule et al., 2011), although occupant kinematics were not the focus in that study. Driving simulators have been used to study occupant kinematics in other studies (Kempter et al., 2022), and thus it should be possible to introduce a threat in a driver simulator study and to record occupant kinematics. Potentially, the volunteers from those studies represent a more realistic behavior of occupants in a critical situation. Although, if the occupant is not aware of any threat, and the car is performing the evasive maneuver autonomously, it is possible that the volunteers where no threat was introduced are more representative.

6.1 METHOD CHOICES

The active HBMs have several applications, where one of the main uses is to predict the occupant state when transitioning from an evasive maneuver to a crash. As indicated in the introduction it is important to predict position, posture, and muscle activity. The most straightforward way to ensure that all relevant parameters are included in the crash simulation is to run Whole-Sequence simulations, as was done in (Östh et al., 2022, Wass et al., 2022), and with another HBM in (González-García et al., 2021). While the SAFER HBM is intended to run seamless, so internal stresses and strains, positions, velocities, and muscle activity state are carried over from the maneuver to the crash, studies have been made to consider the evasive maneuver in other ways. For instance, postures from volunteer experiments were used to position the HBM prior to crash simulation (Boyle et al., 2020). This allows investigations of effects from posture and position, while any effects from changed muscle activity are omitted. Further, this approach allows for the creation of a library of postures that could be included with morphing instead of with simulations of maneuvers. Using that approach, AHBM is not needed at all. (Wehrmeyer, 2020) found that the occupant posture/position was the most influential in crash. However, including for instance occupant velocity relative to vehicle, and internal stresses of the occupant when transitioning from a maneuver to a crash yielded closer similarity to a maneuver and crash which was run seamless. This indicates that for accuracy in simulations of crashes preceded by a maneuver, the whole sequence should be considered. However, a fair approximation can be obtained by including only the posture/position, which can be enough depending on what the target with the crash simulation is. As indicated by (Wehrmeyer, 2020), a fair representation of the crash can be obtained if both positions and velocities from one HBM are applied to another, as was done by (Yamada et al., 2016, Matsuda et al., 2018, Dominik Breitfuß, 2022, Öztürk et al., 2019). This also reduces the simulation cost if several crash simulations should be preceded by the exact same maneuver, because the results can be reused. Additionally, by using two different HBMs for simulations of pre-crash and in-crash, as can be done with a transfer of results between two different simulations, different stiffnesses for

maneuver and crash HBMs could be utilized. Commonly, in-crash HBMs are stiffer than humans when subjected to low loading, such as an evasive maneuver (Shelat et al., 2016). Thus, by using the Whole-Sequence approach, larger requirements are placed on the HBM, because it needs to be sufficiently soft to model the maneuver, but sufficiently stiff to be numerically robust in the crash simulation. The stiffness of the SAFER HBM v9 was investigated in (Larsson, 2020), where the original SAFER HBM v9 was found to be stiffer than post-mortem human subjects in low-speed frontal impacts. With updates that softened the flesh and skin, predictions of displacements were improved. These updates to flesh and skin were introduced to the SAFER HBM v10 in (Pipkorn et al., 2021). The low-speed frontal impact validation was rerun with v10 in Paper D, and the model displaced similarly as the validation data when moving forward but rebounded more than the validation data after reaching peak displacement.

Data from several volunteer test series have been used in model comparison in this thesis (papers A, C-E), braking from Ólafsdóttir et al. (Ólafsdóttir et al., 2013), and braking, lane change, and their combinations from Ghaffari et al. (Ghaffari et al., 2018, Ghaffari et al., 2019), parts of which were presented in Paper B. In both test series, the volunteers were exposed to several subsequent maneuvers. In (Ólafsdóttir et al., 2013), the volunteers were exposed to 29 maneuvers, of which 9 were in the passenger seat. In (Ghaffari et al., 2018, Ghaffari et al., 2019) and Paper B, the volunteers were exposed to a minimum of 54 tests, of which a minimum of 33 were in the passenger seat. As indicated by several studies previously (Kirschbichler et al., 2014, Reed et al., 2021, Reed et al., 2018, Chan et al., 2022), Figure 3, including the repeated tests would have most likely led to smaller displacements compared to if the volunteers had only been exposed to one maneuver. Additionally, the volunteers performed MVIC tests prior to in-vehicle testing, further preparing them for the tests. When using the HBM to predict kinematics, it is most likely an unprepared occupant that is of interest. Ideally, the first exposure should have been used in all model comparisons, which was not possible with the chosen comparison data, that is partly presented in this thesis. In Paper B, it would have been possible to include exposure in the regression models. However, exposure was omitted, because of the relatively few numbers of first exposure events compared to the number of tests performed (1 out of 54).

In-vehicle tests have been used exclusively in model comparisons (Paper A, Paper C-E). The test data was used because they represent scenarios and environments close to the intended use, but as indicated by comparing previous work, Figure 4, this could mean that the variability was larger in the studies used in model evaluations, compared to test in a more controlled sled environment. The variability in in-vehicle tests could possibly be attributed to variability in the environment, which was explored in Paper E. Since part of the variability possibly could be attributed to the environment, using in-vehicle tests requires extensively validated environment models. In this thesis, one seat model was used for all simulations where a seat was included. The seat model had been validated with quasi-static indentation tests prior to use (Östh et al., 2012). However, some in-vehicle variability could come from occupants responding to the environment, without being directly linked to variability in environment. For instance, the occupants responding to lane change in (Huber et al., 2015, Huber et al., 2014, Kirschbichler et al., 2014) displaced more when the head moved towards the center of the vehicle compared to when the head moved towards the passenger side window. The same seat and seat belt was used in each tested condition. Although the seat belt is asymmetrical, the difference could also be driven by the occupant activating their muscles differently to avoid interactions with the vehicle interior. A similar hypothesis was proposed in (Reed et al., 2021), where outboard excursions were larger in a larger vehicle, potentially because the subjects restricted their outboard movement more in the smaller vehicle to avoid contacting the vehicle interior. Because the intended use is simulations of in-vehicle environments, as in (Östh et

al., 2022, Wass et al., 2022), comparing the model to in-vehicle tests was prioritized, even though this introduces additional uncertainties. Future work could include validating the model in a more controlled sled environment in addition to the less controlled environments used in this thesis.

To allow for regression of time series in Paper B, principal component analysis was used, as has been done previously (Samuels et al., 2016, Ghaffari, 2021, Reed et al., 2021). The regression models were used to predict one value (PC score) based on the selected predictors (belt, sex, stature, age and BMI, and their interactions). Other methods have been used for parametrization of time series. For instance, in (Reed et al., 2021), cubic splines were used to parametrize the displacements, prior to the use of PCA. 12 parameters were used to describe the cubic splines, and it should be possible to use regression on these 12 parameters without using PCA. By combining the splines with PCA these 12 parameters could be reduced further, since the PCA identifies the most important variability, and vary the spline parameters together. Looking beyond evasive maneuvers, other curve parametrization methods have been used. For instance, for force-deflection curves, a characteristic average curve could be created using normalization and interpolation over monotonously increasing and decreasing windows (Lessley et al., 2004). It should be possible to use regression on peak values used for normalization to create characteristic non-average curves. However, the kinematics in Paper B was not easily divided into distinct monotonously increasing and decreasing windows, making such an approach unfeasible for the data in Paper B.

