Passenger Kinematics and Muscle Responses in Autonomous Braking Events with Standard and Reversible Pre-tensioned Restraints

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Abstract Biofidelic human body models (HBMs) with active muscles are valuable tools for assessing the safety potential of systems that are active immediately before and during a crash. For validation, experimental data including muscle activity are required. This paper provides a data set for front seat passengers in autonomous braking events comprising 20 volunteers (11 male and 9 female) in a passenger car. Volunteers were subjected to two different autonomous braking test cases of 1.1 g, wearing a standard belt and a reversible pre-tensioned belt activated 200 ms before deceleration onset. The following data were collected: muscle activity with electromyography, kinematics with video tracking, footwell force, belt force and belt pay-out. Head and T1 displacements were shorter with a pre-tensioned belt while head rotation was similar for both test cases. Kinematics did not display any significant gender differences. Average muscle activity with a pre-tensioned belt increased rapidly before onset of deceleration for females, but not for males. Muscle activity, predominantly in the cervical and lumbar extensors, increased soon after vehicle deceleration onset for all volunteers wearing the standard belt. All muscles were significantly more active during braking than normal driving. Data are presented in corridors for use when validating active HBMs.

Keywords Emergency braking, EMG, kinematics, reversible pre-tensioned belt, volunteer.

I. INTRODUCTION

Fatalities and severe injuries due to road traffic accidents are a serious health and economical issue in today's society. In recent years, integrated safety systems have been developed to prevent or mitigate traffic accidents and occupant injuries. To effectively evaluate the performance of these safety systems there is a need for biofidelic tools that can estimate the occupant response before and during the crash. Human body models (HBMs) have been used to simulate this response. Utilising HBMs that incorporate active musculature to predict pre-crash occupant kinematics can be valuable for the performance evaluation and optimal design of safety systems active in the pre-crash and the crash phase [1]. However, in order to develop biofidelic HBMs, a thorough validation is required. To validate and improve HBMs that are targeted at front seat passengers and account for active muscle responses, there is a need for passenger validation data that include muscle activity measurements, which is the topic of this paper.

The reversible pre-tensioned belt is an integrated safety technology that has become increasingly prevalent in modern passenger vehicles. If an imminent crash is detected the belt is tightened and the occupant is repositioned before the crash occurs. Repositioning, provoked by belt pre-tension affects the occupant's dynamic behaviour. Furthermore, accident statistics show that drivers commonly apply the brakes before unavoidable frontal collisions [2–4]. With the introduction of autonomous braking systems in vehicles the pre-crash braking scenarios can be expected to occur even more frequently in the future. While intended to reduce accident severity, pre-crash braking will influence the coupling between the occupant and the vehicle. During belt pre-tension or autonomous braking the occupant is likely to become aware of the impending situation. Anticipation of the imminent crash may trigger a reaction in the occupant, causing a dynamic response that is influenced by muscle tension [5–8]. Thus, accounting for muscle activity is essential in order to predict occupant kinematics in emergency events.

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To validate active HBMs for a frontal pre-crash situation, a thorough comparison to volunteer experimental data on kinematics, muscle activity, and vehicle interaction forces is needed. Furthermore, it is essential to have access to such data preceding the pre-crash in order to establish the initial boundary conditions when setting up an HBM simulation. Several studies have reported results from volunteer testing where subjects were exposed to a frontal impact pulse [2], [5], [6], [9–17]. These studies have predominantly been performed with an experimental sled where kinematics and/or muscle activity have been recorded, although most of the studies have focused on the driver response [5], [9], [10], [11]. Drivers behave differently to passengers as their sitting posture is different and they have the option of bracing their arms against the steering wheel. Data on driver kinematics are thus not suitable for the validation of a passenger HBM. Other studies have not included muscle activity data [2], [12–14], have not quantified the muscle activity with respect to maximum voluntary contraction [6], [15] or have only focused on a single body part [16], [17]. The joint conclusion reached in most of these studies is that occupant kinematic response is affected considerably by muscular contraction. Despite this, it remains unclear what levels of contraction and co-contraction are expected in a pre-crash situation.

The aim of this study was to provide a detailed data set suited for validation and initialisation of an active passenger HBM for pre-crash braking simulations. Volunteer tests were performed in order to quantify the muscle contractions in autonomous braking events and relate them to passenger kinematics and boundary conditions for passengers restrained with both standard and reversible pre-tensioned belts. To closely model a real life pre-crash situation, the test was conducted in a passenger car being driven in actual traffic surroundings.

II. METHODS

Front right seat passenger volunteer kinematics, electromyographic (EMG) responses, and vehicle interaction forces were recorded in 1.1 g autonomous braking events in a test vehicle driving on relatively empty rural roads. 20 subjects were exposed to 29 trials each, first 20 in the driver's seat (presented in [18]) followed by 9 in the passenger seat. Three types of test cases were performed in the passenger seat, two of which are reported in this study. In one test case, the volunteers were restrained with a reversible pre-tensioned belt while in the other they were restrained with a standard seat belt. All volunteers performed maximum voluntary contractions (MVCs) for EMG normalisation prior to the test. The protocol of this study was reviewed and approved by the Ethical Review Board at the University of Gothenburg, Sweden.

Subjects

20 healthy volunteers, 11 males and 9 females participated in this study (TABLE I). The volunteers did not have a history of neck pain, poor general health, or other medical conditions that could present an increased risk for injury, as assessed independently by a nurse prior to testing. All volunteers provided informed consent prior to testing by signing a form to that effect, and were compensated with a 45 EUR gift voucher. Volunteers wore a thin sleeveless shirt and their own trousers and shoes during testing.

Three months after the test, volunteers were sent a follow-up questionnaire to ascertain if they had complaints or pain due to participation in the test. All volunteers responded and none had experienced any type of neck pain related to the autonomous braking events.

TABLE I
VOLUNTEER AVERAGE (STANDARD DEVIATION) AGE AND PHYSIQUE.

	Females	Males
No.	9	11
Age [years]	28.8 (5.9)	32.7 (12.5)
Height [cm]	166.6 (5.0)	178.2 (5.2)
Weight [kg]	59.4 (5.2)	77.5 (5.6)
Sitting Height [cm]	89.4 (3.3)	94.7 (2.6)

Test Vehicle and Autonomous Braking Test Cases

The autonomous braking test cases were performed in a commercially available 2012 Volvo V60 T4. Autonomous braking was induced by a PC running Canalyzer v7.6 (Vector GmBH, Stuttgart, Germany), actuated with the vehicle's hydraulic pump and initiated by a test leader pressing a button in the rear seat. Two VN1640 (Vector GmBH, Stuttgart, Germany) cases were used for communicating with the vehicle and triggering the autonomous braking events. The vehicle was fitted with new standard summer tyres (Continental SportContact 3 235/45/R17 inflated with 250 kPa) and the seats were upholstered in leather.

The original retractor for the standard seat belt was replaced by an Active Seatbelt retractor (Autoliv, Stockholm, Sweden), incorporating an electrical motor that allows for reversible pre-tension. In the active mode, a nominal tension of 170 N was applied, while in the passive mode tension was not applied. In the passive mode, the characteristics of the seat belt were identical to the test vehicle's original standard seat belt.

Three autonomous braking test cases were performed. For two cases, reversible seat belt pre-tension was applied and for one case pre-tension was not applied. This paper includes the results from two of the test cases:

Autobrake PT – Autonomous braking combined with 170 N seat belt pre-tension, applied 0.2 s before the initiation of the braking of the vehicle.

