Gender Differences in Occupant Posture and Muscle Activity with Motorized Seat Belts

GENDER DIFFERENCES IN OCCUPANT POSTURE AND MUSCLE ACTIVITY WITH MOTORIZED SEAT BELTS

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ABSTRACT

The aim of this study was to assess gender differences in the posture and muscular activity of occupants in response to pretension from motorized seatbelts. Male and female vehicle occupants were tested in both front seat positions during normal driving and autonomous braking. This data is useful for the development of human body models (HBM), and increases the understanding of the effects of motorized belts.

Kinematics and electromyography (EMG) were analyzed for 18 volunteers (9 male, 9 female) subjected to autonomous braking (11 m/s^2 deceleration) during real driving on rural roads. Two restraint configurations were tested: a standard belt and a motorized belt, activated 240 ms before the initiation of braking. Statistical comparison of volunteers' posture and normalized EMG amplitudes was performed to understand differences incurred by the motorized belts, as well as to compare response across gender and role (occupant position within the vehicle). Data was analyzed both prior to and at vehicle deceleration, which occurred 240 ms after motorized belt onset.

Motorized belts significantly affected all postural metrics, and significantly elevated the activity of all muscles compared to typical riding. Though increases in muscle activity were small at deceleration onset compared with typical riding for male occupants and female passengers, female drivers demonstrated significantly larger increases in muscular activity: between 5 and 13% of the maximum voluntary contraction (MVC). At deceleration onset, standard belts showed little change in posture or muscle activation, with the median changes being well within the ranges exhibited during typical riding for all groups (i.e. not distinguishable from typical riding). Typical riding postures of males and females were similar, as were muscular activation levels—generally less than 5% of the MVC. However, drivers exhibited significantly higher muscular activity in the arm and shoulder muscles than passengers.

Limitations include the repeated nature of the testing, as prior work has shown that habituation across trials alters occupant response compared to that of unaware occupants. However, randomization of the trial order helped mitigate potential habituation effects. Another limitation is the sample size of 18 volunteers.

An important finding of this study is that the increase in occupant muscular activation seen with motorized belts was gender-specific: at deceleration, the change in activation of most muscles was significantly different across gender and belt type, with female drivers exhibiting larger increases in muscular activation than male drivers or passengers of either gender, particularly in the arm muscles. These activations appeared to be startle responses, and may have implications for interactions with the steering wheel and motion during a braking or crash event. This warrants further studies and stresses the importance of quantifying male and female subjects separately in future studies of pre-crash systems.

Keywords: Gender; driver, passenger; reversible, motorized seat belt; braking; EMG

INTRODUCTION

Vehicle safety has improved significantly since the 1950s. Early interventions were in-crash systems; for example, improved vehicle structures and occupant restraints. The three-point seat belt was first introduced in 1957, and reported injury reduction was 40–90% (Bohlin 1967) and more than 35% (Norin et al. 1984). Later, Cummings et al. (2003) estimated a 61% lower risk of death for belted front seat occupants compared to unbelted occupants, based on accident data from 1986–1998. More recently, accident avoidance and pre-crash systems that avoid or mitigate the severity of accidents have been implemented on a large scale. Autonomous braking systems have been introduced (Coelingh et al. 2007; Distner et al. 2009; Schittenhelm 2009). Autonomous braking is beneficial by reducing the vehicle's kinetic energy. However, for impacts with equivalent speed, pre-crash braking increases chest deflection and belt forces (Antona et al. 2010), indicating that injury prevention can be improved with integrated safety systems that reposition the occupant before the impact. One example is autonomous braking in combination with reversible seat belt pretension, which removes belt slack and secures the occupant prior to an impact (Schöeneburg et al. 2011). Motorized seat belts provide reversible pretension before the crash and have the potential to reposition the occupant to an optimal pre-crash position.

Reversible seat belt pretension has been studied in combination with driver emergency braking (Tobata et al. 2003), lateral maneuvers (Mages et al. 2011), pre-impact braking (Woitsch and Sinz 2014), and stationary conditions (Good et al. 2008a; Good et al. 2008b; Develet et al. 2013). In these studies, volunteers or anthropomorphic test devices (ATDs) were used. ATDs have severe limitations as they are too stiff to represent relaxed occupants in low loading conditions (Beeman et al. 2012). Increasingly, computational human body models (HBMs) are used in vehicle safety assessment. Currently available HBMs have mainly been developed and validated in the crash loading regime. Active HBMs, which include occupant muscle response and are validated for braking and similar scenarios (Östh et al. 2015), have the potential to become strong tools for the development and assessment of integrated safety systems. However, the active HBM of Östh et al. (2015) represents the 50th percentile male and there are no active HBMs representing female occupants. Therefore, to study gender differences in occupant response with integrated safety systems, volunteers are the best option to date. With increased knowledge from volunteer experiments, future HBMs can be developed to represent gender differences and study integrated safety systems for the full sequence from pre-crash to crash.

