Société de Biomécanique Young Investigator Award 2019: Upper body behaviour of seated humans *in vivo* under controlled lateral accelerations

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A R T I C L E I N F O

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A B S T R A C T

Background: A deep understanding of human reactions and stabilization strategies is required to predict their kinematics under external dynamic loadings, such as those that occur in vehicle passengers. Low-level frontal accelerations have been thoroughly investigated; however, the human response to different lateral accelerations is not well understood. The objective of this study is to gain insight regarding the responses of seated humans to lateral perturbations from volunteer experiments in different configurations.
Methods: Five volunteers anthropometrically comparable to the 50th-percentile American male, were seated on a sled and submitted to 21 lateral pulses. Seven configurations, each repeated three times, were investigated in this study: a relaxed muscular condition with four pulses, namely, sine and plateau pulses of 0.1 and 0.3 g in a straight spinal posture; a relaxed muscular condition with a plateau pulse of 0.3 g in a sagging spinal posture; and a braced condition with both plateau pulses in a straight spinal posture. Upper body segment kinematics were assessed using inertial measurement units.
Findings: The maximum lateral bending of the head was found to differ significantly among the four acceleration pulses ($p < 0.001$). Braced muscles significantly reduced lateral bending compared to relaxed muscles ($p < 0.001$). However, no significant difference was found in lateral bending between straight and sagging spinal postures ($p = 0.23$).
Interpretation: The study shows that not only pulse amplitude but also pulse shape influences human responses to low accelerations, while spinal posture does not influence lateral head bending. These data can be used to evaluate numerical active human body models.

1. Introduction

Evasive manoeuvres, performed either manually or automatically by advanced driver assistance systems in highly automated vehicles, influence the kinematics and position of occupants within the vehicle interior (Graci et al., 2020; Holt et al., 2020). Understanding human reactions and stabilization strategies under such load cases is therefore a key factor in the continuous optimization of integrated safety for future vehicles. Tools such as active human body models (AHBMs) are useful to predict occupant kinematics while taking into account human muscular reactions during a pre-crash phase, which can influence occupants' interaction with the restraint system. Numerous research studies have been conducted utilizing postmortem human subjects (PMHS) (Park et al., 2013; Serre et al., 2007); however, these data can be used only to validate the passive properties of AHBMs, as PMHS lack neuromuscular control (Beeman et al., 2012; Higuchi et al., 2020; Lopez-Valdes et al., 2017; Parenteau, 2006; Parenteau et al., 2002). Hence, additional validation data obtained from volunteer experiments under conditions resembling vehicle evasive manoeuvres are needed. Factors such as pulse level and shape, different postures, and different muscular states need to be considered due to the wide range of conditions observed in the real world.

In recent years, human body responses have been investigated in volunteer tests during non-injurious sled loading (0.2 to 3.7 g), focusing mainly on frontal and rear accelerations (Beeman et al., 2011; Blouin et al., 2003; Dehner et al., 2007, 2008, 2013; Ejima et al., 2007, 2008;

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Hernández et al., 2005; Kemper et al., 2014; Ewing et al., 1976; Kirschbichler et al., 2011; Kumar et al., 2000, 2003; Lopez-Valdes et al., 2017; Kirschbichler et al., 2014; Sandoz et al., 2016). A less studied phenomenon is human reaction to lateral accelerations, for instance, those experienced by occupants during a lane change manoeuvre. Ejima et al. (2012) subjected three male volunteers to two plateau lateral pulses of 0.4 and 0.6 g for 600 ms; the volunteers were asked to either relax or brace their muscles. Under the latter muscular condition, lateral rotation of the torso was reduced by 5° and 10° at 0.4 and 0.6 g, respectively. Likewise, van Rooij et al. (2013a, 2013b) and Holt et al. (2018) reported a decrease in lateral movement of the head under 0.5 g sine and plateau and 0.75 g sine pulses, respectively, for the braced muscle condition compared to the relaxed condition. Common trends of increased lateral displacement and rotation under higher loads and a decrease in the braced muscle condition have been demonstrated. However, the influence of different pulse shapes on human reactions has not been thoroughly investigated. Likewise, the influence of different upper body postures has not yet been investigated under lateral pulses even if, in reality, a wide variety of postures are adopted by vehicle occupants (Morris et al., 2005). Additionally, different sitting postures lead to different initial muscle tone under static conditions (Caneiro et al., 2010) and to different kinematics during frontal accelerations (Graci et al., 2019; Mackenzie et al., 2022; Reed et al., 2018a, 2018b; Reed et al., 2021).

