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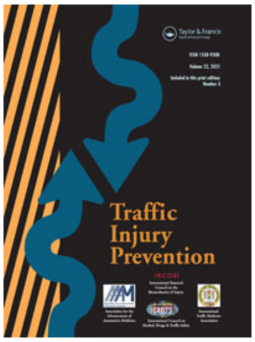

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Female kinematics and muscle responses in lane change and lane change with braking maneuvers

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ABSTRACT

Objective: The primary aim of this article is to extensively study female occupant kinematics and muscle activations in vehicle maneuvers potentially occurring in precrash situations and with different seat belt configurations. The secondary aim is to provide validation data for active human body models (AHBMs) of female occupants in representative precrash loading situations.
Methods: Front seat female passengers wearing a 3-point seat belt, with either standard or pre-pretensioning functionality, were subjected to multiple autonomously carried-out lane change and lane change with braking maneuvers while traveling at 73 km/h. This article quantifies the head center of gravity and T1 vertebra body (T1) linear and rotational displacements. This article also includes surface electromyography (EMG) data collected from 38 muscles in the neck, torso, and upper and lower extremities, all normalized by maximum voluntary contraction (MVC). The raw EMG data were filtered, rectified, and smoothed. Separate Wilcoxon signed-rank tests were performed on EMG onset and amplitude as well as peak displacements of head and T1 considering 2 paired samples with the belt configuration as an independent variable.
Results: Significantly smaller lateral and forward displacements for head and T1 were found with the pre-pretensioner belt versus the standard belt ($P < .05$). Averaged muscle activity, mainly in the neck, lumbar extensor, and abdominal muscles, increased up to 16% MVC immediately after the vehicle accelerated in the lateral direction. Muscles in the right and left sides of the body displayed differences in activation time and amplitude relative to the vehicle's lateral motion. For specific muscles, lane changes with the pre-pretensioner belt resulted in earlier muscle activation onsets and significantly smaller activation amplitudes compared to the standard belt ($P < .05$).
Conclusions: The presented results from female passengers complement the previously published results from male passengers subjected to the same loading scenarios. The data provided in this article can be used for validation of AHBMs of female occupants in both sagittal and lateral loading scenarios potentially occurring prior to a crash. Additionally, our results show that a pre-pretensioner belt decreases muscle activation onset and amplitude as well as forward and lateral displacements of head and T1 compared to a standard belt, confirming previously published results.

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KEYWORDS
Body kinematics; EMG; human body model; lane change; pretensioner belt; volunteer

Introduction

To ensure that the occupant protection systems in vehicles work effectively in crash situations, an understanding of precrash situations is needed. Mathematical tools developed to simulate vehicle occupants' responses in precrash loadings are commonly referred to as active human body models (AHBMs). Some AHBMs can be used to predict occupant responses in the in-crash phases and are therefore useful for the development of advanced and integrated safety technologies. However, AHBMs need to be validated with biomechanical data in different and representative potential precrash loading situations.

Today, AHBMs are mostly developed to represent male occupants (Iwamoto et al. 2012; Östh et al. 2015; Subit et al. 2016), though field data have shown that females are at higher risk of traffic injuries than males. For instance, it has been confirmed that in case of whiplash-associated disorders, the risk for female occupants is approximately double or even higher than that for male occupants in comparable crash circumstances (Morris and Thomas 1996; Temming and Zobel 1998; Krafft et al. 2003; Jakobsson et al. 2004; Carstensen et al. 2012; A. Carlsson et al. 2014). In addition, Bose et al. (2011) have reported that the risk of more severe injuries is higher for females than for males in analogous crash conditions.