The aim of this study was to analyze how pretension with motorized seatbelts changed occupant posture and muscle activity, and to assess differences in response based on gender and occupant position in the vehicle. This was done with statistical analyses of volunteer data from drivers and front seat passengers with and without motorized seat belt pretension prior to autonomous braking interventions. The results have the potential to enhance traffic safety for all occupants, males and females, by providing an increased understanding of how motorized belts affect occupants and by providing data for enhancement of simulation tools, such as HBMs representing both genders.

METHODS

This study analyzed data from volunteer experiments, approved by the Ethical Review Board at the University of Gothenburg, Sweden, where 20 volunteers were exposed to 29 braking interventions (11 m/s^2) as drivers (Östh et al. 2013) and passengers (Ólafsdóttir et al. 2013). Volunteers with incomplete data sets were excluded, and therefore 9 females and 9 males were incuded in this study. For each volunteer, 18 braking interventions were analyzed (12 driver, 6 passenger). The tests were conducted in a passenger car equipped with a motorized belt. Interventions were performed in a randomized order with two seat belt retractor configurations: a standard configuration that locked at 4 m/s^2 vehicle deceleration or when the belt pay out acceleration was 15 m/s^2 (subsequently denoted "standard"), and a reversible configuration where the electrical motor provided 170 N of belt pretension force at an approximate maximum retraction speed of 300 mm/s (subsequently denoted "motorized"). Each braking intervention was triggered without prior notification to the volunteer. Data were analyzed for two different time periods, before triggering the intervention (termed "typical riding") and at the onset of vehcile deceleration (termed "initial braking"). Vehicle deceleration occurred on average 350 ms after triggering; for trials with the motorized belt, pretension occurred on average after 110 ms.

Prior to testing, volunteers' anthropometry was measured (Table 1). Sitting height was the distance between the superior aspect of the head and the seated surface in the mid-sagittal plane, with volunteers sitting on a stool (Schneider et al. 1983). Surface electromyography (EMG) electrodes were applied bilaterally to the volunteers (Figure 1). Volunteers were positioned in a Maximum Voluntary Contraction (MVC) rig, designed to provide a posture that resembled the driving position, and three repetitions of MVC were performed for each tested muscle (Östh et al. 2013). Within the test vehicle, volunteers could partially adjust the driver seat and steering wheel to find a comfortable driving position and were told to keep their hands symmetrically on the steering wheel. Allowed adjustments were translation of the seat, change of the inclination angle of the seat back, and steering wheel position and angle. The passenger seat was fixed (in the mid fore/aft position with a seat back angle of 22°), and volunteers were instructed to keep their feet symmetrical to the midline of the footwell and rest their hands on their lap.

Surface EMG was recorded with a sampling rate of 2048 Hz using a Compumedics Grael (Compumedics, Abbotsford, Australia) for eight muscle groups: sternocleidomastoid, cervical paravertebrals, rectus abdominis, lumbar paravertebrals, biceps brachii, triceps brachii, anterior deltoid, and posterior deltoid (Figure 1). EMG data were normalized using MVC. In typical riding, EMG data were averaged over a one-second interval, from 1.5 to 0.5 s before trigger. In initial braking, EMG data were averaged over a 20 ms interval starting at deceleration onset. The change in activation in initial braking compared to typical riding was investigated per trial as the the absolute increase or decrease in activation. Then, the median change across trials of a given condition was used to represent each volunteer's response.

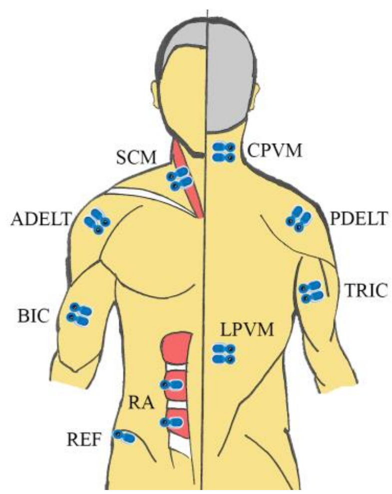

Figure 1. EMG electrode placement (anterior locations on left, posterior locations on right). SCM: sternocleidomastoid; CPVM: cervical paravertebrals; ADELT: anterior deltoid; PDELT: posterior deltoid; BIC: biceps brachii; TRIC: triceps brachii; LPVM: lumbar paravertebrals; RA: rectus abdominis; REF: reference electrode. Adapted from Östh et al. (2013) by permission of The Stapp Association.

Kinematic data was acquired at 50 Hz through film analysis (TEMA Automotive, Image Systems, Linköping, Sweden). Posture was measured with video tracking of markers on the volunteer's head and chest. In the present study, the head center of gravity (CG) position was calculated from film markers close to the ear and eye (Östh et al. 2013). Kinematic posture data were collected in a vehicle-fixed coordinate system, with positive X forward and positive Z upward (Figure 2). Head rotation was the angle between the horizontal plane and the Frankfort plane, in the sagittal plane, with positive rotation representing extension. Relative head-to-sternum distance was the difference between the calculated head CG and the chest marker, effectively measuring the posture of the neck: an individual with a larger head-to-sternum X in one position compared to another would have a more retracted posture in the former. Head-to-head restraint distance was the difference between the head CG and the mid-point on the anterior surface of the head restraint, in the mid-sagittal plane. For typical riding, kinematic data was analyzed and defined as the average value over the first 100 ms after trigger. The median across all trials of a given condition was used to represent each volunteer's response. For initial braking, the postural metrics were collected at deceleration onset. For each trial, the volunteer's change in metric was taken as the difference between the response at deceleration onset and at typical riding, with the median change across all trials of a given condition representing each volunteer's response.