Autobrake SB – Autonomous braking without seat belt pre-tension, active belt operated in passive mode.

The initial velocity in all cases was approximately 70 km/h and the target was to reach maximum vehicle deceleration. The duration of the autonomous braking interventions was chosen to yield a velocity change of 50 km/h. The average applied acceleration pulses in the Autobrake PT and Autobrake SB test cases are plotted in Fig. 1. The onset of the acceleration pulse is plotted with a dashed line, in the figure, and the belt pre-tension onset with a dotted line. In both cases onset is defined as 5% of the peak of the respective average curves.

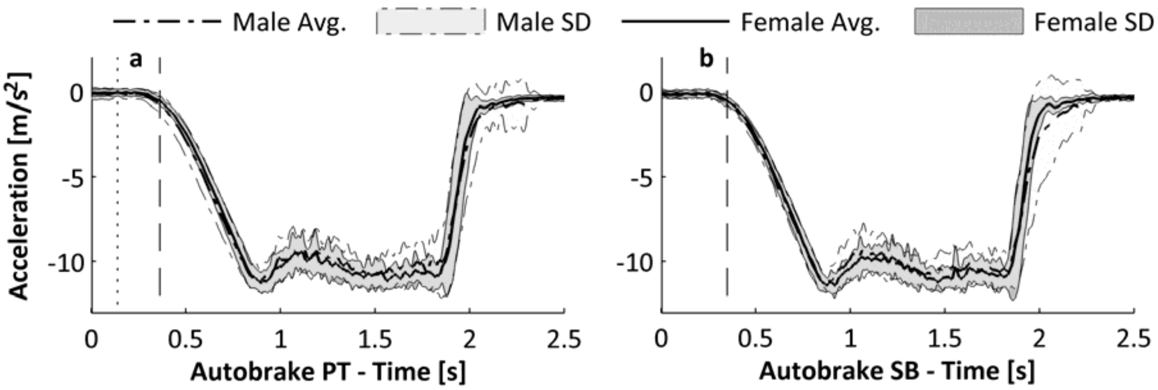

Fig. 1. Average ± SD (Standard Deviation) vehicle acceleration in Autobrake PT (left) and Autobrake SB (right) for males (light grey, dashed dotted lines) and females (dark grey, solid lines). Vertical dashed lines represent acceleration onset and dotted line shows the belt pre-tension onset.

Instrumentation

Volunteer's motion was captured with a UI-5220CP-C Gigabit Ethernet CMOS colour camera (IDS GmbH, Obersulm, Germany) at 50 frames/s and retrieved with the film analysis software, TEMA Automotive (Image Systems, Linköping, Sweden). Film markers were attached to the volunteers, see Fig. 2. Three markers, 25 mm in diameter, were attached in a triangular pattern on the left side of the head. The base of the triangle was parallel to the Frankfort plane, approximately 20 mm forward of the external auditory canal. A 50 mm marker attached on a T-shaped holder was taped on top of the manubrium; the marker centre positioned 25 mm below the sternal notch and 35 mm in front of the chest. A white 25 mm diameter circular half sphere was placed on the skin covering the C7 process, located by palpation. Four 50 mm markers were attached to the interior passenger side of the vehicle; one on the seat head rest, one on the A-pillar, and two on the door.

EMG activity was measured with surface electrodes placed bilaterally on the sternocleidomastoid (SCM), cervical paravertebral muscles (CPVM), rectus abdominis (RA), lumbar paravertebral muscles (LPVM), biceps

(BIC), triceps (TRIC), anterior deltoid (ADELT) and posterior deltoid (PDELT), see Fig. 3. A single electrode was placed on the anterior superior iliac spine for reference. Electrode placement is described in TABLE III in the Appendix. EMG was recorded with a sampling rate of 2048 Hz using a Compumedics Grael (Compumedics, Abbotsford, Australia). Paired Ag/AgCl electrodes (Blue Sensor N–00–S, Ambu A/S, Ballerup, Denmark), with an interelectrode spacing of 20 mm, were used. Before application of the electrodes, the skin was prepared by shaving, abrading with sandpaper (P180) and wiping with 70% ethanol solution.

Vehicle longitudinal acceleration was measured with a 10 g accelerometer (EGCS-D1CM-10, Entran, Hampton, Virginia) mounted on a bracket to the left front bolt holding the passenger seat to the body-in-white. The vehicle footwell force was measured by mounting a PZ1,0 force transducer (Load Indicator AB, Göteborg, Sweden) on a balance plate under the original passenger foot support in the footwell. Shoulder and lap belt forces were measured using low-range (<6 kN) belt force transducers (DK11-13, Messring GmbH, Krailling, Germany and EL20-S458-7B, Altheris bv, Den Haag, Netherlands). Finally, belt pay-out was measured using an optical belt movement sensor (IES 2098, IES, Braunschweig, Germany).

Sensor excitation and signal amplification were made with multiple DAQ-P Bridge B (Dewetron, Wernau, Germany) bridge amplifiers. All analogue to digital (AD) signals were recorded using a USB-6251 AD-converter (National Instruments, Austin, Texas) running at 2048 Hz. Film and EMG data were received over Ethernet through a Gigabit Ethernet switch (GS108, Netgear, San Jose, CA).

Fig. 2. The camera view in the vehicle.

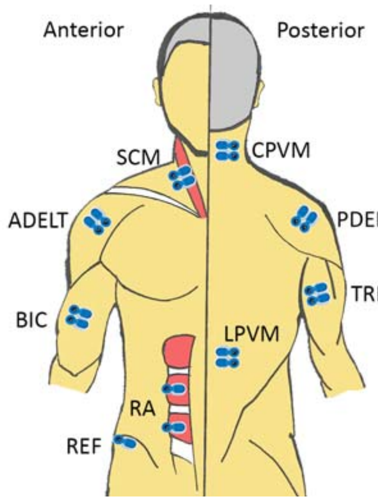

Fig. 3. Electrode placement [18].

Test Procedure

Before undergoing the autonomous braking events the volunteers performed MVCs in a custom made test rig, for EMG normalisation. In the test rig the volunteers were asked to adopt a posture that resembled a typical sitting posture in a vehicle. Volunteer MVCs were recorded in isometric conditions for each muscle tested. Three repetitions of MVCs were performed for each muscle with a minimum of 30 s rest between contractions. Volunteers were provided with a visual feedback on their exerted force levels. The volunteers had a minimum of 20 minutes break between the MVCs and the beginning of the braking events. The test rig construction and the MVC procedure are described in further detail in [18].

The braking tests were conducted in two parts. Volunteers were given a minimum of 10 minutes break between the two parts. The first part consisted of 20 braking events where the volunteer sat in the driver's seat, as reported in [18]. In the second part, the focus of this study, the volunteers underwent autonomous braking events in the front passenger seat. During these braking events the passenger seat was positioned according to the Euro NCAP frontal impact test protocol (Euro NCAP 2012), i.e., mid fore/aft position, with the seat back angle according to the manufacturer's specification (22°). The volunteers were instructed to keep their feet symmetrical to the midline of the footwell and to rest their hands on their lap during the test. Each volunteer underwent nine autonomous braking trials in the passenger seat in total, i.e., three trials for each test case. The three different cases were randomised, with the exception of the first trial which was always the Autobrake PT. A test assistant drove the car and the braking events were initiated by the test leader seated in the rear right seat pushing a trigger button. The volunteers were not notified prior to trial initiation.