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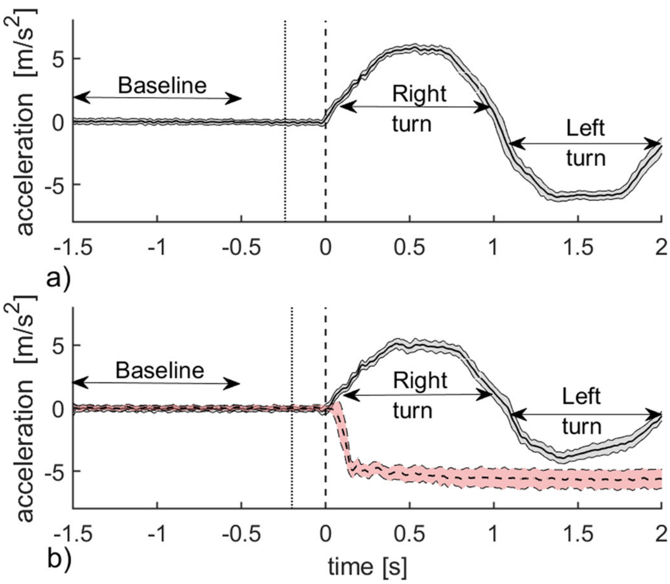

Figure 1. (a) Lateral vehicle acceleration (solid gray) in LSB and LPT and (b) lateral and longitudinal vehicle acceleration (solid gray and dashed pink, respectively) in LBSB and LBPT. Vertical dashed lines present time 0 and vertical dotted lines present the onset time of the pretensioned belt. $n = 17$, 18, 17, and 17 for LSB, LPT, LBSB, and LBPT, respectively.

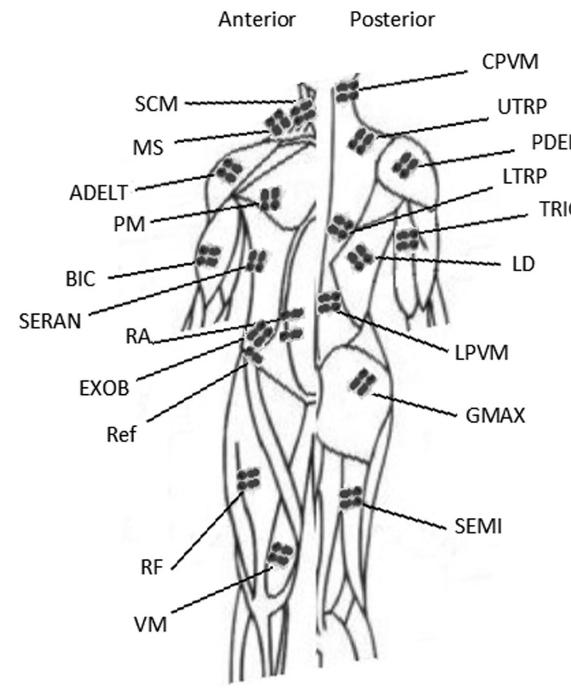

Figure 2. Electrode placement on the anterior and posterior sides of the body shown to the left and right, respectively. Muscle name abbreviations according to Table A.II.

To develop and validate female AHBMs, biomechanical data for females in representative precrash loading situations are essential. Particularly, physiological differences between males and females might affect their muscle responses and body kinematics in response to different loading scenarios. For instance, Seacrist et al. (2012) reported that cervical spine flexion differed significantly between males and females subjected to a maximum 1 g inertial loading in the posterior–anterior direction when the neck musculature was relaxed.

Behr et al. (2010) gathered data from 2 females subjected to braking conditions in the driver seat and proposed a simplified configuration including pedal loading as well as activation values for 4 groups of muscles in the lower extremities. Carlsson and Davidsson (2011) collected body kinematics data from 8 females subjected to hard braking as drivers and passengers and they found a larger forward motion for females than for males of the same sitting height. Östh et al. (2013) and Ólafsdóttir et al. (2013) provided validation data for AHBMs from 9 females subjected to braking conditions in the driver and passenger seats. In general, they found higher average normalized muscle activity and faster response for females than males, suggesting that females might be more prone to startle responses. Huber et al. (2015) presented body kinematics data from 6 female passengers subjected to braking and lane change maneuvers. Reed et al. (2018) conducted a quantitative study on 44 female passengers subjected to braking and lane change events. They demonstrated no significant difference in head excursion between females and males after accounting for body size.