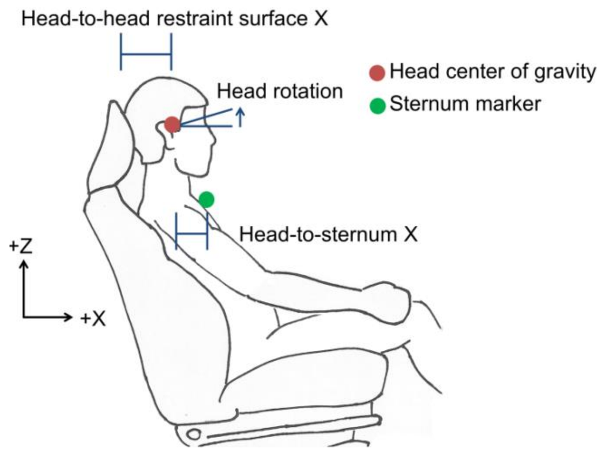

Figure 2. Schematic of postural metrics used in this study.

Statistical analysis included a normality and homoscedasticity assessment and a general test for group differences: either a parametric repeated-measures ANCOVA, with sitting height as a covariate, or a nonparametric Friedman test (when data did not meet normality or homoscedasticity assumptions). A 5% significance level was used. For typical riding, two-way repeated measures designs were used, with factors being role (driver or passenger) and gender. For initial braking, three-way repeated measures designs were used, with belt type (motorized or standard) as an additional repeated measure. To compensate for the seat belt loading asymmetry, outboard muscles of the drivers and passengers were compared, and likewise for inboard muscles. Data processing was performed in MATLAB (Version 8.0.0, MathWorks, Natick, MA), while statistical analyses were implemented in SAS (Version 9.3, SAS Institute Inc., Cary, NC).

RESULTS

The volunteer sitting height was normally distributed (919 ± 39 mm). A post-hoc t-test indicated a significant difference ($p < 0.05$) in male and female sitting height, with females (894 ± 33 mm) being shorter than males (945 ± 27 mm). Table 1 lists the average anthropometric measurements for females and males.

Table 1.
Volunteer anthropometric data. SD: Standard deviation.

Volunteers	Age (years)	Height (cm)	Weight (kg)	Sitting Height (mm)
Male Mean (SD)	34.1 (12.7)	178.1 (4.8)	77.2 (5.8)	945 (27)
Female Mean (SD)	28.8 (5.9)	166.6 (5.0)	59.4 (5.2)	894 (33)

Motorized belts significantly affected all postural metrics (Figure 3). The occupant position at initial braking with motorized belts is illustrated in Figure 4 for one male and one female volunteer. Notably, the median sternal X location was shifted posteriorly by more than 10 mm (depending on gender and role) with motorized belts, and the median head-to-sternum X distance was reduced by 12 to 13 mm for females (depending on role) and by 6 to 7 mm for males. The lower values for males may be due to the higher inertia (mass) of male volunteers compared to female volunteers. The range (difference between the 75th and 25th percentiles) of change in postural metrics seen with standard belts during initial braking was well within the range exhibited during typical riding, for all comparable metrics, indicating that there were no other factors except the belt influencing the occupant response. The typical riding position did not seem to depend on gender, as the only significant effect of gender was found for the head-to-sternum X, which also showed significant covariance with sitting height. Role had a significant effect on the head-to-sternum X, indicating differences in neck curvature. Passengers had more than 20 mm larger median head-to-sternum X distances than drivers.