Data Analysis

Lens distortion compensation was made in the film analysis software. The angle of the camera with respect to the average motion plane of the volunteers was also accommodated for. Occasional lost frames in the image sequences were detected and replaced with a copy of the preceding frame. The average number of lost frames per image sequence was 2.7 out of 201 frames. For each frame, a time stamp was recorded. Using the stored time of frame acquisition, a restored image sequence remained synchronous with other measurements such as EMG. The accuracy evaluation of the film analysis, assessed as described in [18], revealed that in a worst case scenario a 9% length distortion could be present. Sensitivity of the camera to the acceleration loads applied during the breaking events was assessed. The markers on the passenger door were found to move less than 0.5 mm in both X and Z direction due to these loads.

Displacement data are presented in a vehicle fixed coordinate system defined in accordance with ISO8855 [19], i.e., positive X is forward and positive Z upwards. Head centre of gravity (COG) coordinates were calculated based on head COG data from the literature [20–22] by transforming head marker displacements as defined in [18]. Head rotation data were represented as the angle of the Frankfort plane with respect to the horizontal plane. The marker above the C7 process was used to represent T1 displacements. All displacement data, rotational as well as linear, are provided relative to the respective positions at trial initiation.

The raw EMG signals were band-pass filtered (4th order Butterworth, 10-350 Hz), fullwave rectified, and smoothed with a 40 ms (82 sample) root mean square (RMS) window. For the RA and the LPVM the low-pass filter limit was increased to 50 Hz to remedy heart rate artifacts. Furthermore, EMG was normalised with the maximum 1 s RMS window found out of all the three MVCs made for each muscle.

Footwell force, belt force and belt pay-out measurements were filtered and the measurements from the footwell force transducer were corrected by removing the inertial loading of the effective mass of the balance plate and sensor.

Differences between male and female subjects and between the Autobrake PT and Autobrake SB test cases were analysed using a one-way ANOVA blocked by subject and a post-hoc test at a 5% significance level. Before performing the ANOVA, data were checked for normality and a homoscedasticity assessment was also made. A non-parametric test (Friedman test) with a 1% significance level was applied to any non-normal data (here only applied to EMG data). For simplification during the application of this test, any unbalanced set of trials was removed. All differences analysed with aforementioned statistical methods were assessed in the time interval 1.6 – 1.8 s, here after referred to as steady state braking, or for short, steady state. When comparing differences between muscle activity in steady state braking and quiet sitting, i.e., when volunteers were stationary in the sitting position, the quiet sitting measurements represent the average muscle activity during a one second period half a second before the onset of the braking event.

III. RESULTS

All data are presented as average curves with corridors of one standard deviation (SD). Males and females are plotted separately. All measurements from one Autobrake PT trial with one of the male volunteers had to be removed as this volunteer intentionally pushed as hard as he could on the footwell during the trial. Therefore the output from this trial was deemed unrepresentative of the volunteer's response in pre-crash braking.

Kinematics

Kinematic corridors are presented in Fig. 4. The figure shows the average forward and upward displacements of head COG and T1. T1 displacements were available for 7 out of 11 male volunteers and all female volunteers. Head rotation is also presented in Fig. 4. Displacements were similar for males and females, in magnitude, as well as phase for both test cases, except for the head upward displacements and the T1 forward displacement in the Autobrake SB case. These differences were, however, not found to be significant in steady state braking at the 5% significance level. Average initial positions of the head and T1 are given in TABLE IV and displacements in the XZ coordinate system are plotted in Fig. 9 in the Appendix.

In the Autobrake PT, the head and T1 moved slightly backwards when the belt was pulled. In the Autobrake SB test case, i.e., with standard belt, no such backwards motion was noticed. Pre-tensing the belt resulted in significantly less steady state head and T1 forward displacements ($p<0.05$) for both genders (see Fig. 4a, b, g and h). Head rotations were, however, not affected.

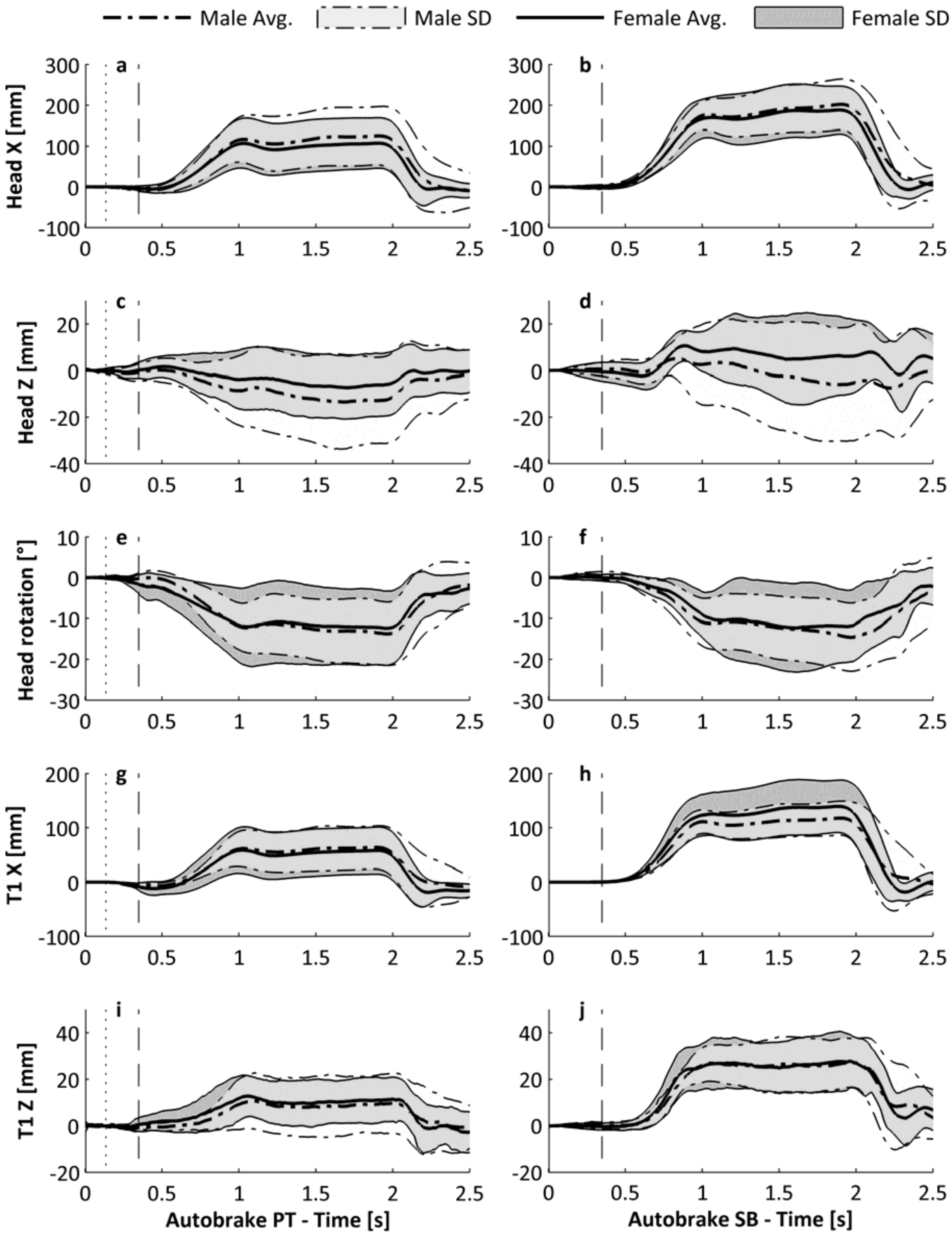

Fig. 4. Average ± SD head COG and T1 displacements, and head COG rotation in Autobrake PT (left) and Autobrake SB (right) for males (light grey, dashed dotted lines) and females (dark grey, solid lines). Vertical dashed lines represent acceleration onset and dotted lines show the belt pre-tension onset. For females *n* = 27 for plots *a – j*. For males *n* = 32 in plots *a, c, e*; *n* = 33 in plots *b, d, f*; *n* = 20 in plots *g, i*; *n* = 21 in plots *h, j*.