The present research hypothesizes that initial posture affects the initial muscle tone and hence the stabilization strategies of the human body under low to moderate lateral accelerations. Not only the initial posture but also the characteristics of the pulse, such as jerk, duration and level of acceleration, could affect human responses. Finally, in previous studies, it has been shown that muscle state influences kinematics. However, most of these studies have been conducted with restricted upper body movement due to seatbacks, some of which were equipped with side bolsters and seatbelts. Consequently, it remains an open question whether this is also the case for a lower level of external upper body motion restriction.

The aim of this research is to study the influence of posture, acceleration level and shape, and muscular state on human upper body kinematics during lateral accelerations. Therefore, data were acquired from five male volunteers seated on a rigid sled without any upper body support subjected to different lateral pulses in two body postures with two muscular states.

2. Methods

An existing sled setup was optimized to apply lateral accelerations to seated volunteers (Sandoz et al., 2019). Five male volunteers were seated on a rigid seat plate that was accelerated to their left. Each of them was submitted to a total of 30 lateral pulses, of which 21 were analysed in this paper (Appendix 1). The experimental matrix includes various postural and muscular configurations, as well as levels of acceleration. Each configuration was repeated three times. The nine additional pulses not presented in this paper were performed between the 12th and 13th pulses (Appendix 1).

2.1. Ethics

The following protocol was approved by the French Ethics Committee CPP IDF VII number 15–018. All volunteers were fully informed and signed a consent form.

2.2. Volunteer recruitment and instrumentation

The inclusion criteria were as follows: males aged 18 to 50 years whose anthropometry was comparable to that of the 50th-percentile American male (77 ± 5 kg, 175 ± 3 cm). Volunteers were excluded if they had any recent muscular soreness or pain.

Volunteers were instrumented with four inertial measurement units (IMU, Xsens MTx) on the top of the head, the first thoracic vertebra (T1), the sternum and the sacrum (S1), and the orientations and accelerations were recorded at 100 Hz. This acquisition frequency is similar to that of IMUs in similar previous experiments (Mackenzie et al., 2022; Reed et al., 2018a, 2018b; Sandoz et al., 2016; Sandoz et al., 2019). The head IMU was firmly attached with cable ties and tape on an adjustable rigid head strap, tightened on the volunteer's head. The T1, sternum and S1 IMUs were fixed to the body with double-sided tape and secured using hypoallergenic adhesive strips. The activity of the following bilateral muscle pairs was recorded at 2000 Hz using wireless electromyography (EMG) sensors (Trigno Delsys): the sternocleidomastoids, trapezius, latissimus dorsi, erector spinae, external obliques and rectus abdominis on the upper body, as well as the vastus lateralis and biceps femoris on the legs.

2.3. Sled description and instrumentation

Volunteers were seated on an instrumented four-metre-long sled and secured by a Velcro belt across the thighs (Figs. 1.a. and 1.b.). The initial tension of the belt was 120 N, measured by a handheld load cell (Kern company, capacity 50 kg, precision ±0.05 kg) (Fig. 1.b.). Pelvic brackets 90 mm high were adjusted to the left and right of the pelvis to minimize its lateral movement and prevent sliding (Fig. 1.b. and 1.c.). An on-board video camera filmed the volunteers from behind at 50 fps. An additional IMU (Xsens MTx, 100 Hz) was fixed under the seat plate to record its acceleration.

All sensors were synchronized with a common trigger signal. The sled pulses were programmed and validated using precise and reproducible closed-loop control (Sandoz et al., 2014). The dimensions and configuration of the sled are available in Appendix 2.

2.4. Lateral acceleration profiles

Four acceleration profiles were applied to the seat plate (Fig. 2). Sine Low (SL) and Sine High (SH) were single-cycle sine-shaped pulses with peaks of 0.1 and 0.3 g, respectively, and a duration of 2 s. Plateau Low (PL) and Plateau High (PH) were single-cycle plateau-shaped pulses with maximum accelerations of 0.1 and 0.3 g, respectively, and a duration of 3 and 2 s, respectively. The deceleration phases were symmetrical to the acceleration phases for the four pulses.