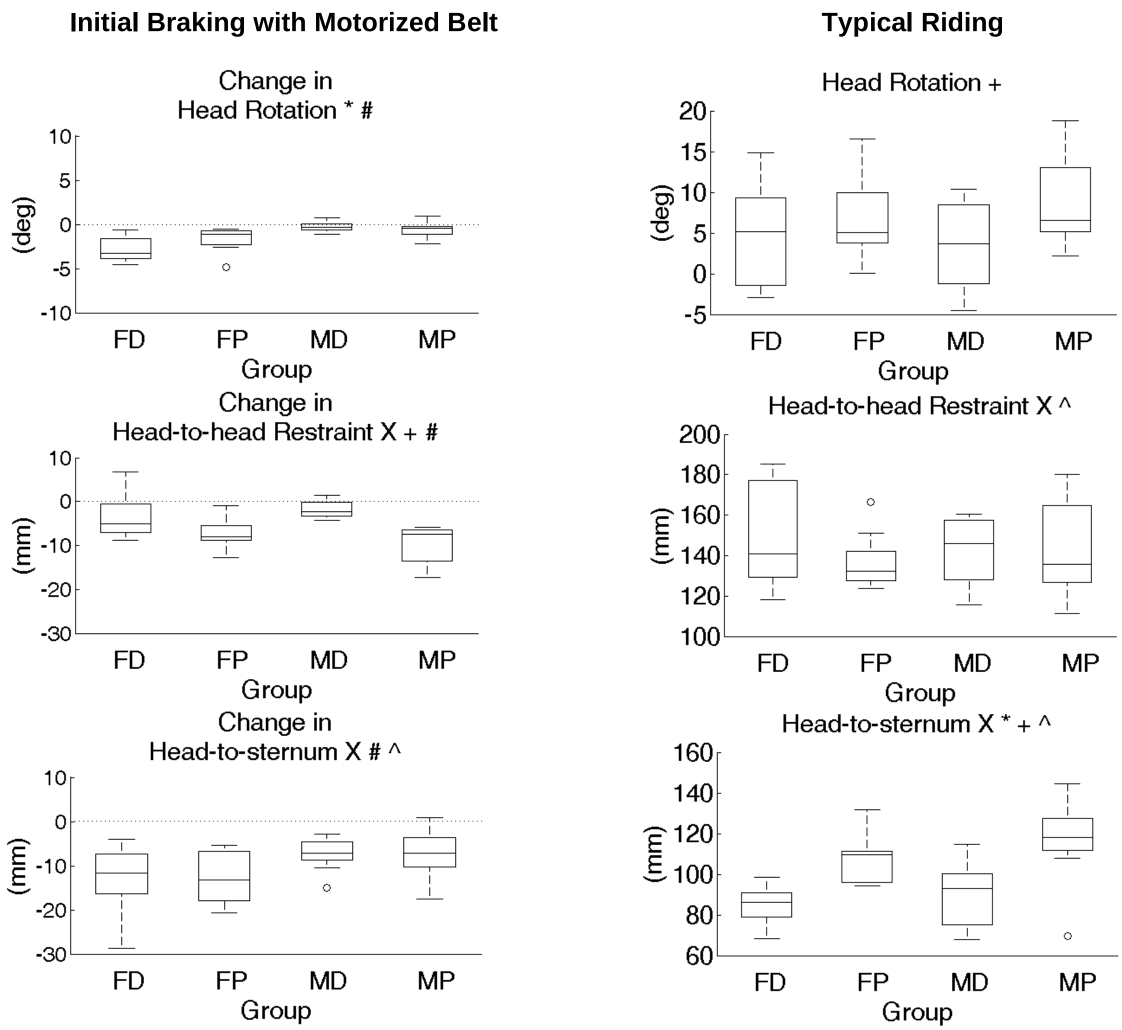

*Figure 3. Median change in postural metrics with motorized belts for initial braking compared to typical riding (left) and median postural metrics during typical riding (right). The interquartile ranges are indicated with boxes and outliers by circles. All head metrics are defined at the head center of gravity. FD: female drivers; FP: female passengers; MD: male drivers; MP: male passengers. *=significant main effect of gender, +=significant main effect of role, #=significant main effect of belt type, ^=significant covariance with sitting height.*

Figure 4. Pictures of one female (left) and one male (right) volunteer in the driver (top) and passenger (bottom) position during initial braking with a motorized seatbelt (~250 ms after belt pretension).

Gender differences were not found for the muscle activity during typical riding (Table 2). Muscle activity was not normally distributed, and was analyzed nonparametrically. Drivers exhibited significantly higher acivity in the arm and shoulder muscles than passengers. Though nonsignificant, this trend was evident for most neck and trunk muscles. Notably, while muscles generally displayed interquartile ranges of less than 5% MVC, the lumbar paravertebral muscles had muscle activity up to 20% MVC. With standard belts, the median changes in muscle activity during initial braking were well within the ranges during typical riding for all groups (typically <1% MVC, LPVM <5% MVC), in line with the postural metrics.

Motorized belts significantly increased muscle activity for all muscles (Figures 5 and 6, Table 3). Gender generally displayed significant effects on the arm, shoulder, and trunk muscles, with females showing larger changes than males. The increase in muscle activity was generally small (less than 5% MVC), except for female drivers. In the group of female drivers, for the measured muscles activity increased by 5-13% MVC, and inboard muscles typically displayed slightly larger increases in activity than outboard muscles. With motorized belts, a trend of drivers displaying larger median changes in muscle activity than passengers was seen for most muscles. The lumbar paravertebral muscle had higher increases in activity with motorized belts compared to the other muscles: 9-11% MVC for female drivers, 7-11% for female passengers, 6% for male drivers, and 2-5% for male passengers (Figure 6). Role significantly affected the outboard and inboard cervical paravertebral muscles, with drivers showing larger changes than passengers (Figure 6). Role also significantly affected the outboard rectus abdominis and inboard anterior deltoid.