Belt Interactions and Footwell Forces

The belt was free to move before pre-tensioning of the belt. In the Autobrake PT test case, the shoulder belt force began to increase at the onset of the pre-tension, i.e., before the braking event started, and peaked at around 0.94 s for both genders, approximately 354 N for males and 296 N for females (Fig. 5). In the Autobrake SB case, shoulder belt forces were lower and started to increase approximately 320 ms after the acceleration onset, i.e., when the retractor locked. In Autobrake SB the first force peak was at approximately 0.93 s at 232 N for males and around 0.90 s at 165 N for females. Peak lap belt forces in Autobrake PT were 128 N for males and 110 N for females and in Autobrake SB, 89 N for males and 77 N for females.

The maximum amount of belt slack removed by the pre-tensioning system was not significantly different between genders at the 5% level. There was a statistically significant difference between slack removal in the first Autobrake PT trial and the two following trials (Fig. 6). On average the maximum slack removal was 66 mm (n = 19) in the first trials and 53 mm (n = 34) in the two subsequent trials (Fig. 6c). The amount of webbing paid out before the retractor locked in the Autobrake SB test cases was not significantly different between genders or trials at the 5% level. The average maximum pay-out was 41 mm (n = 52). Due to a measurement artifact, data in Fig. 6a-b are presented from 320 ms and onward. In Autobrake PT complete belt pay-out data were only available for 11 first trials and 6 trials for males and 6 for females in trials 2 – 3. In Autobrake SB 14 trials for males and 6 for females were available.

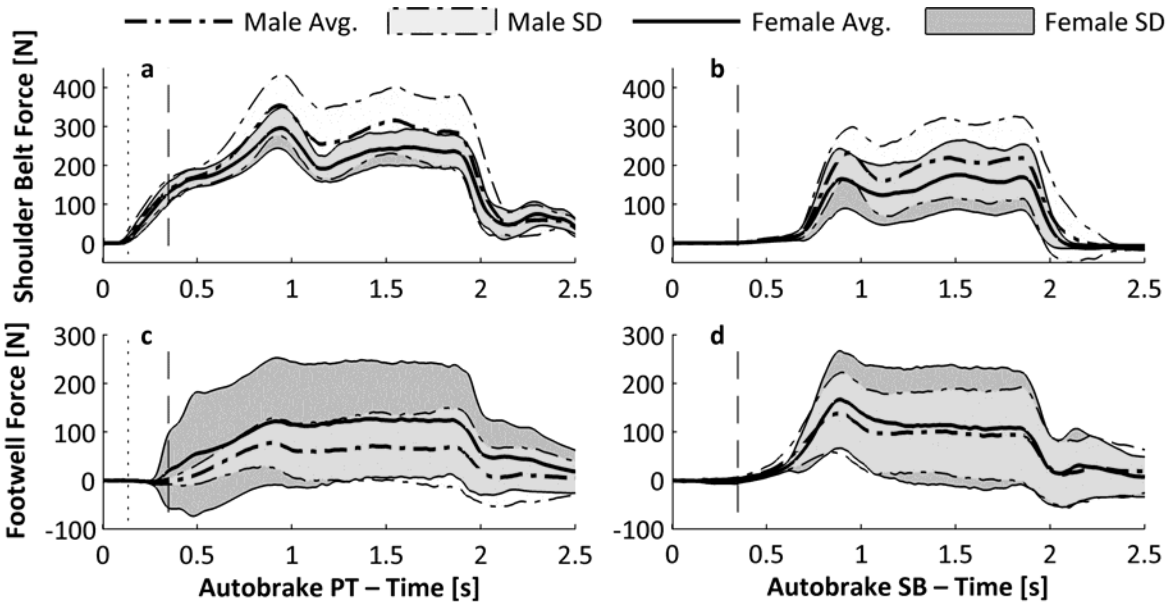

Fig. 5. Average ± SD shoulder belt and footwell force changes in Autobrake PT (left) and Autobrake SB (right) for males (light grey, dashed dotted lines, n = 32 in PT and n = 33 in SB) and females (dark grey, solid lines, n = 27). Vertical dashed lines represent acceleration onset and dotted lines show the belt pre-tension onset.

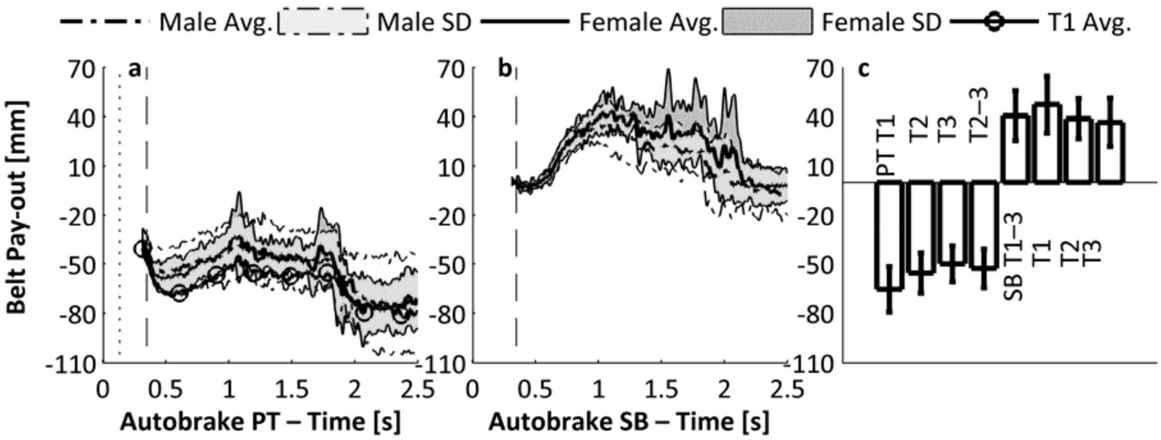

Fig. 6. Average ± SD belt pay-out in Autobrake PT (left) and Autobrake SB (middle) for males (light grey, dashed dotted lines) and females (dark grey, solid lines). Vertical dashed lines represent acceleration onset and vertical dotted lines show the belt pre-tension onset. Maximum ± SD belt pay-out for both genders (right), T1: Trial 1. In plot a the corridors are for trials 2-3 and the circled curve is for trial 1.

In Fig. 5 footwell force is plotted relative to the exerted force at trial initiation, see TABLE IV in the Appendix. Hence, negative values represent unloading of the footwell. In Autobrake SB footwell force increased soon after deceleration began for both genders. In the Autobrake PT case, the average force for the males was lower than in the Autobrake SB. The female response in Autobrake PT was noticeably different from the male response in magnitude and onset, and the spread was much greater for the females.