These volunteer studies have provided some understanding of the occupant kinematics and activity of a small number of muscles when the volunteers were subjected to precrash situations, though none have provided a more complete set of body kinematics and muscle activity data in the form of corridors as well as the boundary conditions particularly for use in developing and validating the current models for female occupants. Recently, Ghaffari et al. (2018, 2019) provided passengers' head center of gravity and T1 body displacement data, muscle activation data, and boundary conditions such as seat belt forces and position as well as vehicle dynamics for male volunteers traveling in a regular car and exposed to lane changes and lane changes with braking maneuvers. The present article provides data from female volunteers in the passenger seat collected in the same test series as presented in the same 2 studies and with the overall objective to provide data that enable the development and validation of AHBMs.

Methods

The use of human volunteers was approved by the Ethics Review Board at the University of Göteborg (application 602-15). The test procedure, instrumentation, and data acquisition systems were as explained in Ghaffari et al. (2018, 2019). Information regarding maximum voluntary contractions (MVC) measurements, test vehicle, seat position, and pre-pretensioner belt as well as camera information are provided in Appendixes M and N (see online supplement). Figure 1 shows the lateral and longitudinal vehicle accelerations.

Sixty-nine tests including 9 female volunteers (Table A.I, online supplement) were included in this study. The recruitment of volunteers, inclusion criteria, and test cases are explained in Appendix O (see online supplement). Further information on instrumentation, data acquisition, and data analysis is provided in Appendix P (see online supplement). Placement of electromyography (EMG) electrodes and abbreviations used for naming the muscles are as described in Figure 2 and Table A.II (see online supplement).

Data analysis

Data analysis was performed using MATLAB v2015a. For each loading condition, head center of gravity (head), T1 vertebra body (T1) kinematics, and EMG response corridors were established using mean and mean ± 1 SD for all available test data excluding outliers. The muscle activation amplitude and onset were averaged across repetitions per volunteer in each loading scenario. The data were considered as 2 paired samples; that is, the belt configuration was set as a variable with 2 levels: standard and pre-pretensioning. The sample size was $n = 9$ volunteers except for a few cases of outliers or undetectable onset times that were removed from the statistical analysis because the Wilcoxon signed-rank test requires balanced data sets. Statistical analyses were performed on EMG activation level and onset time. Nonparametric Wilcoxon signed ranks tests were also used for comparing peak kinematics (head and T1 forward and lateral displacements) between LSB (lane change with standard belt) and LPT (lane change with pre-pretensioner belt), as well as separately between LBSB (lane change with braking and with standard belt) and LBPT (lane change with braking and with pre-pretensioner belt) groups. All statistical analyses were done using IBM SPSS Statistics v22.

Film analysis and kinematics postprocessing

A 3D film analysis of the data from the front, side, and rear cameras was performed using TEMA Automotive (Image Systems, Linköping, Sweden). Linear displacement of head and T1 were estimated using same method as explained in equations (3) and (4) in Ghaffari et al. (2018), respectively, except for T1, which was estimated using the average coordinate of the markers attached to the T1 level and the sternum and the term with a rotation matrix was removed from the calculation. It was to eliminate the potential error because the upper torso is not a rigid body. In addition, instead of using Euler angles, which were difficult to visualize, rotation angles of the head and upper torso around 3 axes were estimated using projection of angles on each plane made of 2 axes.

Projected rotation angles

For calculation of head and upper torso rotations around the x-, y-, and z-axes the projected angles on y–z, x–z, and x–y planes were used, respectively. For head rotation around the x-axis (Eq. (1)), the y- and z-coordinates of the 2 markers on right and left sides of the head were used. Calculation of head rotation around the *y-axis* (Eq. (2)) was done using x- and z-coordinates of the 2 markers on the front and rear sides of the head, and head rotation around the z-axis (Eq. (3)) was estimated using x- and y-coordinates of these front and rear markers. For upper torso rotation around the x-axis (Eq. (4)) y- and z-coordinates of the 2 markers on the T1 level and left acromion were used. Calculation of upper torso rotation around the y-axis (Eq. (5)) was done using x- and z-coordinates of the 2 markers on the sternum and T1 level, and upper torso rotation around the z-axis (Eq. (6)) was estimated using x- and y-coordinates of these markers.