These pulses were chosen in accordance with the mechanical characteristics of the sled. In previous vehicle experiments, lane change acceleration pulses usually follow a sinusoidal shape with different levels of acceleration (Ghaffari et al., 2018; Huber et al., 2015; Reed et al., 2018a, 2018b). However, there is no single, universal vehicle lane change pulse. Therefore, in addition to the sinusoidal pulses, plateau pulses with greater jerk and a constant acceleration level were defined with respect to the hypothesis.

2.5. Test protocol

Prior to each sled pulse, the volunteers placed their hands on their thighs and feet on the footrest. Depending on the configuration, they were asked to assume one of the three physical states: a sagging spinal posture with relaxed muscles or a straight spinal posture with either relaxed or braced muscles (Fig. 1.b.). These were chosen expecting that the sagging spinal posture would induce low initial muscle tone, the straight spinal posture with relaxed muscles would induce additional muscular contraction to keep the upper body straight, and the straight spine posture with braced muscles would induce the highest muscular contraction of the three conditions.

A horizontal laser beam line projected on the back of each volunteer ensured the same initial straight posture preceding each pulse. A countdown warned the volunteers before each pulse, but they were not aware of the characteristics of the upcoming acceleration pulses. In the

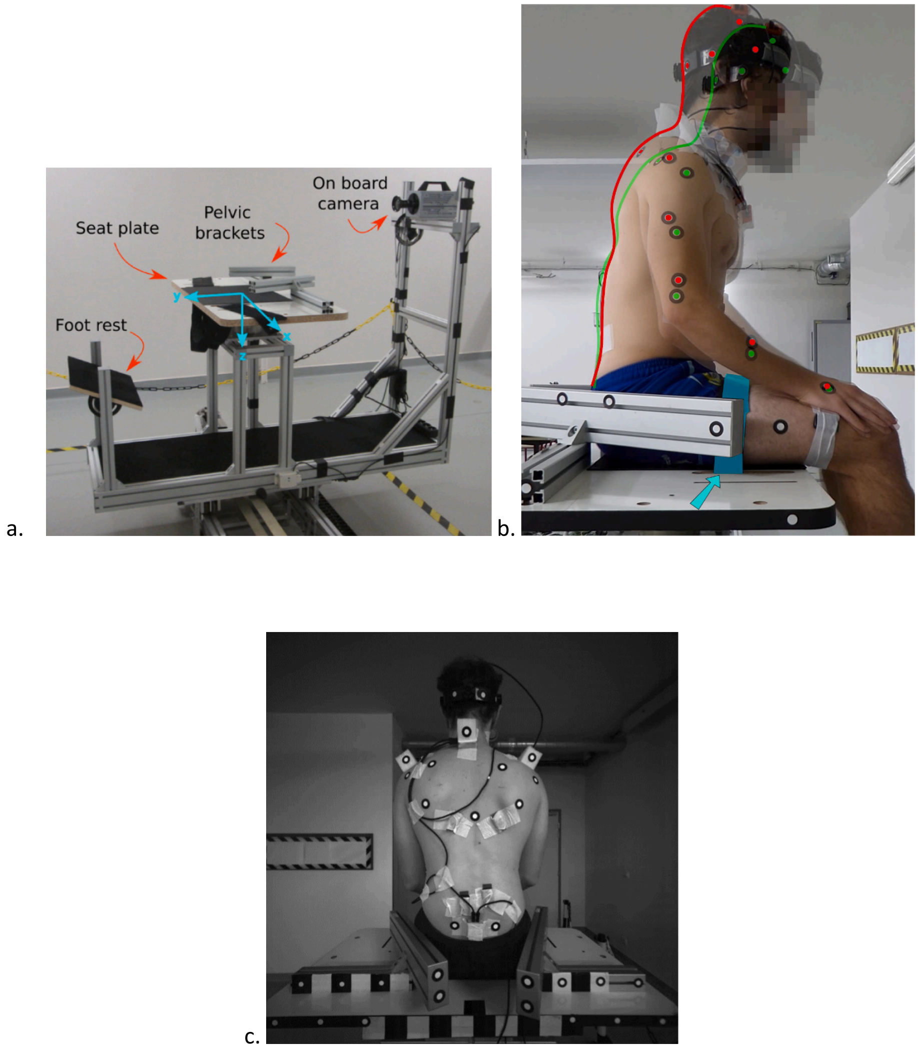


Fig. 1. Sled in its initial position. a. Global view. b. Superimposed lateral views of a straight (red) and sagging (green) spinal posture. The arrow points to the lap belt. c. Posterior view of the volunteer from the on-board camera. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