EMG Response

Average EMG data for the muscles on the left side of the body are presented in Fig. 7 and Fig. 8. For average right muscle activity in steady state braking, refer to TABLE II. In steady state, muscle activity was not significantly different between the left and right side except for BIC in males in Autobrake PT, CPVM and ADELT in females, in both test cases and SCM in females for Autobrake SB ($p<0.01$).

For females exposed to Autobrake PT, average muscle activity increased rapidly in all muscles after the retractor began to pull the belt (Fig. 7). Thereafter, and before the vehicle acceleration had fully ramped up, average female SCM, BIC, TRIC, ADELT and PDELT activity had already settled to less than 5% of MVC (%MVC). For the average male, no or slight increase (<7 %MVC) in activity of the EMG response was noted in these muscles. For both genders the cervical and lumbar extensors were highly active on average or between 14 – 35 %MVC in steady state, see TABLE II. Females had somewhat higher average activity levels than males in all muscles except for RA in both studied test cases.

In Autobrake SB, muscle activity increased soon after the vehicle deceleration began (Fig. 8). As in Autobrake PT, CPVM and LPVM muscles were the most active reaching levels between 16 – 36 %MVC in steady state. In contrast, all other muscles measured were relatively relaxed with average activity levels below 5 %MVC, except for left RA in males and right SCM in females.

All muscles were significantly more active during steady state braking in the Autobrake PT and Autobrake SB test cases compared to quiet sitting for both genders ($p<0.01$). In quiet sitting all muscles were activated less than 5 %MVC on average except for LPVM in both genders, see TABLE II.

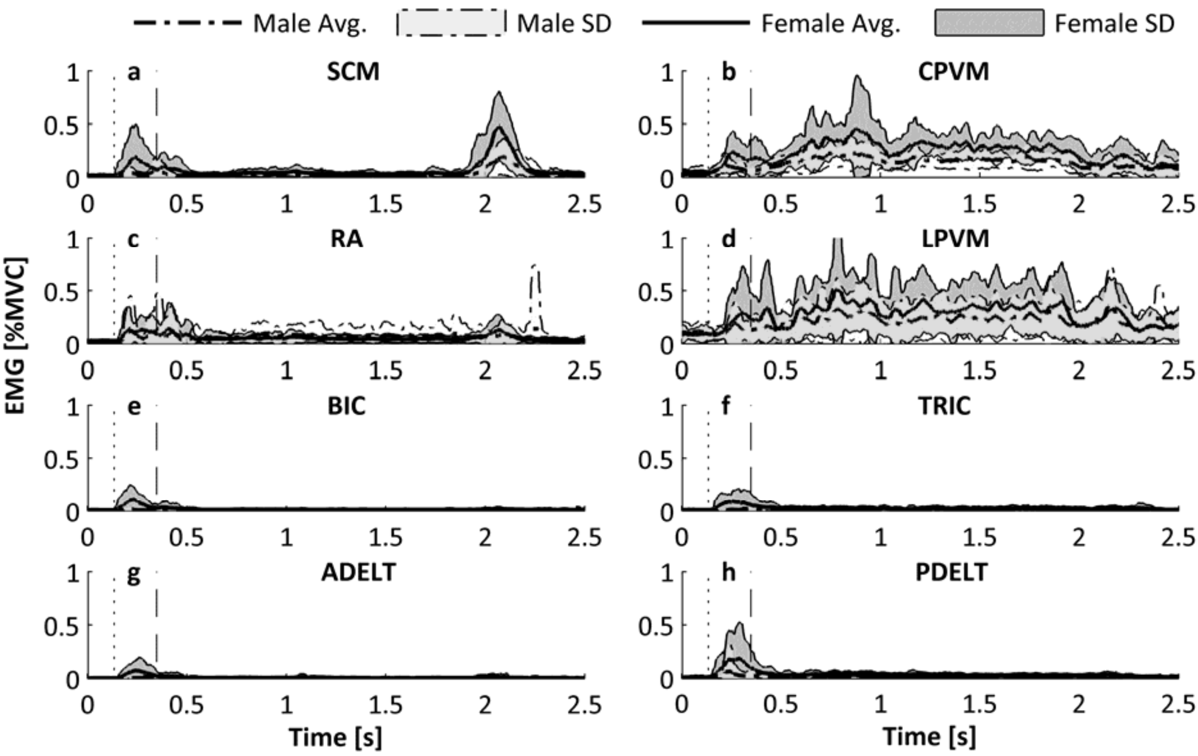

Fig. 7. Average EMG response normalised to MVC for left muscles in Autobrake PT for males (light grey, dashed dotted lines, *n* = 32) and females (dark grey, solid lines, *n* = 27). Vertical dashed lines represent acceleration onset and dotted lines show the belt pre-tension onset. SCM: sternocleidomastoid, CPVM: cervical paravertebral muscles, RA: rectus abdominis, LPVM: lumbar paravertebral muscles, BIC: biceps, TRIC: triceps, ADELT: anterior deltoid and PDELT: posterior deltoid.

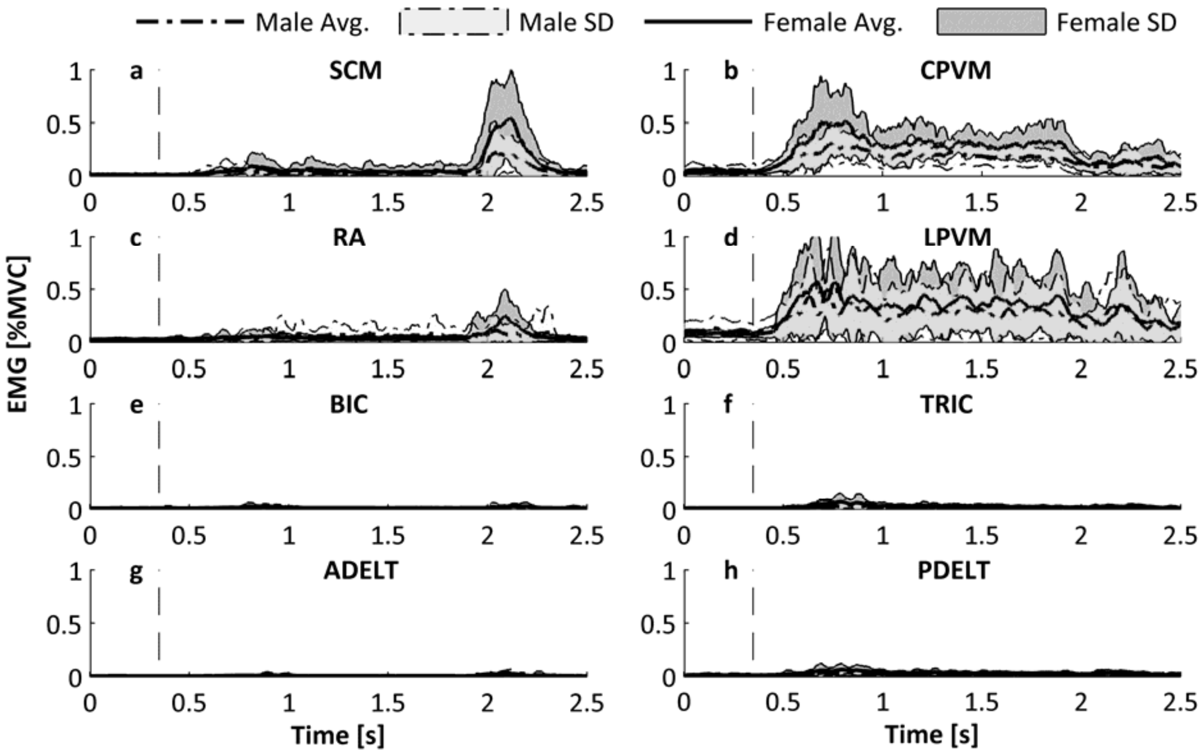

Fig. 8. Average EMG response normalised to MVC for left muscles in Autobrake SB for males (light grey, dashed dotted lines, *n* = 33) and females (dark grey, solid lines, *n* = 27). Vertical dashed lines represent acceleration onset. SCM: sternocleidomastoid, CPVM: cervical paravertebral muscles, RA: rectus abdominis, LPVM: lumbar paravertebral muscles, BIC: biceps, TRIC: triceps, ADELT: anterior deltoid and PDELT: posterior deltoid.