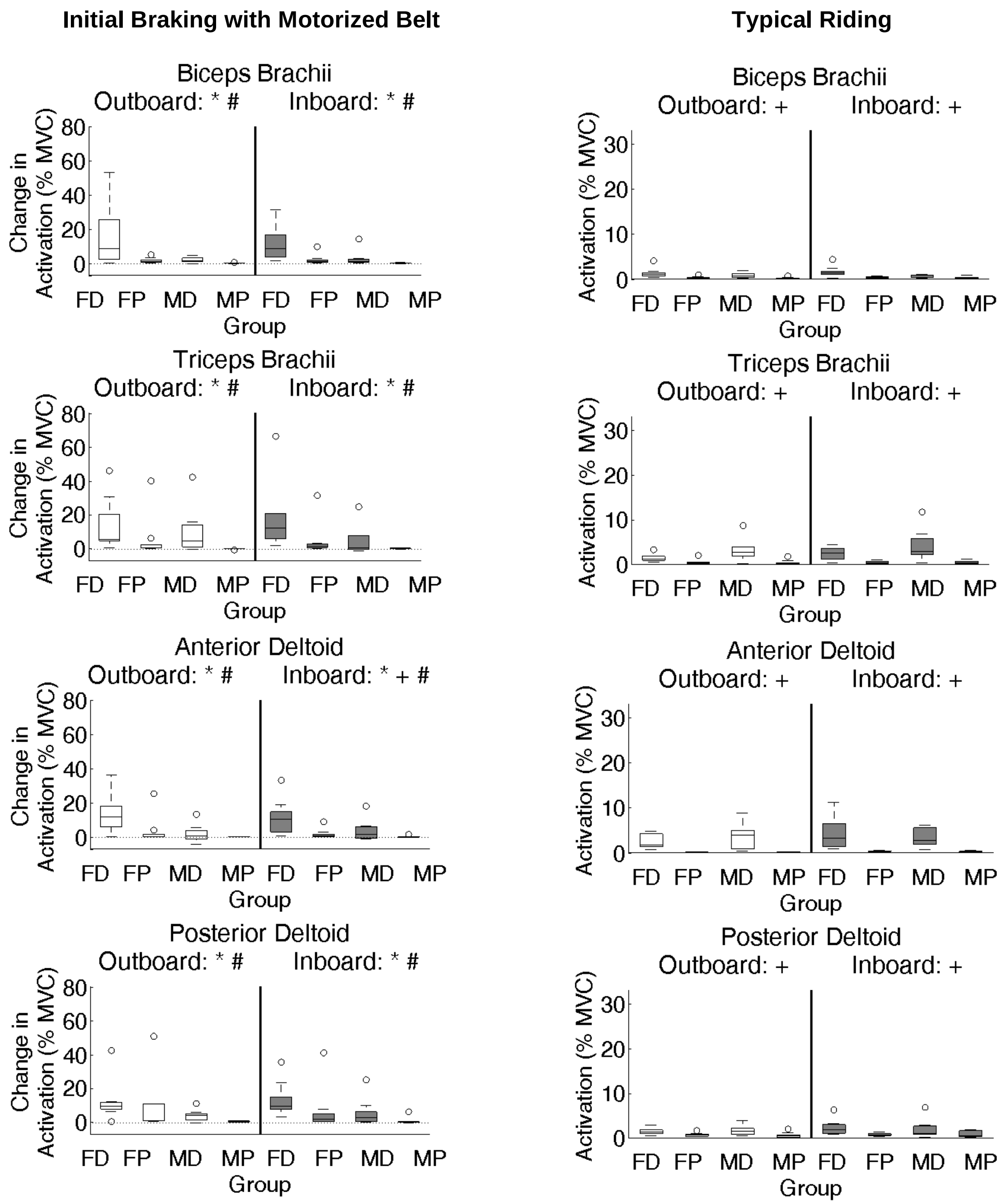

Figure 5. Median change in arm and shoulder muscle activity with motorized belts for initial braking compared to typical riding (left) and median muscle activity during typical riding (right). The interquartile ranges are indicated with boxes (white for outboard muscles, grey for inboard muscles) and outliers by circles.

FD: female drivers; FP: female passengers; MD: male drivers; MP: male passengers.