$$\tan\psi = \frac{z\ \text{right} - z\ \text{left}}{y\ \text{right} - y\ \text{left}} \tag{1}$$

$$\tan\Theta = -\frac{z\ \text{front} - z\ \text{rear}}{x\ \text{front} - x\ \text{rear}} \tag{2}$$

$$\tan\varphi = \frac{y\ \text{front} - y\ \text{rear}}{x\ \text{front} - x\ \text{rear}} \tag{3}$$

$$\tan\psi = \frac{z\ \text{T1} - z\ \text{left acr}}{y\ \text{T1} - y\ \text{left acr}} \tag{4}$$

$$\tan\Theta = -\frac{z\ \text{sternum} - z\ \text{T1}}{x\ \text{sternum} - x\ \text{T1}} \tag{5}$$

$$\tan\varphi = \frac{y\ \text{sternum} - y\ \text{T1}}{x\ \text{sternum} - x\ \text{T1}} \tag{6}$$

Results

For each loading scenario, vehicle dynamics including lateral and longitudinal accelerations (Figure 1) and angular displacements around the x-, y-, and z-axes (Figure B.1, see online supplement); shoulder and lap belt interaction forces (Figure C.1, see online supplement); and volunteer kinematics corridors (Figures D.1–D.8, see online supplement), were established using mean and mean ± 1 standard deviation with $n = 17$, 18, 17, and 17 for LSB, LPT, LBSB, and LBPT, respectively. Complete sets of EMG data corridors from 38 muscles and EMG data deemed to be outliers for 4 types of loading scenarios (LSB, LPT, LBSB and LBPT) are presented in Appendixes G–J (see online supplement).

Kinematics

For the LSB scenario, corridors of head kinematics are illustrated in Figure D.1. As shown in Figure D.1, the head appears to have greatest displacement on the y-axis compared to the x- and z-axes. Corridors of T1 displacement are illustrated in Figure D.2. Again, the greatest T1 displacement is found on the y-axis and its direction is consistent with head lateral displacement. Pelvis forward displacement was found to be negligible according to the pressure distribution on the passenger seat.

Comparing corridors for head, T1, and upper torso kinematics in LSB (Figures D.1 and D.2, respectively) with the same corridors in LPT (Figures D.3 and D.4) indicates that in the first second of the maneuvers, volunteers have significantly less head and T1 lateral displacement in LPT (maximum 95 mm for head and 50 mm for T1) than in LSB (maximum 140 mm for head and 97 mm for T1; $P = .008$). The same comparison between corridors in LBSB (Figures D.5 and D.6) and corridors in LBPT (Figures D.7 and D.8)

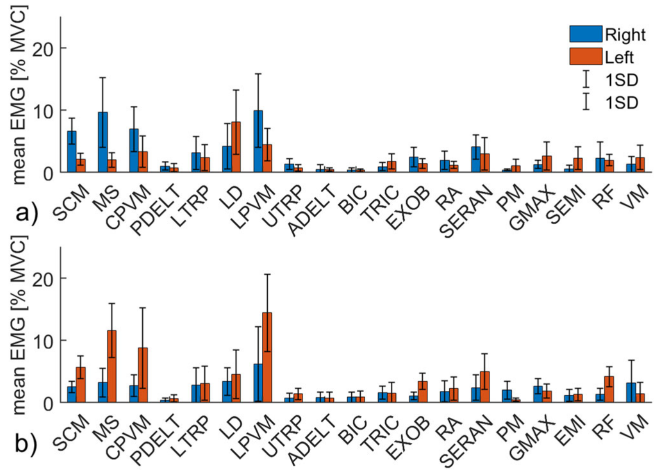

Figure 3. Grand average and standard deviation of EMG for LSB maneuver, during the (a) right turn phase and (b) left turn phase for all muscles on the right (blue bars) and the left (red bars) sides.