TABLE II
AVERAGE (SD) MUSCLE ACTIVITY AS %MVC IN STEADY STATE (1.6 – 1.8 S) FOR BOTH TEST CASES AND QUIET SITTING BEFORE BRAKING OCCURS (-1.5 – -0.5 S).

		Quiet Sitting				Autobrake PT				Autobrake SB			
Muscle	Side	Male		Female		Male		Female		Male		Female	
SCM	L	1.4	(1.3)	1.2	(0.7)	2.3	(1.5)	3.6	(2.7)	2.8	(2.5)	4.4	(6.5)
	R	1.6	(1.4)	1.6	(1.0)	2.7	(1.7)	4.8	(2.6)	3.3	(2.7)	6.6	(9.3)
CPVM	L	4.7	(4.1)	3.8	(1.6)	16.5	(7.2)	27.1	(8.9)	18.3	(7.5)	29.9	(12.0)
	R	3.3	(2.6)	3.3	(1.1)	14.2	(8.4)	19.9	(5.8)	16.2	(10.0)	21.4	(7.8)
RA	L	1.2	(0.9)	1.4	(0.6)	7.6	(9.6)	4.6	(3.2)	6.7	(10.1)	3.7	(1.6)
	R	1.1	(0.9)	1.4	(0.8)	5.9	(5.7)	4.0	(2.6)	4.9	(5.8)	3.5	(2.0)
LPVM	L	9.0	(8.5)	5.2	(6.1)	23.9	(17.2)	34.8	(19.4)	29.4	(24.6)	36.1	(24.4)
	R	11.1	(10.1)	6.6	(6.2)	21.2	(14.5)	32.2	(22.1)	28.0	(21.6)	32.7	(22.5)
BIC	L	0.4	(0.4)	0.5	(0.6)	0.9	(0.6)	1.3	(0.6)	0.8	(0.5)	1.3	(0.6)
	R	0.3	(0.3)	0.8	(2.4)	0.6	(0.5)	1.1	(0.5)	0.5	(0.3)	1.1	(0.4)
TRIC	L	0.5	(0.5)	0.6	(0.4)	1.0	(1.0)	2.2	(2.2)	1.5	(1.4)	2.1	(1.4)
	R	0.4	(0.5)	0.7	(0.6)	0.8	(0.6)	2.2	(1.1)	0.9	(0.9)	1.8	(0.8)
ADELT	L	0.3	(0.2)	0.4	(0.1)	0.8	(0.8)	0.9	(0.4)	0.6	(0.4)	0.9	(0.4)
	R	0.2	(0.1)	0.4	(1.4)	0.5	(0.2)	1.0	(1.5)	0.4	(0.2)	0.7	(0.4)
PDELT	L	0.9	(0.7)	0.9	(0.4)	2.4	(1.6)	3.2	(1.4)	2.4	(1.6)	3.0	(1.4)
	R	0.6	(0.6)	0.9	(0.9)	2.0	(1.2)	3.6	(1.6)	2.8	(2.1)	4.2	(2.1)

IV. DISCUSSION

In the present study, male and female front seat passengers were subjected to autonomous braking events where kinematic, EMG and boundary condition data were presented and analysed. Two test cases were studied, one with a standard belt and one with an active seat belt, equipped with a retractor to pre-tension the belt before the onset of deceleration. The analysis of the data from these tests resulted in corridors that are suitable for validation of active HBMs intended for safety system development and evaluation.

The use of a test vehicle in the present study, opposed to previous studies using a sled setup in a laboratory environment [5], [6], [10], [11], [13], [15–17], had both its advantages and disadvantages. To give the volunteers a mind-set resembling that of a person exposed to an actual pre-crash event, in order to reduce the influence of experimental circumstances on the volunteer response, test conditions were selected to closely resemble the target environment by performing the tests in regular traffic conditions. The vehicle environment limited what type of kinematic data could be acquired, i.e., the legs and hips of the passenger were obscured from most angles. The use of a car seat instead of a simplified rigid seat commonly used in laboratory set-ups ensured a biofidelic sitting posture. A simplified seat however, when compared to the more complex car seat, has proved to be more beneficial for model generation and validation. The interaction between the volunteers and the passenger seat was not as well defined as in a laboratory setting and may prove to be a challenge when validating the HBM and seat contact interaction. However, these drawbacks were deemed acceptable.

A single vehicle model, a Volvo V60, was used in this study. Nevertheless, the leather seat and the belt are representative of premium passenger cars. When loaded in quasi-static conditions the stiffness of the rear portion of the seat was approximately 21 N/mm, while a seat with textile upholstery is less stiff at approximately 13 N/mm [23]. It is not anticipated that a seat of different stiffness will change the kinematics of the occupant considerably. The speed of the belt retractor may change the occupant reflex muscle response, here the initial belt force increased by approximately 530 N/s inducing reflex muscle reactions in female but not male volunteers on average. Females are thus expected to react to belt pre-tension by a reflex response in vehicles with a similar or more aggressive retractor, while it is not known at what lower threshold the reflex response would disappear. Similarly, males may show reflex responses if exposed to a more aggressive retractor.

The deceleration was 1.1 g for both test cases, a more prominent braking than most currently available commercial autonomous braking systems apply (0.4 - 0.5 g) [24], [25]. Driver initiated emergency braking events reach up to approximately 1 g with a deceleration rate that is higher than that of the autonomous braking events in this study [2], [18], reaching 1 g in approximately 200 ms compared to 600 ms. The slower acceleration ramp up phase provides more time for muscle contractions. Consequently, muscle activity is expected to be different in this phase. After ramp up, however (approximately 800 ms), similar activity levels, and subsequently kinematics, are expected. Future autonomous braking systems are likely to have deceleration levels and rates approaching those in driver braking events, as auto braking technology advances [26]. Therefore, it seems reasonable that the acceleration level in this study is adequate for the validation of HBMs aimed at evaluating and improving future pre-crash systems.

The data were acquired in the vehicle fixed coordinate system. According to Volvo chassis engineers, the test vehicle can pitch up to 1.7° in full braking. The rotation of the coordinate system was considered to have minor influences on the measurements and was not compensated for in the presented data.

Muscle habituation effect was not examined in detail in the present study. All volunteers initially underwent 20 braking events in the driver's seat, 12 of which were analogous to the braking events they were exposed to in the passenger seat. Bluoin et al. [27] reported that volunteers exposed to repeated analogous acceleration pulses modified their EMG responses, adopting a feed-forward strategy so as to minimise EMG activity. However, a preliminary analysis on increased head forward displacement, which might be expected in decreasing muscle activity in subsequent trials, showed no such general trends. That also indicates that potential muscle fatigue had a limited influence on the results.