****=significant main effect of gender, +=significant main effect of role, #=significant main effect of belt type.***

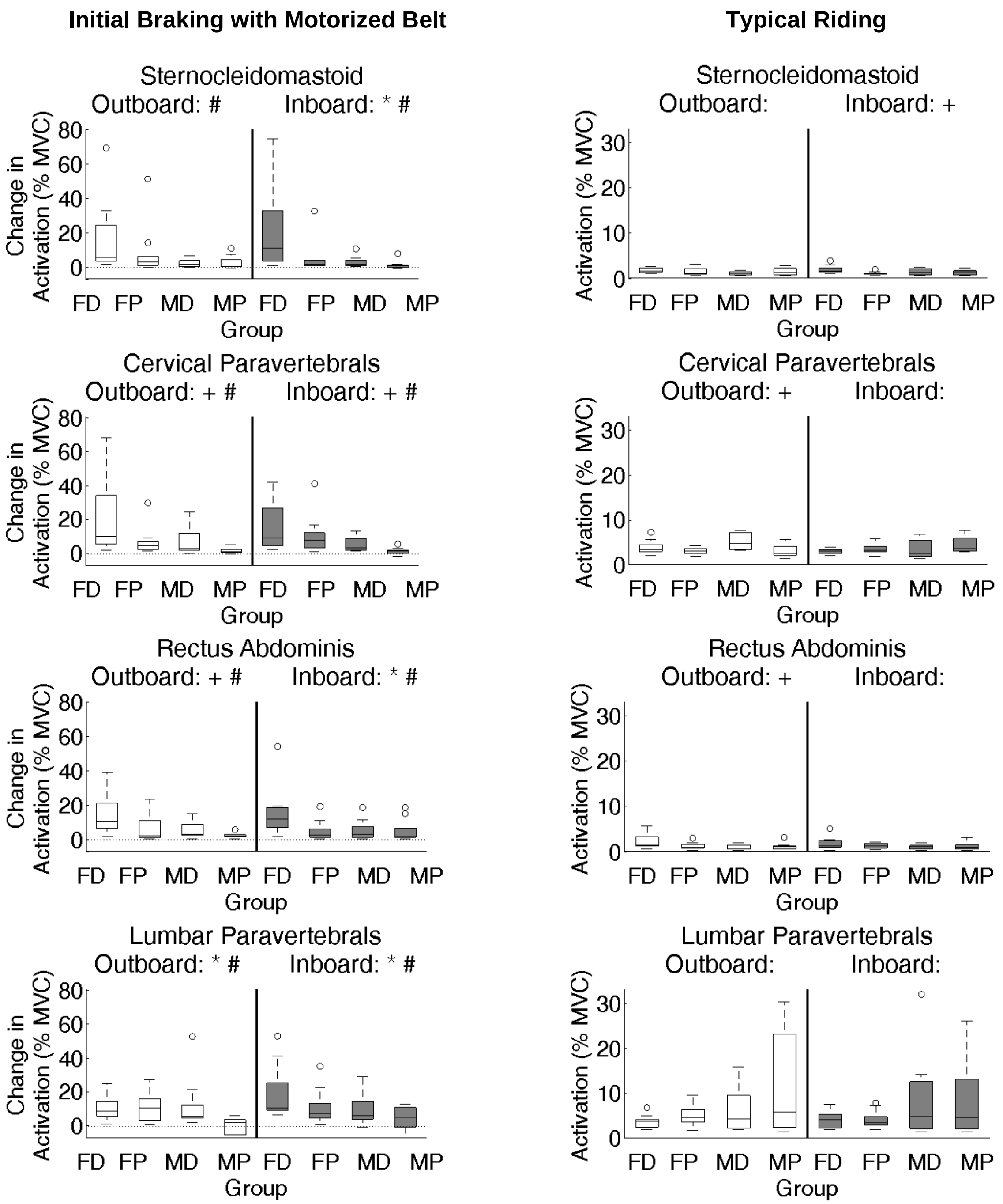

Figure 6. Median change in neck and trunk muscle activity with motorized belts for initial braking compared to typical riding (left) and median muscle activity during typical riding (right). The interquartile ranges are indicated with boxes (white for outboard muscles, grey for inboard muscles) and outliers by circles.
FD: female drivers; FP: female passengers; MD: male drivers; MP: male passengers.
**=significant main effect of gender, +=significant main effect of role, #=significant main effect of belt type.*

Table 2.
Median (25^{th}, 75^{th} percentile) Muscle Activity (%MVC) in Typical Riding.
* significant main effect of gender, + significant main effect of role.

Group	SCM	CPVM	RA	LPVM	BIC	TRIC	ADELT	PDELT
Outboard muscles								
		+	+		+	+	+	+
Female Drivers	1.6 (1.3, 2.3)	3.4 (3.0, 4.5)	1.4 (1.2, 3.3)	3.8 (2.5, 4.2)	1.0 (0.7, 1.4)	1.2 (1.0, 1.9)	1.7 (1.4, 4.3)	1.5 (1.0, 1.9)
Female Passengers	1.1 (0.9, 2.1)	3.1 (2.7, 3.6)	1.0 (0.8, 1.7)	4.7 (3.6, 6.4)	0.3 (0.2, 0.5)	0.4 (0.3, 0.5)	0.2 (0.1, 0.2)	0.7 (0.5, 1.0)
Male Drivers	1.1 (0.7, 1.5)	4.8 (3.4, 7.1)	1.4 (0.6, 1.5)	4.4 (2.3, 9.5)	0.7 (0.4, 1.3)	2.7 (2.0, 3.9)	4.0 (0.9, 5.1)	1.7 (0.9, 2.2)
Male Passengers	1.2 (0.8, 2.2)	2.6 (2.2, 4.2)	1.0 (0.5, 1.3)	5.8 (2.4, 23.1)	0.1 (0.1, 0.3)	0.3 (0.2, 0.5)	0.2 (0.1, 0.3)	0.5 (0.2, 0.8)
Inboard muscles								
	+				+	+	+	+
Female Drivers	1.6 (1.4, 2.3)	3.1 (2.6, 3.5)	1.3 (1.0, 2.5)	4.2 (2.3, 5.4)	1.4 (1.1, 1.9)	2.6 (1.2, 3.6)	3.3 (1.4, 6.5)	2.0 (1.1, 3.1)
Female Passengers	1.0 (0.9, 1.2)	3.2 (3.0, 4.2)	1.2 (0.8, 1.8)	3.4 (2.9, 4.7)	0.4 (0.3, 0.5)	0.5 (0.3, 0.7)	0.4 (0.3, 0.5)	0.8 (0.6, 1.1)
Male Drivers	1.3 (0.8, 2.2)	2.6 (2.0, 5.5)	1.1 (0.6, 1.4)	4.8 (2.1, 12.5)	0.5 (0.4, 0.9)	3.0 (2.2, 5.8)	2.7 (1.9, 5.6)	1.0 (0.9, 2.7)
Male Passengers	1.4 (0.7, 1.7)	3.6 (3.1, 5.9)	1.0 (0.5, 1.6)	4.6 (2.0, 13.1)	0.3 (0.2, 0.5)	0.5 (0.2, 0.8)	0.3 (0.2, 0.4)	0.5 (0.4, 1.7)