The displacement corridors created in Paper B were created by combining results for males and females, and two different belt systems wherever available, and then accounting for the differences by regression. Kinematics were presented separately for males and females in Ólafsdóttir, Östh et al. (Ólafsdóttir et al., 2013, Östh et al., 2013), Chan et al. (Chan et al., 2022, Chan et al., 2021), while they were presented together in Graci et al. 2022 (Graci et al., 2022), Huber, Kirschbichler et al. (Kirschbichler et al., 2014, Huber et al., 2013, Huber et al., 2015, Huber et al., 2014) Kempter et al. (Kempter et al., 2018, Kempter et al., 2022), and Reed et al. 2018 and 2021 (Reed et al., 2018, Reed et al., 2021). In Paper B, the displacements were similar enough between males and females, and belt systems, to not distinguish the different groups by visual comparison. For belt forces and belt position, it was possible to distinguish between a standard, inertia-reel belt and a pre-pretensioned belt with visual inspection, because two distinct groups were formed. Thus, the two conditions were analyzed separately for belt forces, but not for displacements. Visual inspection was used prior to PCA in Paper B to avoid a situation as in Figure 7, where two distinctly different groups are analyzed together and result in a less meaningful PCA. However, with this it is possible that analyzing the belt systems together, even though they were visually similar, nuances between the groups could have been masked. Using the same reasoning as for the belt forces, the displacements and belt forces from different maneuvers were analyzed separately, instead of combining all maneuvers into one common analysis and using the regression models to differentiate between them.

In Paper C, neck muscle controller gain tuning was performed using a sub-model consisting of head-neck only, and volunteer torso kinematics from braking and lane change were applied to T1 of the sub-model, similar as was done by (Putra et al., 2021). Prior to tuning, a comparison was made between a full HBM and an isolated head-neck model. The results from the isolated head-neck agreed well with the results from the full model. However, the gains have been tuned using boundary conditions from volunteers and not from a simulation model. If the HBM would fail to predict the torso movements, a model tuned using model data could compensate for poor torso predictions in the neck muscle controller, and possibly produce better head kinematic predictions compared to a model tuned using volunteer data. Although better predictions are of course preferable, compensating for poor performance in one part with over- or underpredicted muscle

forces in another part of the model is undesirable. Therefore, volunteer kinematics were used as boundary conditions in Paper C, even though this might have led to slightly worse head kinematic predictions when combined with a torso model that has not been tuned. Based on results from Paper E, the HBM torso forward displacements were similar to volunteer torso forward displacements, although no objective rating was performed in that study.

In Paper D and E, the M-DRM method (Zhang and Pandey, 2014) was used to analyze the sensitivity of displacements in braking to variations of HBM characteristics (Paper D) and boundary conditions (Paper E). M-DRM was used instead of Monte Carlo based methods because the simulations of braking with the SAFER HBM are relatively costly (in Paper D, each simulation took approximately 65 hours on 32 CPU cores), making thousands of evaluations unfeasible. Because more parameters were investigated in Paper E (8 parameters) compared to in Paper D (7 parameters), the number of Gauss points was reduced from 5 in Paper D to 3 in Paper E. No investigation of convergence was performed. The method was used to screen among the investigated parameters, and because the method is approximative, and no convergence was investigated, the results should be seen as indicative. Other effective methods for screening could have been used as well, such as the Morris method, where one parameter at a time is changed (Saltelli et al., 2008). With the Morris method, $N(n+1)$ evaluations are needed, where N is the number of changes (suggested to be at least 5 (Confalonieri et al., 2010)) and n the number of trajectories (number of elementary effects that can be calculated). With this method, the number of evaluations needed to capture elementary effects are also reduced compared to using Monte Carlo based methods. The M-DRM was used in this thesis because of needing fewer evaluations compared to the Morris method.

The synthetic population in Paper E was created by Latin Hypercube sampling of six parameters, three HBM characteristics (spinal alignment PC and PC2, and muscle PCSA), and three boundary conditions (set position, belt retractor stiffness and velocity change). 20 evaluation point were used for the synthetic population, because this was a typical sample size in volunteer testing, Figure 2. However, each volunteer is typically exposed to several repetitions of the same maneuver, and thus the number of tests included in Paper E was lower compared to in the volunteer tests. However, in (Reed et al., 2018) correlation between displacements from the same occupant was found, indicating that rerunning a test with the same volunteer might not add as much new information as running the same test with a new volunteer.

In all simulation papers (A, C-E) the HBMs were gravity settled, with varying durations, to position the HBM in the seat and remove belt slack, and to allow the model to find a relatively stable position before setting reference position. Head center of gravity and T1 (full-body simulations only) were constrained in lateral and longitudinal direction, but not vertical direction prior to setting reference position, while after setting reference time the posture was maintained by the muscle controllers. In papers A and C, for full body model, a gravity settling of 750 ms was used. In Paper A, this was done to allow for a separate reference time for APF and MLF controllers, and in Paper C for consistency between paper A and C. In tuning simulations with an isolated head-neck sub-system, the gravity settling duration was reduced to 250 ms, because no belt or seat was used. In Paper D, 400 ms was used, while it was increased slightly to 500 ms in Paper F to allow the belt a longer slack removal time. A recent study found that when using gravity settling to position an HBM in a seat, a minimum of 400 ms gravity settling time should be used (Kleeck et al., 2023). In that study a different HBM and a different seat was used, but their results still indicate that all full-body HBMs in this thesis should have been reasonably settled in the seat. Further, in Paper D and E, the posture was found to influence kinematics, and since gravity settling without posture constraints was used to position the models in the seat in papers A, C-E, the posture was not strictly defined in the simulations. Thus, the

gravity settling procedure could have influenced the kinematics slightly, compared to if other positioning methods with more strict posture control had been used.

6.2 MODEL DEVELOPMENT

In Paper C, a new controller was developed, that responded to head rotations instead of head translations. The controller combined rotations in 3D to determine muscle activation and muscle load sharing, allowing for one PID controller to respond to rotations around three axes. This combined approach contrasts with what others have done previously. For instance, the THUMS (Kato et al., 2018, Kato et al., 2017) and GHBMC (Devane et al., 2022) both have three controllers for head rotations, and the resulting muscle activity from each of the three controllers are superimposed. If the same approach had been used in Paper C, the controller might have performed better in terms of kinematic predictions, because the three controllers could have been tuned with separate gains. The combined approach was used to allow for antagonist activity and muscle synergies based on data collected from human volunteers, because human intermuscular load sharing cannot be determined by only considering the muscle geometry (Fice et al., 2018, Vasavada et al., 2002). An alternative approach could have been to optimize load sharing patterns based on the model, and for instance combine force generation in the model and (metabolic) cost minimization (de Bruijn et al., 2016). Since there was enough volunteer data available for both controllers, these volunteer data were used instead of deriving patterns based on the model geometry, to ensure that the patterns were based on muscle activity from humans.

In Paper A, the APF controllers, aimed at emulating VCR, yielded better results compared to the MLF controller, aimed at emulating muscle stretch reflexes. Further, the APF implementations more analogous to VCR yielded better results compared to the APF implementations more analogous to muscle stretch reflexes. For that reason, the subsequent studies (papers C-E) used controllers aimed at emulating VCR. Since the model in Paper A used 1D muscle elements without routing, it is possible that the muscle elements did not lengthen in a similar way as they would in a human. In Paper D, routing of posterior lumbar and cervical spine muscles was introduced, by using several 1D elements in series, and connecting the elements using pulley elements attached to vertebrae and ribs, Figure 19. Using this type of routing could be comparable in terms of line of action to combining passive 3D elements and active 1D elements, as has been used in for instance GHBMC (Barker and Cronin, 2021), as long as the muscles remain close to the bony structure it is attached to. Potentially, this routing would make the lengthening of muscles more physiological, however, no investigation of MLF kinematic predictions after routing was done.