present study, seven different configurations, repeated three times each, were analysed. The first twelve pulses were randomized in terms of acceleration pulse characteristics to avoid habituation. In this configuration, the muscles were relaxed, and the spine was straight. The following configurations were used: three PH acceleration pulses with a sagging spine and relaxed muscles, three PH acceleration pulses with a straight spine and braced muscles, and three PL acceleration pulses with a straight spine and braced muscles (detailed test matrix in Appendix 1).

2.6. *Studied parameters and data processing*

The first twelve pulses were used to study the effect of acceleration level and shape. The effect of muscle state was studied by comparing pulses {2; 8; 10} and {19; 20; 21} for the PL input accelerations and pulses {5; 9; 11} and {16; 17; 18} for the PH input accelerations. Pulses {5; 9; 11} and {13; 14; 15} were compared to study the effect of spinal posture.

Acceleration and lateral bending of the upper body were studied at each inertial measurement unit sensor location. The lateral bending measurements from the IMU on top of the head, hereafter referred to as lateral head bending, are presented in detail in this paper. No additional filter has been applied to the IMU sensor output regarding lateral bending or accelerations. The orthogonal reference frames of the IMU sensors were numerically oriented in the same directions for each volunteer's initial posture: +X was directed to the left of the volunteer in the direction of sled motion, +Y was directed to the front of the volunteer, and + Z was directed toward the floor (Fig. 1.a.). The present paper focuses on lateral head bending about the Y-axis, since it is the main movement under pure lateral pulses. All data, including accelerations and lateral bending of the other body segments, are provided (Appendix 5 and supplementary data).

Lateral bending and acceleration data are provided in the form of corridors. These were defined as the area between the maximum and minimum value of all the volunteers for the specific represented measurement and condition. All corridors also include the mean and ± 1 standard deviation (SD). Boxplots (Appendix 3) present the median, 1st and 3rd quartiles, and extreme values.

2.7. *Statistics*

Maximum bending angles during the acceleration phase (noted as positive) and the deceleration phase (noted as negative) of the volunteers were considered as parameters of interest, *i.e.*, the dependent variables to analyse the differences between configurations. The independent variables were the acceleration profiles (SL, SH, PL, or PH), the muscle contraction (relaxed or braced) and the spine position (straight