Table 3.
Median (25^{th}, 75^{th} percentile) Change of Muscle Activity (%MVC) with Motorized Belts.
* significant main effect of gender, + significant main effect of role, # significant main effect of belt type.

Group	SCM	CPVM	RA	LPVM	BIC	TRIC	ADELT	PDELT
Outboard muscles								
	#	+ #	+ #	* #	* #	* #	* #	* #
Female Drivers	5.5 (3.6, 24.3)	10.4 (5.5, 34.1)	10.7 (6.4, 21.2)	8.9 (5.9, 14.7)	8.7 (5.6, 5.0)	5.6 (5.0, 20.5)	12.0 (5.9, 18.2)	9.6 (8.1, 12.0)
Female Passengers	2.9 (1.0, 6.2)	4.9 (2.7, 7.0)	2.2 (1.2, 11.0)	10.6 (3.4, 16.0)	1.3 (0.6, 2.2)	1.0 (0.6, 2.5)	0.4 (0.3, 1.8)	1.4 (1.1, 10.9)
Male Drivers	1.6 (0.2, 3.7)	3.0 (2.1, 11.9)	3.1 (2.7, 8.7)	5.6 (4.6, 12.3)	1.6 (1.0, 3.2)	5.0 (1.5, 14.3)	1.0 (-1.1, 3.7)	4.6 (1.5, 5.2)
Male Passengers	0.4 (0.2, 4.1)	1.5 (0.9, 2.8)	2.2 (1.5, 2.8)	2.1 (-5.1, 3.8)	0.2 (0.1, 0.3)	0.2 (0.2, 0.4)	0.2 (0.2, 0.4)	0.7 (0.4, 1.1)
Inboard muscles								
	* #	+ #	* #	* #	* #	* #	* + #	* #
Female Drivers	11.1 (3.6, 32.6)	9.3 (4.7, 26.8)	11.7 (6.9, 18.6)	10.5 (9.4, 25.2)	8.9 (3.7, 16.7)	12.5 (5.9, 20.8)	10.6 (3.2, 14.8)	9.7 (7.9, 15.2)
Female Passengers	1.6 (0.8, 3.9)	7.9 (3.5, 12.2)	2.7 (1.4, 6.1)	7.4 (4.9, 13.4)	1.2 (0.6, 2.0)	1.8 (1.0, 3.2)	0.8 (0.3, 1.5)	2.3 (0.9, 5.5)
Male Drivers	1.6 (0.8, 4.0)	3.6 (2.3, 8.9)	2.8 (1.4, 7.6)	6.4 (4.1, 14.7)	1.3 (0.7, 2.7)	0.7 (0.1, 7.9)	1.6 (-0.4, 6.3)	3.0 (0.9, 6.4)
Male Passengers	0.6 (0.0, 1.2)	1.7 (0.4, 2.2)	1.8 (1.4, 6.5)	5.2 (-0.5,10.9)	0.2 (0.1, 0.4)	0.2 (0.2, 0.6)	0.1 (0.1, 0.3)	0.5 (0.3, 0.7)

LIMITATIONS

There are limitations inherent in the experimental procedure, such as the use of a specific test vehicle or the level of autonomous deceleration applied. With respect to the current investigation, all braking events analyzed were autonomous events with volunteers who were unaware of the impending deceleration. As prior work has shown that habituation across trials alters occupant response compared to that of unaware occupants (Blouin et al. 2003a, Siegmund et al. 2003b), the repeated nature of the testing introduces some limitations. However, randomization of the trial order helped mitigate potential habituation effects. As the passenger seat was fixed, the typical passenger riding posture does not take into account potential gender differences in seat adjustment. However, it is plausible that passengers adjust the seats to a lesser extent than drivers during real driving. On the other hand, the driver data capture differences in seat adjustment and are more representative of real life driving postures. Another limitation is the sample size of 18 volunteers.

The change in occupant metrics due to the motorized seat belt was evaluated approximately 250 ms after the start of pretension. At this time, the deceleration had started to build up and was a few percent of the maximum value. It was assumed that the deceleration during initial braking would not influence the occupant response. To ensure that the influence of factors other than the belt pretension was negligible, braking interventions with a standard belt were analyzed at initial braking and compared to the typical riding metrics. The small changes (often less than 1%MVC) in muscular activation from typical riding values seen with standard belts are consistent with the studies by Ejima et al. (2007, 2008, and 2009), which did not find major muscle activity between 0 and 100 ms after deceleration initiation for the muscles tested. They are also in line with trends seen in frontal perturbations, in which normalized integrated EMG values from the sternocleidomastoid and cervical paravertebral muscles were less than 5% and 15%MVC, respectively, for the first 25 ms after deceleration initiation (Blouin et al. 2003b). The reported median changes were within the 25th -75th percentiles of values for the typical riding (i.e. not distinguishable from typical riding) for comparable metrics. Hence, the change in muscle activity compared to typical riding can be considered an effect of motorized belt pretension, and not of deceleration onset.