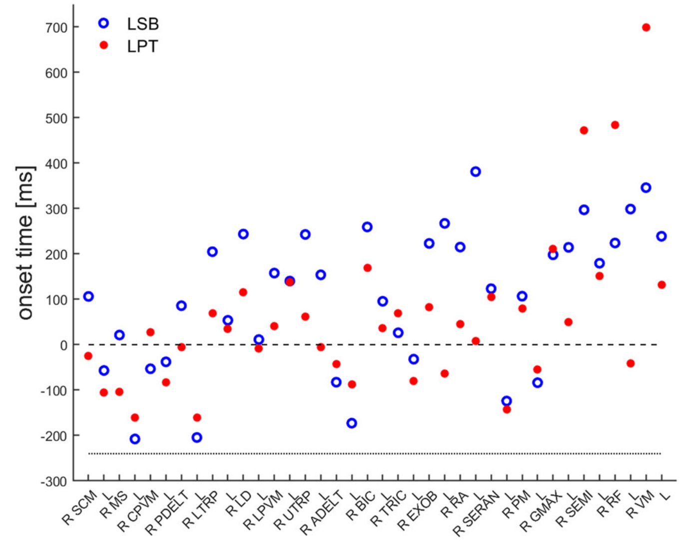

Figure 4. Average EMG onset times showing 2 channels (i.e., right [R] and left [L]) for each muscle during the right turn phase for LSB and LPT. Horizontal dashed line presents time 0 and the dotted line presents the onset time of the pre-pretensioner belt.

results in significantly less head and T1 lateral displacement in LBPT (maximum 95 mm for head and 59 mm for T1) than in LBSB (maximum 150 mm for head and 117 mm for T1; $P = .012$).

Head and T1 forward displacement were significantly less in LBPT (maximum 74 mm for head and 30 mm for T1) than in LBSB (maximum 112 mm for head and 72 mm for T1) compared to corresponding corridors ($P = 0.012$). Comparing head and T1 displacements normalized by seated height reveals the same statistically significant results. Peak of head and T1 forward and lateral displacement, raw values and values normalized by seated height, are presented in Tables F.I and F.II (see online supplement). Head and T1 displacements on the z-axis were much less than the displacement on the other 2 axes. Comparing the mean of corridors shows that displacement on the z-axis was even less in maneuvers with a pre-pretensioner than with a standard belt, although there were relatively large standard deviations. Head and upper torso rotations did not indicate noticeable differences between those loading conditions. Mean kinematics of the head and T1 during 4 loading scenarios are presented in Figures E.1. and E.2 (see online supplement).

Grand average of EMG values

During the baseline phase, all muscles were activated less than 2% MVC on average, except for the lumbar extensor muscle (LPVM approximately 3.3% MVC). The activation levels in the baseline phase are similar for all loading scenarios, indicating that the volunteers were in similar muscle states before the different events (Appendix L, see online supplement). Figure 3 illustrates the grand average of EMG values for the LSB maneuvers. As shown in this figure, for muscles that were noticeably active on both sides, the right side is more active during the right turn phase and the left side is more active during the left turn phase. This pattern was also noticed in 3 other maneuvers (Figures K.1–K.3, see online supplement).

The results show that the LPVMs have the highest average activation levels during the right turn phase and left turn phase (9%–16% MVC) for all maneuvers and the largest standard deviation (Figure 3 and Appendix L). The neck muscles (SCM, MS, CPVM), the LPVMs, and the abdominal muscles (EXOB) have noticeable activation of 3%–16% MVC on both sides of the body (Figure 3 and Appendix L). Muscles in the upper extremities (PDELT, ADELT, BIC, and TRIC) consistently have very low muscle activity (<2% MVC) except for left TRIC in LBSB (<4% MVC).