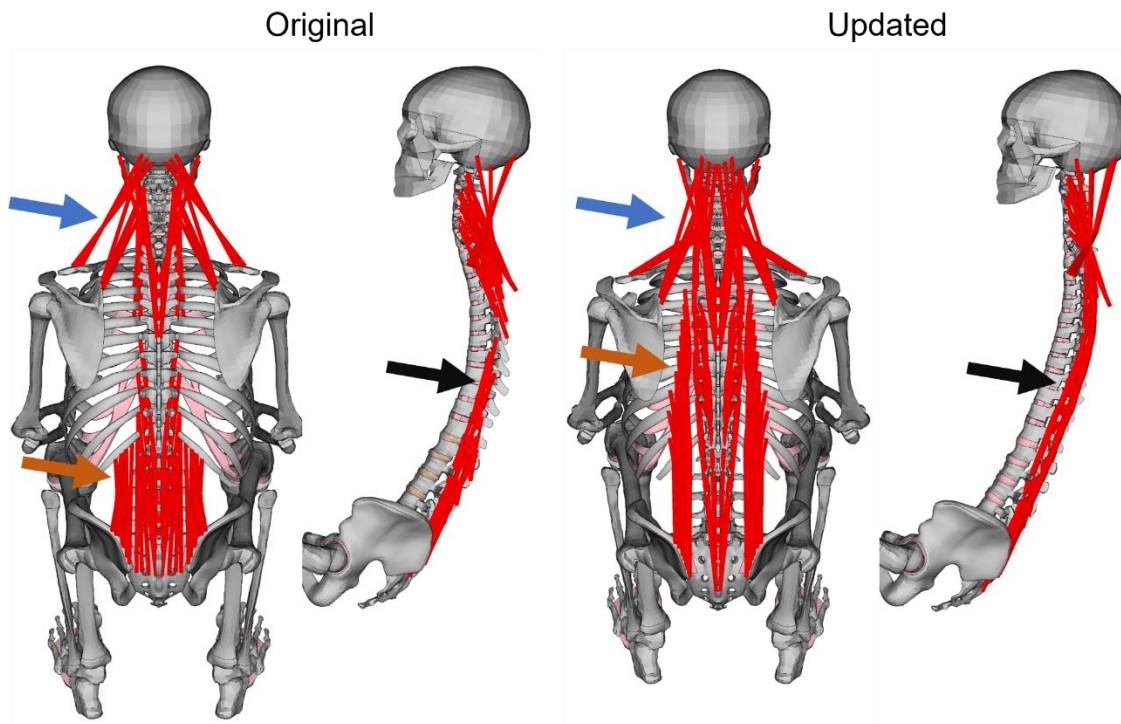


Figure 19. Muscle routing from v10, and updated muscle routing in Paper D. The blue arrow highlights the updated routing for trapezius muscle elements, the orange arrow highlights the updated erector spinae iliocostalis lumborum muscle elements, and the black arrow highlights the updated routing for erector spinae iliocostalis lumborum and erector spinae longissimus thoracis.

Humans can use several types of input to sense a vehicle maneuver, as described in Section 4.1.1, such as head acceleration, head orientation in gravity, muscle length, joint position, skin pressure, tendon force, and vision. Any controller that uses similar inputs could be argued to have some physiological basis, because a human could sense and respond to changes in any of these sensors. Analogies to these different sensory inputs have been implemented in several HBM muscle controllers, Table 2.

Table 2. Analogies to physiological sensors used in HBMs.

Sensors	Models
Head rotation and translation	SAFER HBM v9 (Ólafsdóttir et al., 2019) Paper A GHBMC (Devane et al., 2022, Devane et al., 2019) GHBMC (Correia et al., 2021) THUMS (Kato et al., 2017, Kato et al., 2018) MADYMO (Meijer et al., 2012, Meijer et al., 2013) detailed head-neck model (Happee et al., 2017)
Muscle stretch	SAFER HBM v9 (Ólafsdóttir et al., 2019) Paper A GHBMC (Correia et al., 2021) A-THUMS-D (Martynenko et al., 2019) detailed head-neck model (Happee et al., 2017)
Tendon tension	

Joint position	MADYMO model (Meijer et al., 2012, Meijer et al., 2013)
Skin pressure	THUMS (Kato et al., 2017, Kato et al., 2018)
Vision	

Although the human could sense these inputs, the input itself says little about how the human would respond to the input. Thus, to model the response, some assumptions need to be made. In the SAFER HBM controllers, the assumption has been that the initial posture should be maintained. The posture maintenance has been implemented assuming physiologically based short term reflexes that exist in humans (Section 4.1.2) extrapolated to longer durations. Most controllers are implemented to maintain posture, either the initial posture or some other posture defined to represent for instance bracing. In humans, VCR together with muscle stretch reflexes have been suggested to be important for posture maintenance (Goldberg and Cullen, 2011).

Vision has been identified to be important in sensing vehicle accelerations (Kenney et al., 2020), together with vestibular input, but neither of the implemented controllers in any of the models have modelled vision control, possibly due to the complexity of vision processing. Instead, all controllers, in this thesis and other available models, represent simplifications of human control systems, which integrate input from all available sensors. For instance, when the neck is flexed in a braking maneuver, humans could sense this flexion with otoliths (change in gravitational alignment and linear acceleration), semicircular canals (head rotational acceleration), muscle lengthening in extensor muscles, joint position changes in cervical spine joints, through vision, and possibly also in changes in tendon force.

Under the assumption of posture maintenance, the response to different sensory inputs would be similar regardless of which sensory input is used. For instance, the head rotations would be counteracted with extensor activity and the lengthening of extensors would be counteracted extensor activity. The joint position sensors would also trigger extensor activity. The only contrasting input would be tendon tension and tendon tension reflexes, which de-activate muscles if the tendon force is too large (Marieb and Hoehn, 2019). The tendon tension reflex is not implemented in any of the controllers.

6.2.1 Reflexes

All controllers (Paper A, Paper C-E) were aimed at emulating the VCR, which act on a short time scale (<100 ms) (Kurtzer, 2015). In the controllers, the same controllers were used throughout the simulations, such that these reflexes were extrapolated to relatively long-duration maneuvers (>2 s). In humans, these long time scales (>100 ms) would invoke voluntary muscle activation (Kurtzer, 2015). In Paper E, the corridors were more similar early in the maneuver compared to later in the maneuver, when corridors for volunteers became wider while synthetic experiments corridor remained at the same width. Volitional control could explain why volunteer corridors widen while simulation corridors remain at a similar width throughout the maneuver. This suggests that the initial response from volunteers was dominated by a reflex response, similar for both repetitions and different volunteers, while further into the maneuver, the volunteers adopted different voluntary strategies, throughout the maneuver, between the repetitions of the maneuver and between the different volunteers. In contrast, the model employed the same reflex-based control strategy for the entire duration and for all repetitions. Furthermore, a reflex response would be expected from an unprepared occupant. If the occupant is alerted of the maneuver in some way, the occupant will be prepared for the maneuver. If the occupant is prepared, the implemented controllers might not be representative of the occupant response, as differences between prepared or braced or aware and

unprepared occupants have been found in several experiments (Kirschbichler et al., 2014, Beeman et al., 2011, Chan et al., 2022, Van Rooij et al., 2013a, Van Rooij et al., 2013b). It has been shown that during crash avoidance maneuvers or driver-initiated braking, drivers brace themselves against the steering wheel (Hault-Dubrulle et al., 2011, Östh et al., 2013). As proposed by Östh et al. (Östh et al., 2014), it would be viable to include the bracing by adapting the reference posture, an approach that might be feasible with the new controllers as well.