DISCUSSION

During typical riding, posture and muscular activity was similar across gender. Differences were seen between drivers and passengers. Drivers displayed significantly smaller (by over 20 mm) head-to-sternum X values than passengers, meaning that drivers adopted a more protracted head posture than passengers. The typical riding postures found here are in line with other studies. For instance, Carlsson and Davidsson (2011) reported an average head-to-sternum X distance of approximately 82 mm in a similar study, which is slightly below our median values of 86 - 118 mm depending on group. Likewise, a significant effect of role was seen for muscles in the arm, which is expected due to driver interactions with the steering wheel. However, though significant differences in activation across role were present, median activation levels during typical riding were low: all median values were below 6% MVC and many were below 2% MVC. These values are in line with the maximum pre-impact activation levels of 1-5% MVC reported in the rear impact tests of Szabo and Welcher (1996) for the muscles investigated (sternocleidomastoid, suboccipital cervical extensors, superior trapezius, and paralumbar muscles). They are also in line with reported median muscle activation levels prior to sled perturbation of between 0.6 and 3.7% MVC (Ólafsdóttir et al. 2014), for several cervical muscles (sternocleidomastoid, trapezius, levator scapulae, splenius capitis, semispinalis capitis, semispinalis cervicis, and multifidus).

Motorized belts significantly altered all postural metrics. Though the median changes seen with motorized belts were small for metrics associated with the head, they were slightly larger for metrics associated with the sternum. Indeed, the sternum marker moved typically more than 10 mm posteriorly with the motorized belts, likely because of the direct interaction between the belt and the sternum. These differences were present throughout the braking event: occupants with motorized belts displayed significantly less forward displacement than occupants with standard belts (Ólafsdóttir et al. 2013, Östh et al. 2013). Similarly, Schöeneburg et al. (2011) reported that reversible belt tension reduced the median peak forward chest and neck displacements of volunteers with 42% and 34%, respectively, in braking tests with reversible belt tension compared to tests without.

Likewise, motorized belts significantly increased muscle activity. The changes were interesting as they occurred early in time relative to the total braking event, as steady-state deceleration levels were not reached until 0.8 seconds post-trigger (Ólafsdóttir et al. 2013, Östh et al. 2013). Since motorized belt pretension fires before deceleration starts, it is hypothesized that these muscular reactions occur as part of a startle or reflex response to the belt (Ólafsdóttir et al. 2013, Östh et al. 2013). Such startle contractions would result in higher muscle activity at deceleration onset with motorized belts compared to standard belts, which is an important consideration for human body modeling. To the best of the authors' knowledge, there are no comparable studies investigating muscular effects of motorized belts.

Furthermore, the effect of motorized belts appeared to be gender-specific. The gender difference in the current study was largely driven by female drivers, whose changes in muscle activity ranged between 5 and 13% MVC. As passengers, their changes were less than 5% MVC for all muscles except the lumbar paravertebrals. Also, male changes in muscle activity were typically less than 5% MVC, both as drivers and passengers. However, gender differences in reflex time and activation onset may be contributing to the differences observed in this study. Females have faster stretch reflex times than males for neck flexor (sternocleidomastoid) and extensor (semispinalis capitis, splenius capitis) muscles (Foust et al. 1973). Furthermore, Siegmund et al. (2003a) found significantly different muscle onset times between males and females in a series of frontal sled tests, with female activation occurring 5 and 3 ms before male activation for the sternocleidomastoid and cervical paraspinal muscles, respectively. Taken together, these effects could contribute to the gender differences in activation observed in the current study.

Though the presence of startle was not rigorously investigated in this study, the results presented here indicate that the initial muscular activity provoked by motorized belt pretension is different between males and females. These differences are consistent with differences found later in the braking event, where average activation levels during steady-state braking were higher for females than males for drivers (Östh et al. 2013) and passengers (Ólafsdóttir et al. 2013). Hence, we recommend further investigation of female drivers, exploring the potential startle effect that motorized belt pretension seems to induce for this group of occupants.

CONCLUSIONS

Motorized belts significantly changed the occupant posture, especially with respect to the chest, and significantly increased muscle activity. Gender did not seem to influence the typical riding postural metrics or muscle activation levels. In contrast, drivers and passengers had significantly different metrics for posture and muscle activity in typical riding. The effect of motorized belt pretension was gender-specific. When belt pretension was applied, though changes in postural metrics were similar for males and females, significant differences in activation were observed across gender. This gender difference at initial braking was driven by high changes in activation for female drivers. Therefore, further studies with a focus on female drivers are needed to explore the startle effect that motorized belt pretension seems to induce for this group of occupants.

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