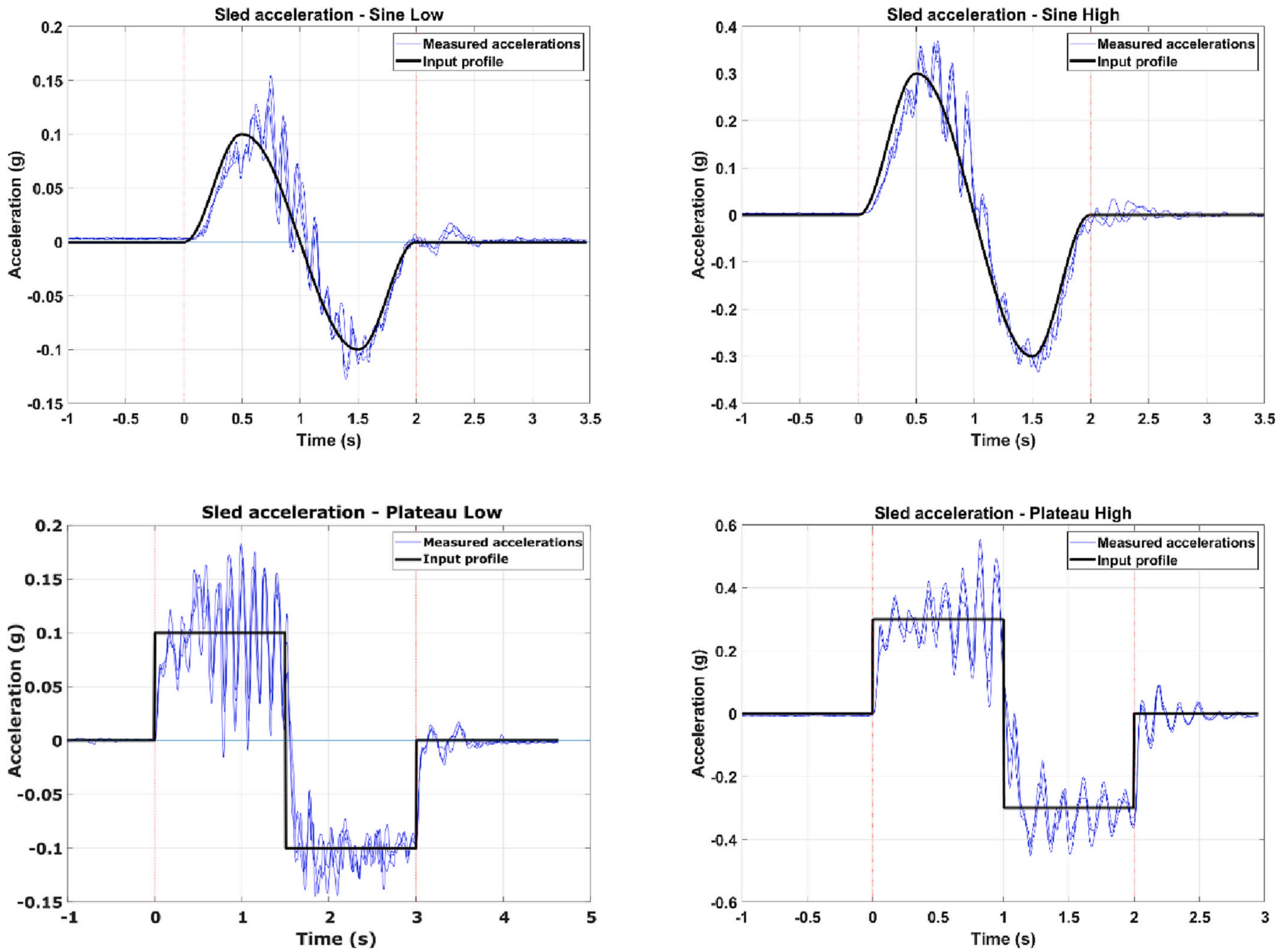


Fig. 2. Sled acceleration profiles on one volunteer for three pulses: Sine Low, Sine High, Plateau Low and Plateau High. Input profiles (black line) and measured acceleration under the seat pan (blue line). Time 0 s is the beginning of the sled motion. Positive values: acceleration phase. Negative values: deceleration phase. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

or sagging).

The significance levels of the differences between all combinations of the configurations were tested using multivariate analysis of variance (MANOVA). Then, the normality of the considered angle of lateral head bending distributions was verified using a Lilliefors test, adapted for small sample sizes. Depending on the result of the normality test, the significance levels of the differences between two configurations were calculated using a paired Student's *t*-test for normal distributions and the Wilcoxon test for nonnormal distributions. The significance threshold was set at $p < 0.05$.

3. Results

3.1. Volunteers

Thirteen males volunteered to participate in the experiment. After the inclusion and exclusion criteria were applied, five healthy males were finally accepted for the study. The average volunteer was 26 to 36 years old, 175 cm tall (SD 2.3), and 76 kg in mass (SD 3.4), with a body mass index of 24.6 kg.m^{-2} (0.8 SD) (Table 1). Volunteers felt no pain or discomfort during or within two days of the experiments. No habituation was observed on kinematics or EMG data.

Table 1
Volunteers' anthropometry.

Volunteer #	Height (cm)	Mass (kg)	Body Mass Index (kg. m^{-2})	Age (years)
1	175	78	25.5	26
2	174	73	24.1	33
3	179	78	24.3	27
4	172	71	24.0	27
5	175	80	26.1	36
Mean (SD)	175 (2.3)	76 (3.4)	24.8 (0.8)	30 (4.1)

3.2. Lateral bending measured at the top of the head

The MANOVA showed $p < 0.0001$. All p values calculated for the differences of peak lateral bending angles between various configurations, as measured at the top of the head, are summarized in Table 2.