The neck and lumbar muscle controllers (referred to as APF and used in Paper A, Paper C-E) respond to translations of the head relative the torso or torso relative the pelvis, and was created to emulate the VCR (Ólafsdóttir et al., 2019). The rotational controller (Paper C) was also created to emulate this reflex. The VCR responds to translational and rotational accelerations of the head and predominantly affects the neck muscles (Binder et al., 2009b), and is sometimes referred to "head-in-space" control. The original APF controller implementation (Ólafsdóttir et al., 2019) maintained the posture of the head relative to the torso (or torso relative pelvis) in the global coordinate system and responded input based on linear translations of the head relative torso (or torso relative pelvis). This implementation let the controller respond to similar types of input as the human would. In Paper A, the reference coordinate system was updated to allow for simulations of cases with large vehicle yaw rotations, such as U-turns. It was concluded that a hybrid reference system, partially connected to the global coordinate system and partially connected to the HBM, was the most suitable for the neck and lumbar muscle controllers. Updating the reference system was a pragmatic approach to allow for simulation of more complex scenarios but moved the controller one step away from the reflex it was intended to emulate. Further, it was concluded that the controller predicted some rotational displacements with poor accuracy, hypothesized to relate to that the controller responded to translations but not rotations of the head. Based on results from Paper A, a hybrid system of the APF was introduced in the SAFER HBM in (Pipkorn et al., 2021), and tuned in Paper C. A controller that responded to rotations instead of translations was developed and tuned in Paper C. The rotational controller developed also in Paper C maintained the head in the global coordinate system, similar as the original implementation of the APF controller (Ólafsdóttir et al., 2019).

The best results in Paper A were obtained with a single controller, closest in resemblance to the VCR, agreeing with findings in simulations of rear impacts (Putra et al., 2019). In the volunteer tests however, the volunteers likely responded to combination of several types of input. For instance, humans sense acceleration faster if it is presented using visual cues and vestibular cues together, compared to if one of the cues is presented before the other (Kenney et al., 2020). Since the volunteers in Paper B and (Ólafsdóttir et al., 2013) had their eyes open during the experiments, they likely used both visual and vestibular input to respond to the maneuvers. Furthermore, most likely other input is integrated as well to determine the response of the volunteers. For that reason, input from muscle spindles have been used to activate muscles in HBMs, but combining this input to input from the vestibular system was found not to improve predictions of kinematics in Paper A or (Putra et al., 2019). In contrast to the findings in Paper A and (Putra et al., 2019), (Correia et al., 2021) used a combined MLF and APF controller for the neck muscles in simulations of frontal and lateral impacts. In that study, no comparison between pure APF, pure MLF and combinations was included. For the elbow joint, (Devane and Gayzik, 2023) found that the best kinematic predictions were obtained when combining MLF and APF. Since the elbow joint does not have a vestibular system, it is not surprising that their study found a different control strategy provided better predictions.

HBM head displacements were compared to those of the volunteers, in Paper A, revealing that combining APF and MLF controllers lowered the correlation to volunteer kinematics, although these changes were small. Based on those results, the MLF controller was deemed superfluous. However,

as shown in (Putra et al., 2019), (Ólafsdóttir et al., 2019), and (Happee et al., 2017), an MLF controller used for the cervical muscles prevents vertebral rotation and spine buckling. Besides controlling the initial curvature in Paper D and E, the spinal curvature during simulation was not evaluated, and potentially the spine could exhibit non-physiological vertebral rotations in the simulations. During further model development and validation, it would be of importance to include evaluation of the spinal curvature to ensure that non-physical vertebral rotations are avoided.

6.2.2 Feedback controller

In the feedback loops in all simulation studies (Papers A, C-E), there were several components: PID controller, saturation, spatial tuning, activation dynamics, and baseline activity. Although all components are important for the controller, the sequence in which they should occur was less straightforward. The order was updated between Papers A and D and Papers C and E, Figure 20. In Papers A and D, baseline activity was placed last and was used if the control signal was below the baseline value, while in Papers C and E, the baseline activity was added on top of the control signal and was placed before the activation dynamics to prevent discontinuities in the signal. In Paper A, the saturation was placed last, while in Paper C, the saturation was placed before spatial tuning to ensure that the muscles always maintained the load sharing based on the volunteer data. This order differs slightly from for instance the MADYMO model (Meijer et al., 2012, Meijer et al., 2013), where the activation dynamics (for the MADYMO modeled as a frequency dependent delay) was placed before the muscle recruitment (comparable to spatial tuning). The THUMS (Kato et al., 2017) and GHBM (Devane et al., 2019) both use the same structure as in Paper A. In THUMS-D (Martyntenko et al., 2019), one controller per muscle is used, and thus no muscle recruitment was needed in that loop.

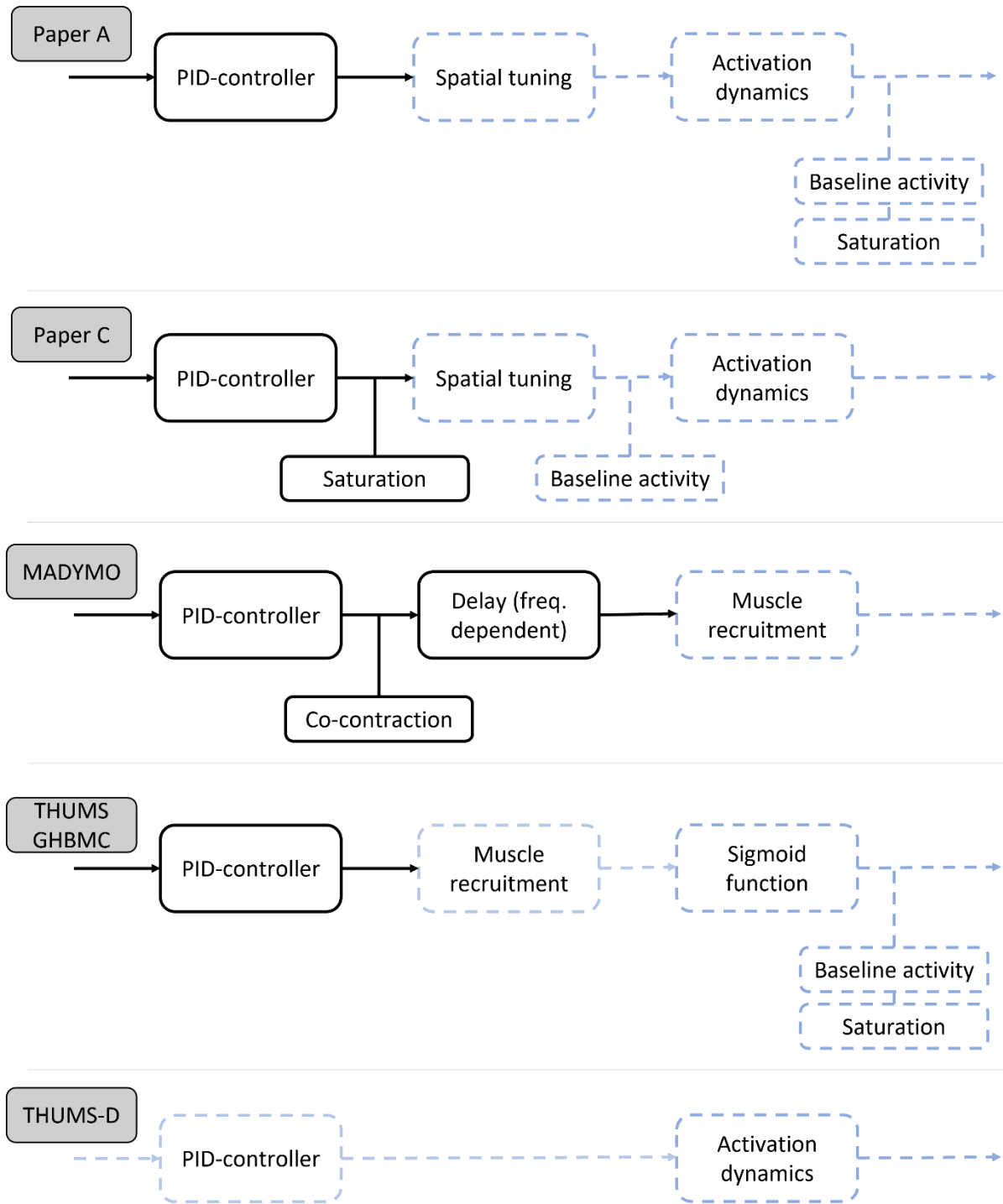


Figure 20. APF controller sequence in Papers A and D, Papers C and E, MADYMO model (Meijer et al., 2012, Meijer et al., 2013), THUMS (Kato et al., 2017), GHBMC (Devane et al., 2019) and THUMS-D model (Martynenko et al., 2019). Black solid lines indicate signals common for all muscles, while blue dashed lines indicate muscle-specific signals.