The active belt affected the average passenger kinematic response. Pre-tension of 170 N prior to the brakes being applied in the vehicle, resulted in significantly less forward displacement of the head and T1 ($p<0.05$). Male volunteers (7 out of 11) experienced on average a 66 mm less forward head displacement in steady state when restrained with the reversible pre-tensioned belt compared to the standard belt. For females, the differences were more pronounced, on average 81 mm less forward head and T1 displacements. The difference in T1 displacement for males was 52 mm on average which is comparable with the findings of Schoeneburg et al. [2] who reported 46 mm less median forward neck displacement in emergency braking (1 g) with a reversible pre-tensioned belt.

The belt slack removed by the retractor was significantly different ($p<0.05$) in the first Autobrake PT trials compared to the two subsequent trials. In the first Autobrake PT trial initial slack was removed which was not paid out again when the belt was released. Thus for the subsequent trials, the belt was more tightly positioned on the volunteers. To ensure uniform belt slack conditions for the Autobrake SB test case, each test started with one Autobrake PT trial and all subsequent trials were randomised test cases. The difference in initial slack did not affect the passenger kinematics substantially in the Autobrake PT test case. Schoeneburg et al. [2] reported 26 mm slack removal (belt pay-in) and 21 mm belt pay-out with and without the reversible pre-tensioned belt, respectively, which is considerably lower than reported in the present study. Possibly, the lower deceleration rate in the present study (23 $m/s^2/s$ compared to 62 $m/s^2/s$ in [2]) caused the retractor to lock more slowly, in turn allowing more webbing to pay out. Furthermore, the fact that the retractor was activated 200 ms prior to braking may explain the larger belt pay-in values.

The footwell and seat belt forces and the belt pay-out corridors primarily provide boundary conditions for the validation of the interaction between an HBM and the car interior model. The change in footwell force depends on inertia effects during deceleration, muscle activity in the lower extremities and to a small extent shifts in sitting posture. Negative force changes were registered for a few volunteers, explained by muscle contractions reducing the interaction between the foot and floor. It is therefore important when validating HBMs to simulate the initial footwell forces (TABLE IV in Appendix). The footwell forces show little gender dependence. It should be noted that the gender differences seen in the Autobrake PT test case is mainly due to the response of a single female volunteer who was extremely startled by the belt pre-tension in all trials, resulting in a footwell force onset approximately 200 ms earlier than the average female and at an average force around 400 N. However, that particular volunteer's kinematic and EMG responses were not exceptionally different from other volunteers in the same trials. This corridor should thus be treated with caution as it includes a potential outlier.

Significant increase ($p<0.01$) in average muscle activity due to the change in loading from 0 to 1.1 g was detected in all muscles. Neck and lumbar extensors recorded had the highest activity levels, 14 – 36 %MVC. These muscles become active to restrict the forward motion caused by the inertial loading. Furthermore, antagonist co-contraction was noticed in the females in RA in particular as well as SCM. Despite the significant increase, activity levels remained below 3 %MVC in the upper extremities, except for PDELT in females (≤4.2 %MVC). As passengers do not have the opportunity to brace their arms against a steering wheel, for instance, muscle activity in the upper extremities play a limited role in passenger response.

Looking more closely at neck extensor activity levels, it was observed that the left neck extensors had higher activity levels, on average, compared to the right ones, although, this difference was only significant for females ($p<0.01$). These results are contrary to the findings of Kumar et al. [17] who reported the same average activity levels for the left and right neck extensors (C4 level) in a 1.4 g frontal sled impact, where volunteers were unaware of the imminent impact. However, the volunteers were restrained by a four-point seat belt, providing symmetrical loading. Thus, it is speculated that the differences in activity between the left and right side muscles is an effect of the unsymmetrical loading of the three-point belt. The right upper part of the torso receives more support from the belt and therefore muscle activity of the left sided muscles needs to increase in order to keep the head from turning. Assuming symmetrical muscle activity for all neck muscles might thus be a crude estimation in case of a three-point belt loading.

In Autobrake PT, muscle activity was related to the onset of the belt pre-tension. Muscle activity increased soon after the belt was pulled and before acceleration started to ramp up. Volunteers, in particular females, responded to the active belt by contracting certain muscles for a short period as if they were startled by the belt pulling. As braking events were initiated by pushing a button in the rear seat, volunteers were not given any visual cues as to the onset of the braking event. The startle response can thus be related either to auditory reflexes to the sound released from the retractor when it was activated or somatosensory reflexes from the shoulder belt pressing against the volunteer's upper body. Acoustic signals can evoke neck muscle responses, in the presence of motor readiness after 58 ms [28] although according to [29] the sound needs to be above 80 dB to cause startle muscle reflexes. The active belt retractor produced a sound below 80 dB and therefore seems unlikely to have evoked a significant startle response. Whether volunteer kinematics between pre-tension and acceleration onset are influenced by the startle response is unclear. Alternatively, it is likely that kinematics at that moment are influenced by the belt pulling the volunteers backwards. In Autobrake SB muscle activity began after the acceleration onset at a relatively slower rate than in Autobrake PT. The breaking acceleration ramp up is a slower process than the belt pre-tension, therefore the muscle contractions in Autobrake SB are unlikely to be startle reflexes.

Information on muscle activity in quiet sitting is important for initialisation and posture control of active HBMs intended for pre-crash simulations. Data in this study indicate the level of muscle activity needed, in the measured muscles, to maintain sitting posture for passengers during driving. Muscles which do not contribute to postural control in sitting, such as neck and lumbar flexors and upper extremity muscles displayed almost no activity (≤1.6 %MVC on average). Neck and lumbar extensors however, were more active, in particular the lumbar extensors. High lumbar extensor activity may be associated with electromagnetic interference as the lumbar EMG signals were the weakest of the recorded signals, or pressure artifacts from the backrest contact. Muscle activity levels in quiet sitting have been presented in the literature [30] for slumped and erect sitting postures on an experimental chair. In ten male volunteers the average activity was 4.8 ± 3.9 %MVC in the RA and 5.4 ± 3.4 %MVC in the lumbar extensors. In the present study, males displayed relatively lower activity levels in the RA (1.1 %MVC) and somewhat higher for lumbar extensors (11.1 %MVC). These differences may be due to several reasons. Watanabe et al. measured activity in lumbar extensors at the L3 level, as opposed to the L1 level in the present study. Furthermore, the incline of the experimental chair backrest was 10° compared to 22° backrest inclination of the vehicle seat. According to Andersson and Örtegren [31] muscle activity is lower at L3 compared to L1 in a reclined sitting posture. However, the study also concluded that a backrest at an increased angle resulted in decreased activity at both L3 and L1 levels. The level of difference found in lumbar extensor activity is thus somewhat surprising. Nevertheless, it should be considered that the test set up in [30] was a stationary chair inside a lab environment while the present study was recorded under dynamic conditions, and the activity levels may have been affected during the quiet sitting by potential vehicle jerks and vibrations related to the road conditions. Szabo et al. [32] presented activity levels for SCM, cervical and lumbar extensors for four male subjects sitting in a relaxed position, in a stationary vehicle, whilst holding the steering wheel. SCM and cervical extensors showed an average activity of 1.2 and 3.5 %MVC respectively, which is similar to the findings in the present study (1.6 and 3.3 %MVC). Average activity levels in the lumbar extensor was 1.9 %MVC which is relatively low compared to both [30] and the levels presented here. In [32] MCVs for the lumbar extensors were performed in a prone horizontal position, not a sitting posture which may have affected the results. Furthermore, it is unclear at which lumbar vertebral level these measurements were taken. The observed differences indicate that vibrations and vehicle jerk during driving may influence the muscle activity and should be accounted for. This study is, to the best of the authors' knowledge, the first study to present EMG data for vehicle occupants, when seated in a passenger seat during driving, that is normalised with MVCs performed in a similar posture.