The effect of the four acceleration pulses on the head bending of each volunteer was studied for a straight spinal posture and relaxed muscular condition. Maximum head bending during acceleration significantly increased from SL to PL, SH and PH pulses ($p < 0.001$), with mean values rising from 3° to 12°, 15° and 20°, respectively (Fig. 3). Maximum head bending during decelerations was significantly different among configurations ($p < 0.001$). Paired comparisons of maximum head bending between SL and SH profiles as well as between PL and PH profiles

Table 2

P-values of paired Student's T-test (when normal distribution) and Wilcoxon test (when nonnormal distribution) of extreme lateral bending angles measured at the top of the head. SL: Sinus Low, SH: Sinus High, PL: Plateau Low, PH: Plateau High, acc: acceleration phase, dec: deceleration phase, br: braced muscles, sag: sagging spinal posture. If not indicated, the condition is straight spine posture and relaxed muscles.

	SH acc	SH dec	PH acc	PH dec	PL acc br	PL dec br	PH acc br	PH dec br	PH acc sag	PH dec sag
SL acc	< 0.001									
SL dec		< 0.001								
PL acc			< 0.001		< 0.001					
PL dec				< 0.001		< 0.001				
PH acc							< 0.005		0.23	
PH dec								< 0.001		0.85

showed significant differences ($p < 0.001$).

In the straight spine condition, the effect of muscle contraction was investigated for both the PL and PH acceleration shapes (Fig. 4). When the muscles were braced, the angles head bending to the right and left during the sled acceleration and deceleration phases were significantly reduced ($p < 0.001$) for both pulses. For the PL profile, mean values decreased from 9° for relaxed to 4° for braced muscles during the acceleration phase and changed from −5° for relaxed to 0° for braced muscles during the deceleration phase. For the PH profile, mean values decreased from 19° for relaxed to 13° for braced muscles during the acceleration phase and changed from −12° for relaxed to −2° for braced muscles during the deceleration phase.

However, in the relaxed muscle condition, no significant effect of spine posture (straight or sagging) was observed on head bending during either the acceleration or deceleration phase of PH acceleration ($p >$ 0.23). The lateral head bending corridors of the five volunteers are shown in Appendix 4. Only one volunteer showed greater lateral head bending during the deceleration phase than during the acceleration phase. For the PL profile, the mean values varied from 19° for the straight posture to 23° for the sagging posture during the acceleration phase, while they varied from −12° for the straight posture to −13° for the sagging posture during the deceleration phase.

4. Discussion

Lateral upper body bending was assessed for five seated volunteers subjected to different lateral accelerations. In previous studies focusing on the 50th-percentile male, the standard deviations of anthropometric variables were, on average, 6 cm for height and approximately 7 kg for weight. Differences in anthropometry were detected as a factor affecting the kinematic results (Reed et al., 2018a, 2018b). To diminish the effect of these differences, volunteers of the present study were recruited with very similar anthropometric characteristics (standard deviations of 2.5 cm in height and 3.8 kg in weight).

Based on several preliminary tests, maximum levels of acceleration were selected to ensure safety and avoid causing fatigue to the volunteers over the course of the test. In total, the 30-pulse experiment lasted <30 min, and none of the volunteers reported fatigue. Furthermore, no habituation effect was observed. Hence, it was hypothesized and verified through randomly repeated pulses over the seven different configurations that the volunteers' behaviour remained relevant throughout the experiment, allowing to include all three repetitions of each configuration when plotting the corridors.

Sled pulse reproducibility was assessed using sled accelerations measured underneath the seat plate. As an example, three of the measured pulses of each acceleration profile are compared with the input profile of the sled motor in Fig. 2. These curves correspond to the test configuration with straight spinal posture and relaxed muscles (pulses 4, 7 and 12 from Appendix 1). The sled accelerations, mechanical behaviour and applied boundary conditions can be considered reproducible and robust. Even if the acceleration measurements showed some vibrations, the sled displacement and velocity were smooth, with no jolts.

The applied accelerations in the current study lasted two or three seconds depending on the pulse shape, while previous sled experiments have been performed with shorter pulse durations: from 86 ms to 600 ms (Ejima et al., 2012; Dehner et al., 2007, 2008; Kumar et al., 2003; Siegmund et al., 2003; Vibert et al., 2001, 2006). Longer acceleration durations allow us to study the stabilization strategies of the human body when subjected to external dynamic perturbations while not triggering pure reflex reactions alone. They are also more representative of lateral avoidance manoeuvres that can occur during emergency driving situations (Ghaffari et al., 2018; Huber et al., 2014, 2015). Additionally, the acceleration pulse shape is often not well described in the literature. The results obtained in this study confirm the hypothesis that not only the amplitude but also the shape (sine or plateau) of the acceleration influences human body kinematics.