Common for the models are some type of activation dynamics, which exists to model the process in the muscle where the activation signal is turned in to force generation (Winters and Stark, 1985, Hatze, 1977), and should thus be placed last.

If the saturation was placed after the spatial tuning, an increasing signal requesting above maximum activity would saturate the agonist before the other muscles. Consequently, the other muscles would increase in activity while the agonist activity would remain at maximum, leading to distorted scaling where eventually all muscles would be at the maximum activity level. It could also lead to a

situation where the controller no longer can control the movement, as the movement created by the controller might be very different in direction compared to expectations of the controller, creating a positive feedback loop. Thus, the saturation point was placed before the spatial tuning, and spatial tuning patterns were scaled accordingly. For a signal that never requires saturation, the placement of saturation is irrelevant, and volunteer EMG data reveal that for most muscles and most cases, the muscle activity of volunteers remain below MVIC (Ghaffari et al., 2019, Ólafsdóttir et al., 2013, Östh et al., 2013).

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

The muscles were controlled using PID controllers (Papers A, C-E), similar to in THUMS (Kato et al., 2018, Kato et al., 2017) and GHBM (Devane et al., 2022). Other muscle controllers exist in literature. For instance, open-loop control was used in (Barker and Cronin, 2021), and the THUMS-D model (Martynenko et al., 2019) uses a hybrid approach, with PID control and open loop control combined, both with satisfactory results. Open-loop could be used when the muscle activation pattern is known a-priori, which is typically not the intended use case for the HBMs, as they should be predictive and should be able to extrapolate to unknown situations. Using a feedback controller allows for extrapolation from one maneuver to another. Looking beyond human body models for vehicle safety, inverse kinematics has been utilized to control muscles for instance in simulations of running (Rasmussen, 2019) and walking (Shourijeh et al., 2016). These inverse kinematic models calculate a muscle activation pattern, or a joint torque pattern based on recorded or desired kinematics, and possibly other boundary conditions such as ground reaction forces. Because the inverse kinematics approach uses the kinematics as input, the methods are retrospective and not predictive in terms of kinematics. In (Rasmussen, 2019), inverse kinematics were used to calculate joint moments for 90 recorded runners. The joint moment time histories were analyzed using PCA, in an attempt of creating statistical models that can predict joint moments based on selected human characteristics, in a similar way as was done for kinematics and belt forces in Paper B. Although this approach would make the models of running predictive, they would only predict running patterns for runners of different characteristics, and not for instance walking. Since HBMs should be able to extrapolate from the evaluated scenarios, an approach such as the one used in (Rasmussen, 2019) alone would be unsuitable for an AHBM muscle controller.

6.3 OCCUPANT VARIABILITY

Just as for previous studies, the results from Paper B indicate that the human characteristics explain parts of the variance between occupants, but considerable variance remained after accounting for these characteristics. The largest influence was from belt system used. The results in Paper D and Paper E indicate that some of the variance could be explained by the seat position and spinal alignment. More specifically, spinal alignment was found influential in vertical kinematics.

Throughout this section, previous volunteer studies will be referenced by the summarizing name from Table 1, and these references will be highlighted in **bold font**.

The results in Paper B identify the same influences from occupant characteristics as **Ghaffari et al.**, Figure 21, which is expected because those publications contain a subset of what was presented in Paper B. For that reason, the **Ghaffari et al.** data will be omitted from further comparisons.

The findings in this thesis identify the relative importance of the belt system. In Paper B, significant differences of kinematics between using a standard inertia-reel belt and a pre-pretensioned belt were identified in several maneuvers. Although not the focus of that study, similar differences were found in Paper A, where simulations of different belt systems were conducted. Further, in Paper E, the belt stiffness was indicated to be of relative importance compared to other vehicle parameters, when varying the belt stiffness after belt locking in line with the differences seen in Paper B. The relative importance of the belt system agrees with findings from **Arbogast et al.**, **Holt et al.**, **Huber, Kirschbichler et al.**, **Ólafsdóttir, Östh et al.**, where different belt systems were compared in volunteer experiments, Figure 21. Neither of the studies analyzed the belt stiffness, as was done in Paper E. Since the results are consistent across studies, it can be concluded that belt pre-pretensioning reduces head displacements in evasive maneuvers.

For occupant characteristics the pattern of influence was less clear when comparing between studies, Figure 21.

- Larger displacements were indicated for males in lane change in Paper B, and partly in **Chan et al.**, while **Graci et al. 2020**, **Reed et al. 2018** and **Ólafsdóttir, Östh et al.** found no differences between sexes. **Huber, Kirschbichler et al.** indicated a difference between males and females, but did not mention if males or females displaced more.
- Stature was found to correlated with displacement in braking in **Huber, Kirschbichler et al.**, while in **Reed et al. 2018**, **Reed et al. 2021** and Paper B, no correlation between stature and displacement in braking was found. In lane change, smaller displacements for taller occupants were found in Paper B, while the opposite trend was found in **Reed et al. 2018** and **Reed et al. 2021**. **Huber, Kirschbichler et al.** found no correlation between displacements and stature in lane change.
- Age was not found to correlate with displacements in Paper B, agreeing with **Graci et al. 2019**, **Holt et al.** and partly with **Reed et al. 2018**, but in contrast to in **Arbogast et al.**, **Graci et al. 2020**, **Reed et al. 2021**, and partly **Reed et al. 2018**. It should be noted that **Arbogast et al.**, **Graci et al. 2019**, **Graci et al. 2020** and **Holt et al.** included children and adults, while in the other studies, only adults were included.
- BMI was typically not correlated to displacements, with the exception of in **Reed et al. 2018** where occupants with larger BMI displaced less.

Since the compared studies, including Paper B, disagree on the effects from occupant characteristics, any true effect from these characteristics is likely small, or there are interactions between these characteristics and other characteristics that not yet have been identified.