The muscles with statistically significant differences between LSB and LPT as well as between LBSB and LBPT grand average EMG values in the right turn phase and left turn phase ($P < .05$) are shown in Tables L.I and L.II (see online supplement). As seen in these tables, the LSB and LBSB demonstrate greater average EMG values than those of the LPT and LBPT, respectively, for all muscles marked with significant differences.

EMG onset time

Average EMG onset time for the right turn phase differs between LSB and LPT (Figure 4). The difference was statistically significant for the right LD, right UTRP, left EXOB, and right RA ($P < .05$). All muscles mentioned above have later onset times in the LSB maneuver than in the LPT maneuver as seen in Figure 4. Moreover, some EMG onsets indicate that muscle activation starts before the lateral acceleration onset, just after activation of the pre-pretensioner belt in LPT maneuvers, or in some cases even in LSB maneuvers (Figure 4). The same comparison of EMG onset time was done between LBSB and LBPT (Figure K.4, see online supplement), whereas the difference was statistically significant only for the left LPVM ($P < .05$). Similarly, EMG onset time was studied for the left turn phase of the maneuvers. However, muscle activation had already started in the right turn phase, and after the first half of the turning maneuver, the volunteers were in different postures in the beginning of the left turn phase. Therefore, the results for onset time in the left turn phase were not meaningful.

Discussion

This study provides quantified head, T1, and upper torso kinematics as well as normalized EMG values with respect to isometric MVCs from 38 muscles of female passengers in response to lateral and longitudinal precrash loading conditions. The volunteer response data together with the vehicle kinematics and volunteer–vehicle interaction forces provided in this study can be used as validation data for female AHBMs.

The results obtained from the comparison between 2 belt configurations confirm the findings reported in Ólafsdóttir et al. (2013), Östh et al. (2013), Holt et al. (2018), and Ghaffari et al. (2018, 2019). In particular, using a pre-pretensioner belt led to a significant decrease ($P < .05$) in the activation level as well as earlier onset times for specific muscles compared to using a standard belt. There was also significantly less head and T1 lateral and forward displacement using a pre-pretensioner than using a standard belt ($P < .05$; Tables F.I and F.II). Activation of the pre-pretensioner causes the belt start to pull in at around 200 ms before time 0 (beginning of lateral acceleration) until reaching the target tension (170 N) at approximately time 0 (Figure C.1). Although it is unknown whether the generated load by the pre-pretensioner is mainly on clavicle or is distributed over the larger part of the upper body, it pulls back the volunteers' upper body. Onset of muscle activity in the studied muscles while using the pre-pretensioner was at least 80 ms after activation of the pre-pretensioner. Whether this was due to a startle response is difficult to determine because of the possible superimposition of muscle activities. A possible explanation for the observed decrease in muscle activity level while using the pre-pretensioner is that volunteers might need less muscle activity to restrict their motions than when using a standard belt. On the other hand, there were some cases of muscle activation before time 0 when using a standard belt, which implies the possibility that volunteers anticipated the maneuvers and muscle activation occurred prior to the event. For instance, hearing the sound produced by the clutch just prior to the event could be a forewarning. Given that, this can affect the muscle responses as well as kinematics and cause potential errors. Using additionally sound-isolated robot systems can help to rule out such errors in future studies.

There were some experimental limitations associated with the test setup in this study, such as the lack of surrounding regular traffic, potential awareness of the upcoming maneuvers, and possible habituation effects on occupant responses to the vehicle maneuvers. However, because the maneuvers were conducted and repeated in a randomized order, habituation effects are less likely. In addition, to avoid possible habituation effects, volunteers were not asked to assume a neutral posture before the beginning of each maneuver. Therefore, about 30% of the test cases had to be excluded where volunteers had different initial postures rather than neutral. Investigation of the results showed that the kinematics corridors were not affected much by the excluded cases. The exclusion primarily reduced SD in the baseline phase of the volunteers' kinematics, whereas the corridors' overall trend in response to the loading scenario were the same. This effect was not investigated for the EMG corridors. There were also some limitations associated with data analysis in this study. One of these limitations is imposed by the filtering of the EMG signals, which is meant to remove electrocardiography and other undesirable components from the recorded signals. However, because the same signal processing methods were applied to all EMG signals, the effect of filtering was the same for all of them. Another limitation lies in the method used for estimation of body part kinematics. In calculation of linear displacement of the T1 vertebra body, the average coordinate of the markers attached to the T1 level and the sternum was used to provide a more precise estimation than using only 1 marker attached to the T1 level. These experimental and analytical limitations were extensively discussed in Ghaffari et al. (2018, 2019).