To create a generic experimental setup, allowing its physical and numerical replicability, the sled seat was designed using rigid structures. The influence of foam or spring deformation and the specific geometry of an individual vehicle seat usually introduce uncertainties and hinder the broad usage of the data. Additionally, to allow free movement of the upper body in the lateral plane, no seatback or shoulder belt was used to restrain the upper body. This also avoids obstructing the field of vision of the high-speed camera. In contrast to in-vehicle experiments, the data generated in this study are not directly transferable to the occupant kinematics expected to occur during emergency manoeuvres or to other very different sled configurations. Lateral head bending movements under similar loading conditions, although conducted in a vehicle and under accelerations up to 1 g, was found to be smaller than the movements observed in the current study (Ghaffari et al., 2018; Huber et al., 2014, 2015; Reed et al., 2018a, 2018b). However, the setup presented here allows for larger lateral excursions, not reduced by restraints, providing a qualified but more generic database for AHBM validation.

All lateral head bending measurements present a delay after the actual sled is set in motion. At the very beginning of the sled motion, the upper body can be considered a passive multibody inverted pendulum. Since the volunteers were seated, the pelvis was accelerated first, and the head was accelerated last. Moreover, under the relaxed muscular condition, the mean maximum head accelerations were always higher than those applied to the sled, reaching 0.30 g for SL, 0.75 g for SH, 0.43 g for PL, and 0.89 g for PH (Appendix 5). Consequently, lateral head bending is significantly larger for pulses with higher acceleration levels than for those with lower acceleration, as well as significantly larger for plateau pulses than for sine pulses.

The muscular state during PL and PH pulses influences the lateral head bending; specifically, bending is reduced when the muscles are braced. This supports the results from previous researchers showing that lateral displacement is reduced under braced conditions (Ejima et al., 2012; Holt et al., 2018; van Rooij et al., 2013a, 2013b), even without any upper body support structure.

Overall, no effect of spinal posture was observed on either head acceleration or lateral bending. Previous studies (Caneiro et al., 2010) reported differences in muscle tone depending on upper body posture in a static environment. In the straight spine condition, upper body posture was checked at shoulder height using a level horizontal laser line prior to each pulse to ensure a homogenous posture. No such check was performed for the sagging condition; therefore, greater variability is