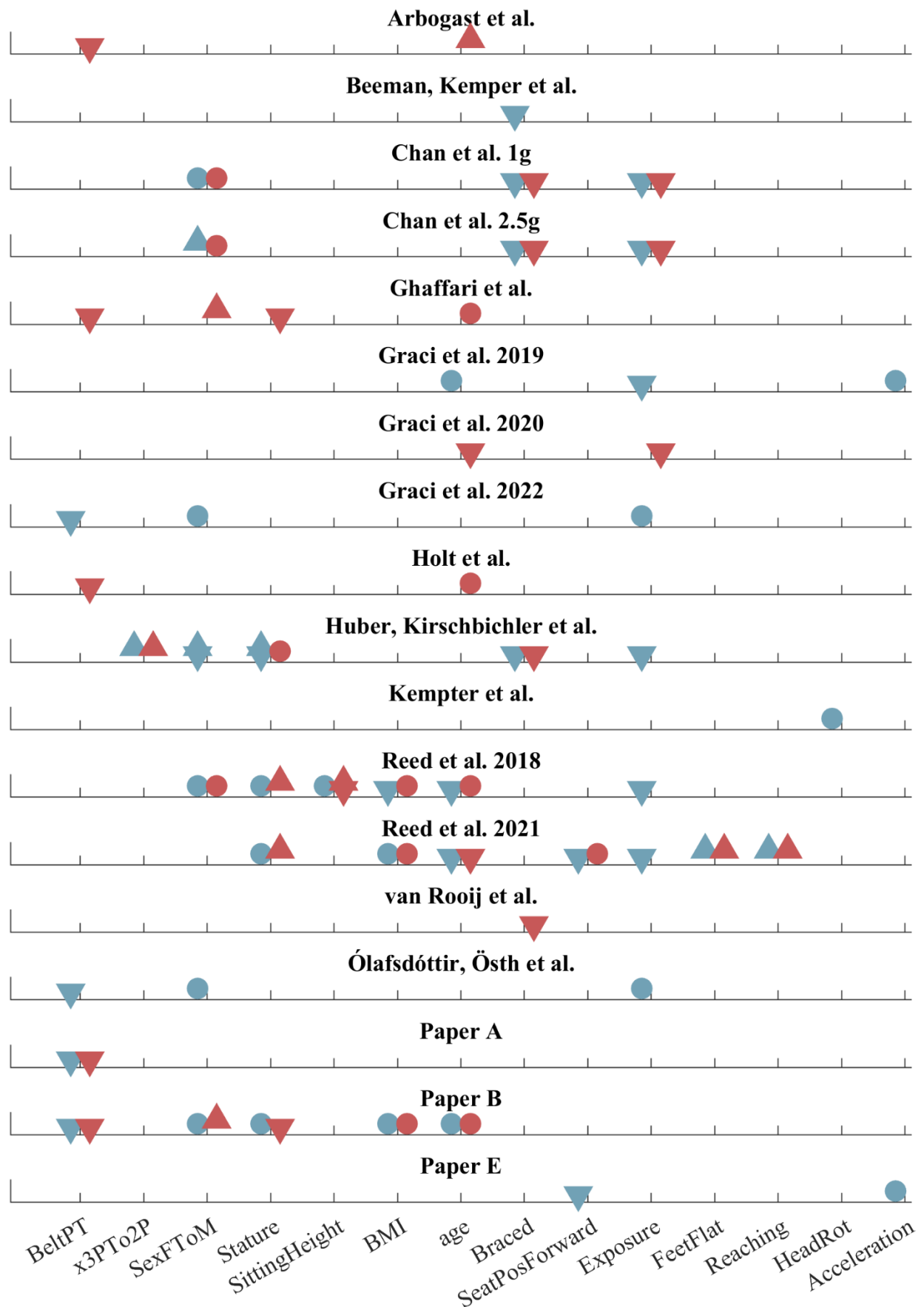


Figure 21. Effect from change in characteristic on head displacement (torso displacement for Arbogast study). Red markers are tests with longitudinal acceleration, blue markers are tests with lateral or oblique accelerations. Arrows indicate the

direction of head displacement difference for that specific characteristics change (upwards indicates larger displacement if change to binary characteristic or increase in numerical characteristic). Overlaid arrows upwards and downwards indicates that there was a difference between conditions, but the direction of change was not specified. Dots show characteristics that were included in the analysis but found not to be influential. Full references are presented in *Table 1*.

Based on the findings in Papers B, D and E, combined with research from others, Figure 21, it is unlikely morphing (alone) is sufficient for capturing variance in pre-crash, because the occupant characteristics have limited explanatory power in evasive maneuvers. However, because the models are run seamless between pre-crash and in-crash (Wass et al., 2022, Östh et al., 2022), morphing capabilities are still needed for the active muscles. Introducing morphing capabilities to the SAFER AHBM should be possible. The spatial tuning patterns in the neck and lumbar APF controllers were created with data from both males and females, and thus the STPs would not need to be updated to represent females. The PCSAs were defined based on males, and because these can influence the results (Paper D and E), the PCSAs should be updated if the control system should be used to model a female occupant.

In Paper E, a method to account for variability from different model parameters was used. The results of the study showed that a more forward seat position reduced forward displacements in braking. The same trend was seen in **Reed et al. 2021**. The predicted reduction in displacement, per distance, was 7% larger in Paper E compared to in **Reed et al. 2021**. These results indicate that the effect from seat longitudinal position was related to geometrical differences between the positions, and not differences in strategy from volunteers, since no volitional control was included in Paper E. Furthermore, the similarities between the results in Paper E and **Reed et al. 2021** indicate that the method and HBM used in Paper E were capable of capturing true effects from geometrical variations. Based on these results, it would be beneficial to keep shoulder belt geometry constant in volunteer tests, and the volunteers positioned such that the geometry is comparable across subjects. Ideally, this should be done while still maintaining the same foot position, as the feet were found by **Reed et al. 2021** to influence kinematics, and the same distance to the vehicle interior, suggested to influence the volunteer behavior in **Huber, Kirschbichler et al.**. If it is not possible to keep the same belt geometry, foot position or distance to vehicle interior, the distance to the vehicle interior should be prioritized, while the foot position and belt geometry should be recorded, because the AHBM can be used to simulate the effect from geometrical changes, while they currently do not capture the change in behavior from changed distance to vehicle interior.

Coefficients of variation (CoV) for peak displacements in Paper B, based on average volunteer (43% male, 36.3 years old, 174.5 cm tall and BMI of 22.8 kg/m²), were similar compared to previous work, while the coefficients of variation for simulations in Paper E were lower compared to most previous studies. The CoV from Paper E was comparable to those from **Beeman, Kemper et al.** and **Chan et al.**, which both were conducted in a more controlled environment (sled) compared to the intended environment in the simulations (in vehicle), Figure 22. This highlights the need to evaluate more than the average response to maneuvers, since the variation is almost 50% of average response for in-vehicle tests, Figure 23, and thus the average response might not be representative of the population. It also highlights the need for understanding where the variability in in-vehicle tests come from, because this should be accounted for.

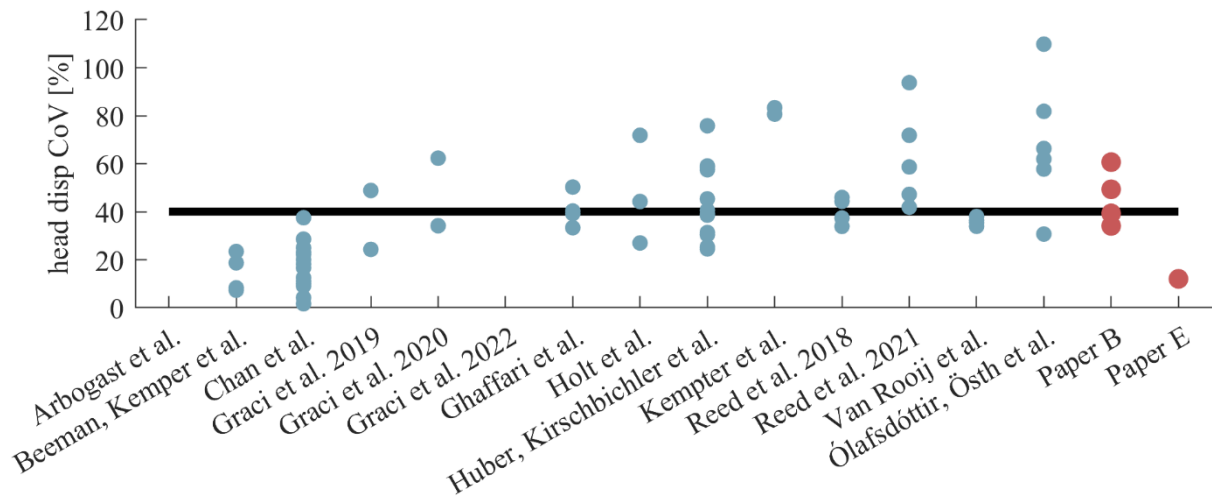


Figure 22. Coefficient of variation for head displacement in different publications (top) and sorted based on environment (bottom). Blue dots are previous work, red dots are results from this thesis. Black horizontal lines show average coefficient of variation, for all previous work (top), and average for previous work for each environment (bottom). Full references are presented in Table 1.