V. CONCLUSIONS

This study presents passenger volunteer kinematic, EMG and boundary condition data for validation of active HBMs restrained with standard and reversible pre-tensioned belts in pre-crash situations. Pre-tensioning the belt 200 ms prior to the braking event proved to affect the volunteers' kinematic responses, with statistically significant reduction of head and T1 forward displacements. Muscle activity was induced by the belt pre-tension, and not the acceleration pulse, as in the standard belt case.

During the braking event, cervical and lumbar extensor muscles, restricting the forward motion due to inertial loading, displayed the highest activity levels. Antagonist co-contractions were also observed. In quiet sitting during driving, activity levels were below 5 %MVC in all muscles, excluding lumbar extensors. Activity levels in quiet sitting along with initial positions and interaction forces provided the initial boundary conditions needed for an active HBM simulation set up.

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VIII. APPENDIX

TABLE III

ELECTRODE PLACEMENT DESCRIPTION. ALL ELECTRODES WERE POSITIONED ON THE MUSCLE BELLY (EXCEPT THE REFERENCE ELECTRODE).

Function	Muscle	Electrode placement	Ref.
Cervical flexion	Sternocleidomastoid (SCM)	1/3 length between the suprasternal notch and the mastoid process.	[33]
Cervical extension	Cervical paravertebral muscles (CPVM)	C4 level, 20 mm lateral of the midline	[27]
Lumbar flexion	Rectus abdominis (RA)	Above and below the umbilicus, approx. 20 mm lateral of the midline.	[34]
Lumbar extension	Lumbar paravertebral muscles (LPVM)	L1 level, 2 fingers lateral of the midline.	[35]
Elbow flexion	Biceps brachii (BIC)	1/3 length from the distal attachment and the medial acromion.	[35]
Elbow extension	Triceps brachii (TRIC)	1/2 length between the olecranon and the crista of the acromion.	[35]
Shoulder flexion	Anterior deltoid (ADELT)	1 finger breadth distal and anterior of the acromion.	[35]
Shoulder extension	Posterior deltoid (PDELT)	2 fingerbreadths behind the angle of the acromion.	[35]
Reference	(REF)	On top of the anterior superior iliac spine.	

TABLE IV

AVERAGE (SD) HEAD AND T1 POSITIONS, HEAD ANGLE AND FOOTWELL FORCES AT TIME ZERO. POSITIONS ARE RELATIVE TO THE ORIGIN OF THE VEHICLE FIXED COORDINATE SYSTEM WHERE THE ORIGIN IS 792 MM REARWARD AND LEVEL WITH THE HEAD OF THE FRONT RIGHT BOLT HOLDING THE DRIVER'S SEAT.

	Male	Female
Head X [mm]	209 (32)	209 (16)
Head Z [mm]	847 (18)	805 (32)
Head Angle [deg]	10 (6)	6 (6)
T1 X [mm]	137 (15)	146 (14)
T1 Z [mm]	696 (16)	669 (22)
Footwell Force [N]	62 (17)	42 (13)

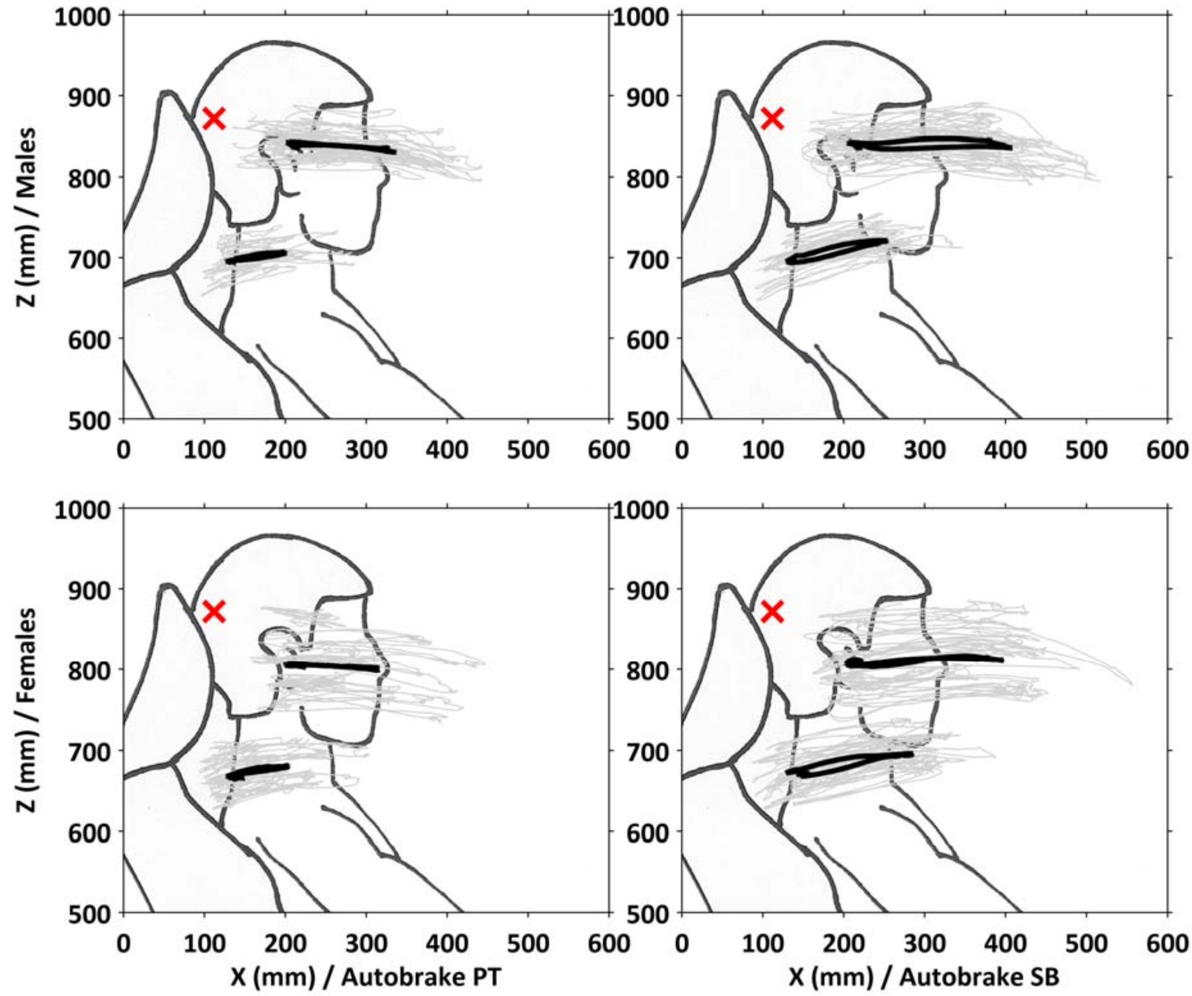

Fig. 9. Individual (grey) and average (black) head and T1 kinematic trajectories in Autobrake PT (left) and Autobrake SB (right) plotted in the vehicle coordinate system. The origin is 792 mm rearward and level with the head of the front right bolt holding the driver's seat. The cross represents the position of the D-ring. The background image is of a 50th percentile male in relative proportions to the data presented.