In line with kinematics of male volunteers reported in Ghaffari et al. (2018), female kinematics showed that T1 had less lateral and forward displacement than the head in all loading conditions. Preliminary comparison between responses of the female occupants presented in this article and male occupants subjected to the same loading scenarios presented in Ghaffari et al. (2018, 2019) indicated that females generally exhibited less lateral and forward displacement of the head and T1 than males. However, this preliminary comparison does not account for the differences in body size between females and males. According to one recent study, in which a data set with a larger sample size was analyzed (Reed et al. 2018), no significant difference was found in head excursion between females and males during braking events and lane change events after accounting for body size. The mean maximum lateral displacement of the head was 118 (mm) in their study of lane changes with lateral acceleration of around 0.7 g and a pulse duration of about 1.2 s. On the other hand, they found that taller volunteers had on average larger lateral head displacement during lane change events. The results presented in this study partly confirm their finding because the female occupants in this study with less head displacement had smaller stature (on average 169 cm) than male occupants (on average 183 cm) reported in Ghaffari et al. (2018). Nevertheless, the presented body kinematics were not scaled with respect to body size because currently there is no proper scaling method. Therefore, it is unknown whether the observed differences between female and male kinematics of are due to the differences in sex or body size. In addition, average muscle activation in females was generally higher in the right turn phase compared to males, whereas in the left turn phase it was lower (up to 16% MVC) compared to males (up to 24% MVC; Ghaffari et al. 2019).

As shown in the recent study by Reed et al. (2018), volunteers' characteristics such as age, stature, and BMI can influence their response to the precrash loading conditions. They found, for instance, that older passengers and those with higher BMI had smaller forward head excursions in braking events. Hence, the results presented in this article together with the previously published papers (Ghaffari et al. 2018, 2019) that cover the quantified responses of volunteers of different sex, age, and body size can be valuable for further statistical investigation of within- and between-subject variations influencing their responses in precrash situations and in response to the seat belt configuration. The focus of this article was to present and analyze female EMG and kinematic data and investigate the effect of a pre-pretensioner belt on their responses. The population heterogeneity, between-subject variation, and statistical regression analyses will be the main objective of an upcoming paper.

Future studies on comparison between the simulation results of the current AHBMs designed for females and this new volunteer data set in lateral loading scenarios are essential to validate the models and to allow for additional parameter studies. Validation of the female AHBMs against the volunteer data, which are fairly representative of the population, will improve the possibility of predicting sex-specific behavior of humans in potential precrash scenarios.

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References

Behr M, Poumarat G, Serre T, Arnoux PJ, Thollon L, Brunet C. 2010. Posture and muscular behaviour in emergency braking: an experimental approach. Accid Anal Prev. 42(3):797–801. doi:10.1016/j.aap.2009.04.010

Bose D, Segui-Gomez M, Crandall JR. 2011. Vulnerability of female drivers involved in motor vehicle crashes: an analysis of US population at risk. Am J Public Health. 101(12):2368–2373. doi:10.2105/AJPH.2011.300275

Carlsson A, Chang F, Lemmen P, Kullgren A, Schmitt K-U, Linder A, Svensson MY. 2014. Anthropometric specifications, development, and evaluation of EvaRID-a 50th percentile female rear impact finite element dummy model. Traffic Inj Prev. 15(8):855–865. doi:10.1080/15389588.2014.885647

Carlsson S, Davidsson J. 2011. Volunteer occupant kinematics during driver initiated and autonomous braking when driving in real traffic environments. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September, Krakow, Poland; p. 14–16.