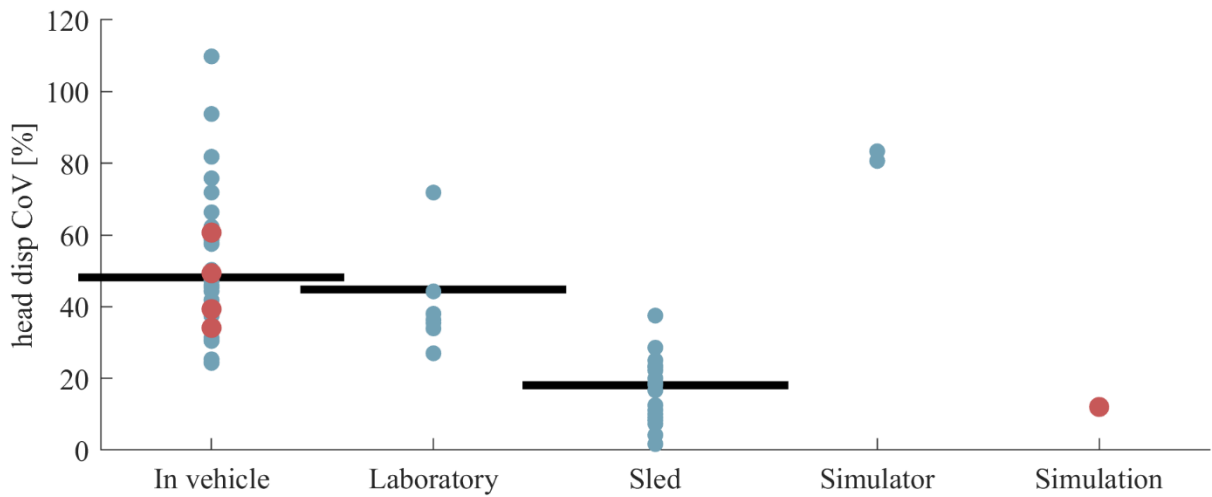


Figure 23. Coefficient of variation (CoV) for head displacement sorted by environment. Blue dots are previous work, red dots are results from this thesis. The black line shows average CoV per environment. No average was calculated for the simulator and simulation environments because only one study (Kempter et al. (Kempter et al., 2018, Kempter et al., 2022)) was classified as a simulator study, and only Paper E as a simulation study. Full references are presented in Table 1.

The subject characteristics and sample sizes were comparable for Paper B and Paper E and the previous studies, Figure 24.

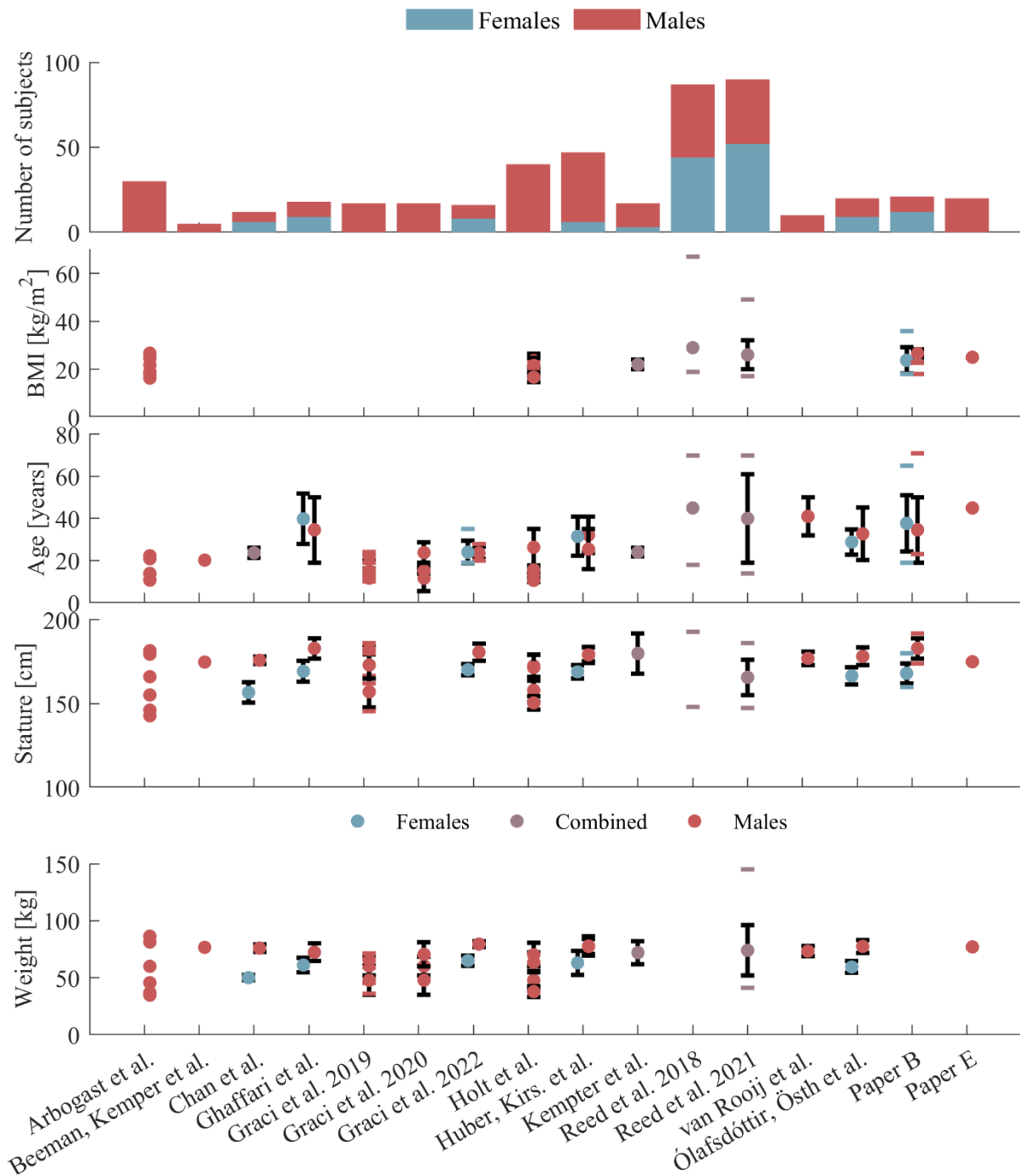


Figure 24. Number of subjects, BMI, age, stature and weight in different publications. Average values are represented with dots, standard deviation with error bars and minimum and maximum with horizontal colored lines. Females are blue and males in red. For some publications, BMI, age, stature and weight was presented for females and males together, these are shown in purple. In some publications, such as Arbogast (Arbogast et al., 2012) BMI, age, stature and weight for several groups were presented, and for those publications, each group was plotted. Full references are presented in Table 1.

6.4 LIMITATIONS AND FUTURE WORK

In this thesis, steps have been taken to quantify the expected variability from different types of characteristics of the occupant and vehicle, but still a lot of variability remains unaccounted for. Future work should include efforts to understand the large variability in displacements in volunteer tests. The results from Paper B could be used to vary the HBM to represent an occupant with larger or smaller displacements compared to the average response, for instance by tuning the controllers such that they match the occupants at specified deviations from average, such as ± 1 standard deviation. However, without understanding of underlying mechanisms behind the large variability, important systematic differences might be overlooked. Based on results in papers D and E, the spinal curvature might be one such systematic difference, but this influence needs to be confirmed with additional studies.

As discussed in Paper D and Paper E, the parameters found influential in the simulations could be investigated from volunteer experiments. The effect from muscle size seen in Paper D could be investigated retrospectively for already existing data sets where strength of volunteers has been presented, such as (Östh et al., 2013). To the best of my knowledge, no existing volunteer studies have investigated correlations between posture and kinematics in maneuvers, and thus new studies are needed to identify any effect from posture. Correlating posture to kinematics could prove challenging, but since the spinal curvature was identified as influential in both Paper D and Paper E, investigating the influence from spinal curvature on kinematics in evasive maneuvers could potentially increase the understanding of occupant variability.

The HBMs used throughout this thesis have been models of an average sized adult male. Both the neck and lumbar APF controllers have been developed with spatial tuning patterns combined from both male and female participants whenever possible. With updates to the physical cross-sectional areas, and possibly re-tuning, the controllers should generalize to the adult female population. Child occupants have been found in some studies to displace differently during maneuvers compared to adult occupants (Graci et al., 2020, Arbogast et al., 2012) and would benefit from a child specific AHBM. The controllers could possibly generalize to the child population as well, but this would require updates to PCSA, re-tuning of the controller parameters, and adjustments to the spatial tuning patterns, as those used in this thesis were collected from adult volunteers.

All included studies have been limited to passengers seated in a nominal or preferred neutral posture, i.e., the posture was based on a person that is asked to sit in a normal position. The influence of spinal alignment was tested in papers D and E, but the alignments were intended to represent variations of a normal posture, as the volunteers in the study used to align the spines (Izumiya et al., 2018, Nishida et al., 2020) were all asked to sit in their normal driving posture. In a recent study, some degree of slouching was found when asking subjects to sit in their preferred posture (Bohman et al., 2023), agreeing with the slouching found when positioning the models in Paper D. When simulating these different normal postures, the belt manages to retain the occupant during the maneuver. As seen in (Reed et al., 2021), adopting different postures, such as moving the feet position or reaching for objects changes the kinematics in the maneuvers. No variations beyond a normal posture, such as those seen in (Reed et al., 2020) have been used in this thesis.

Although indicated to influence predicted crash outcome, little effort has been put on validating the muscle activation levels in this thesis. Attempts were made in Papers A and C, but fundamentally what is produced by the CNS and measured with EMG, and what is produced by the controllers is different, making it difficult to compare the two entities.

If new data is collected with the intention of creating new STPs as input for active HBM controllers, such as the rotational controller, the experiments should target MVIC normalization and dynamic conditions. Ideally, if possible, the experiment should capture the reflex muscle activity as opposed to the activity during voluntary movement. Including also baseline activity recordings, in a representative posture while seated in a vehicle seat would be beneficial. The assumption of symmetry used in the mirroring of left/right side data should be evaluated in pilot testing. If the assumption holds, the test could either be done on half of the STP directions only with bilateral EMG recordings and then mirrored, or unilateral data could be recorded but with a full test matrix. A third option would be to use bilateral recordings and full test matrix and combine both left and right-side muscle as was done in Paper C. If the assumption of symmetry does not hold, bilateral muscle activity and a full test matrix should be used.

One common pre-crash/crash event is run-off-road (Riexinger and Gabler, 2018), a scenario where accelerations in all directions can arise (Jakobsson et al., 2014). The current model is capable of handling horizontal plane loading only. Furthermore, future occupant postures could include reclined postures (Koppel et al., 2019, Nie et al., 2020), which combined with a braking case would introduce similar loading of the spine as vertical acceleration for an upright occupant. The APF controller implementation has been used to simulate repositioning occupants from reclined to upright (Östh et al., 2020), it was noted that the model tensed the extensors during repositioning to upright, but since no validation data for that scenario exists, it cannot be determined if a human would behave in a similar way. Further, if simulating a fully reclined occupant, or an occupant subjected to vertical loading, it is likely that the model does not respond like a human would, because a pure compression of the spine would not be noticed by any of the controllers included in this thesis. Similarly, it is likely that the MLF controller would not respond in a human-like way to this loading, because with a perfect compression the muscles would not lengthen. Therefore, an important extension to the model will be the capability of handling vertical loading. This will likely require updates/extensions to several of the controllers and will also require additional validation and evaluation data. Such data exists in a simplified environment (Kang et al., 2021), but as discussed previously, ideally both simplified and in-vehicle data should be used to evaluate the model. The additional data would ideally be collected from volunteers exposed to low-level vertical loading and maneuvering in a reclined posture, in a representative vehicle environment.

In both Paper D and Paper E, the influence of characteristics on kinematics in simulations of braking was investigated. The effect of similar characteristics on kinematics in other maneuvers, such as lane change, remains to be explored. Based on results from (Reed et al., 2021) and Paper B, the characteristics that were found to be influential differed between maneuvers, it is likely that this holds for also other types of characteristics not included in those studies. Thus, future work could include repeating the process from Papers D and E for other types of maneuvers.

In Paper C, the gains of the neck muscle controllers were tuned to match volunteer kinematics in braking and lane change. No such tuning has been performed on the lumbar muscle controller. This work was initiated, but due to uncertainties in the boundary conditions, the attempts were abandoned in favor of Paper E, where boundary conditions were investigated.

After tuning of lumbar gains, the model should be validated against a different set of volunteer experiments, to ensure that the performance of the model generalizes to simulations in other environments. With the knowledge obtained in Paper E, a suitable validation case would, besides kinematics, contain knowledge on seat position relative D-ring and occupant shoulder, and seat belt stiffness. Ideally, some information regarding spinal alignment should also be present, although to

the best of my knowledge, no such information has been provided in any of the volunteer experiments.

Although the neck and lumbar controllers were developed to emulate vestibular and stretch reflexes, the implemented controllers are simplified, representing pragmatic approaches instead of perfect analogies to the human motor/postural control system. As described in Section 4.1.1, humans could respond to numerous types of input in certain vehicle maneuvers that are currently not included in the model's feedback system, such as visual or tactile input.

7 CONCLUSIONS

In this thesis, the SAFER HBM neck and lumbar muscle controllers were further developed, and the model was used to investigate passenger variability in braking maneuvers. Kinematics and belt forces from a volunteer experiment was analyzed and served as input to subsequent simulation studies. During this thesis:

- The existing muscle controllers in SAFER HBM v9 were further developed to accommodate simulations of maneuvers that include vehicle yaw. The updated neck muscle controller was tuned to volunteer experiments.
- A new neck muscle controller that responds to rotations in three dimensions was developed and tuned to volunteer experiments.
- A volunteer test series was analyzed, and regression models with occupant characteristics and belt systems as predictors for time-series kinematics were presented for a wide range of maneuvers and belt systems.
- The Active SAFER HBM v10 with muscle controller updates from this thesis was used to identify characteristics that induce variability in kinematics in braking.

7.1 MODEL DEVELOPMENT

Including active muscle control improved model kinematic predictions in simulations of evasive maneuvers, as compared to using the same model without active muscles. Before tuning, controllers emulating reflexes from the vestibular system provided better prediction compared to controllers emulating muscle stretch reflexes. The SAFER HBM with tuned controllers could predict passenger head kinematics with good to excellent bio-fidelity, with overall CORA scores of 0.90-0.94 in braking and 0.81-0.82 in lane change. Head translations were better predicted compared to head rotations. The angular position feedback controller provided better predictions compared to the rotational controller. When varying model parameters in braking, the forward displacements were more similar early in the maneuver compared to later in the maneuver.

7.2 OCCUPANT VARIABILITY

From the volunteer study it was concluded that although some of the variations in displacements could be explained by occupant characteristics such as sex, stature, age and BMI, the largest effect was seen when changing belt system. The belt was found influential in 33% of possible prediction models, while the other predictors were influential in 6% or fewer possible prediction models. In subsequent simulation studies, occupant spinal alignment and seat longitudinal position was identified as influential parameters of forward displacements. When varying the three most influential HBM parameters (spinal alignment PC1, spinal alignment PC2 and muscle physiological cross-sectional area) and the three most influential boundary conditions (seat longitudinal position, velocity change and belt stiffness), the vertical displacements varied similarly compared to how volunteers varied. Forward displacement corridors were around 25% of width from volunteer corridors when varying these parameters. The most influential parameter was seat position, moving the seat forward with 66% of the travel range reduced the head displacements by 45 mm, followed by spinal alignment PC2, a more curved spine reduced the displacements, altering the curvature by 2 SD (from -1 SD to +1 SD) reduced head displacements with 28 mm. Together, these two parameters explained 70%-79% of the forward head and torso displacements seen in the simulations.

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