Carstensen TBW, Frostholm L, Oernboel E, Kongsted A, Kasch H, Jensen TS, Fink P. 2012. Are there gender differences in coping with neck pain following acute whiplash trauma? A 12-month follow-up study . Eur J Pain. 16(1):49–60. doi:10.1016/j.ejpain.2011.06.002

Ghaffari G, Brolin K, Bråse D, Pipkorn B, Svanberg B, Jakobsson L, Johan D. 2018. Passenger kinematics in Lane change and Lane change with Braking Manoeuvres using two belt configurations: standard and reversible pre-pretensioner. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September 12–14, Athens, Greece.

Ghaffari G, Brolin K, Pipkorn B, Jakobsson L, Johan D. 2019. Passenger muscle responses in lane change and lane change with braking maneuvers using two belt configurations: standard and reversible pre-pretensioner. Traffic Inj Prev. 20(S1):43–51.

Holt C, Douglas E, Graci V, Seacrist T, Kerrigan J, Kent R, Balasubramanian S, Arbogast K. 2018. Effect of countermeasures on adult kinematics during pre-crash evasive swerving. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September 12–14, Athens, Greece.

Huber P, Kirschbichler S, Pruggler A, Steidl T. 2015. Passenger kinematics in braking, lane change and oblique driving maneuvers. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September 9–11, Lyon, France.

Iwamoto M, Nakahira Y, Kimpara H, Sugiyama T, Min K. 2012. Development of a human body finite element model with multiple muscles and their controller for estimating occupant motions and impact responses in frontal crash situations. Stapp Car Crash J. 56: 231–268.

Jakobsson L, Norin H, Svensson MY. 2004. Parameters influencing AIS 1 neck injury outcome in frontal impacts. Traffic Inj Prev. 5(2): 156–163. doi:10.1080/15389580490435989

Krafft M, Kullgren A, Lie A, Tingvall C. 2003. The risk of whiplash injury in the rear seat compared to the front seat in rear impacts. Traffic Inj Prev. 4(2):136–140. doi:10.1080/15389580309862

Morris AP, Thomas PD. 1996. Neck injuries in the UK co-operative crash injury study. In: Stapp Car Crash Conference. Warrendale (PA): Society of Automotive Engineers; p. 317–329.

Ólafsdóttir JM, Östh J, Davidsson J, Brolin K. 2013. Passenger kinematics and muscle responses in autonomous braking events with standard and reversible pre-tensioned restraints. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September 11–13, Gothenburg, Sweden.

Östh J, Brolin K, Bråse D. 2015. A human body model with active muscles for simulation of pretensioned restraints in autonomous braking interventions. Traffic Inj Prev. 16(3):304–313. doi:10.1080/15389588.2014.931949

Östh J, Ólafsdóttir JM, Davidsson J, Brolin K. 2013. Driver kinematic and muscle responses in braking events with standard and reversible pre-tensioned restraints: validation data for human models. Stapp Car Crash J. 57:1–41.

Reed MP, Ebert SM, Jones MLH, Park BKD, Hallman JJ, Sherony R. 2018. Passenger head kinematics in abrupt braking and lane change events. Traffic Inj Prev. 19(S2):70–77.

Seacrist T, Saffioti J, Balasubramanian S, Kadlowec J, Sterner R, García-España JF, Arbogast KB, Maltese MR. 2012. Passive cervical spine flexion: the effect of age and gender. Clin Biomech (Bristol, Avon). 27(4):326–333. doi:10.1016/j.clinbiomech.2011.10.012

Subit D, Möhler F, Wass J, Pipkorn B. 2016. Robustness of principal and longitudinal strains as fracture predictors in side impact. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September 14–16, Malaga, Spain.

Temming J, Zobel R. 1998. Frequency and risk of cervical spine distortion injuries in passenger car accidents: significance of human factors data. Paper presented at: International Research Council on Biomechanics of Injury (IRCOBI), September 16–18, Gothenburg, Sweden.