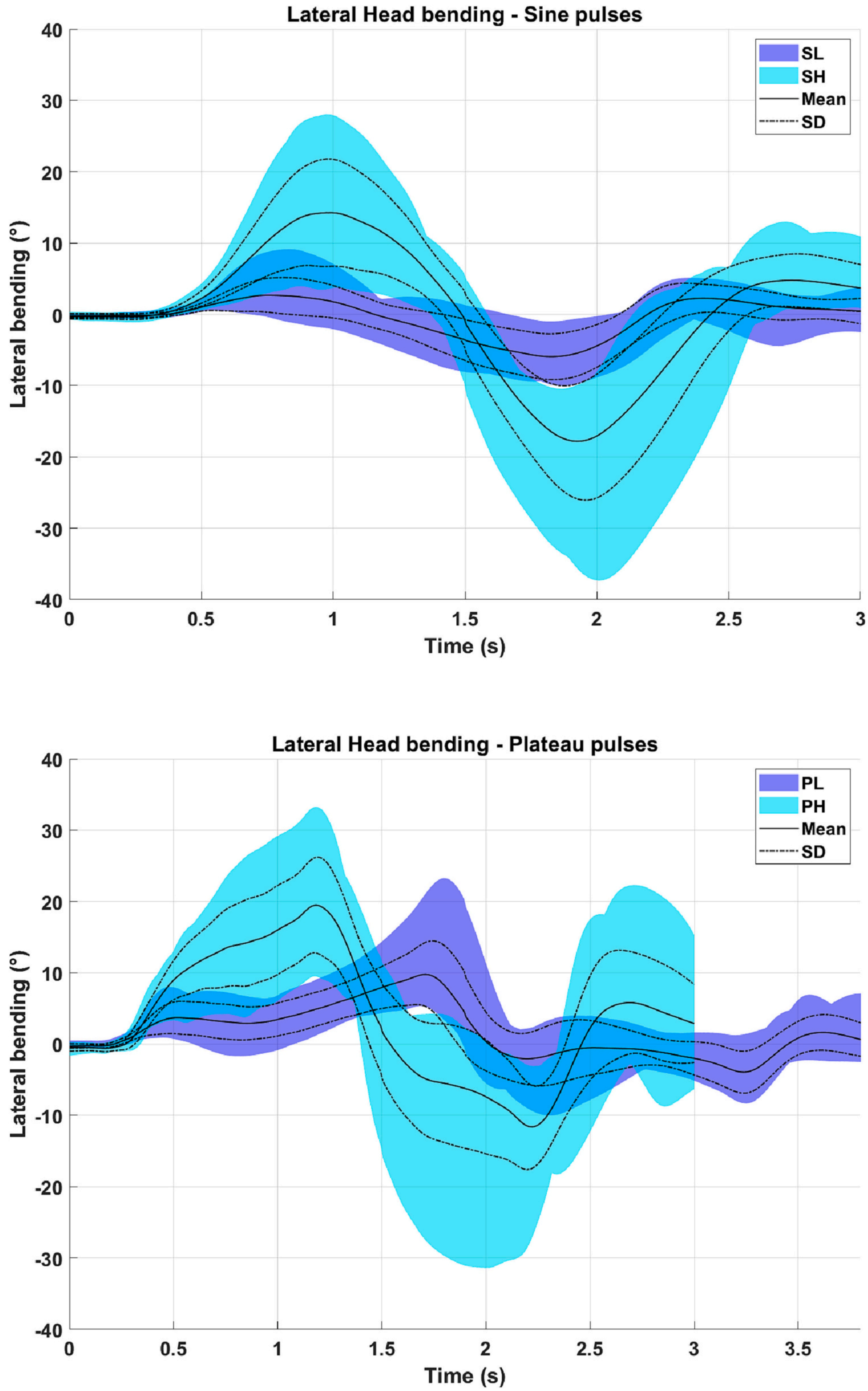
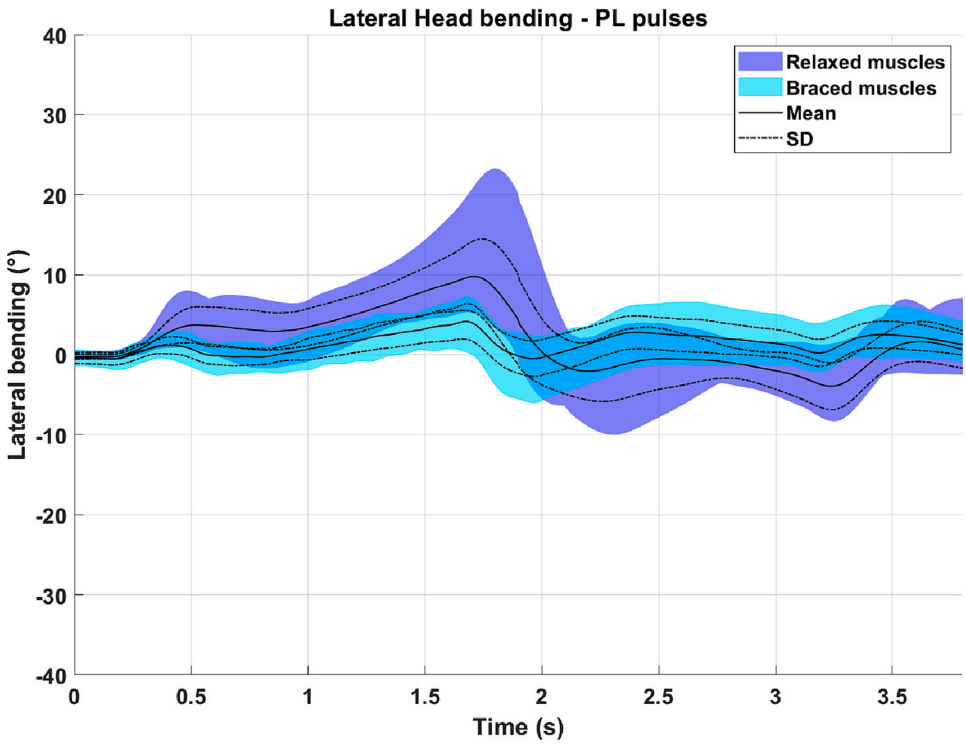


Fig. 3. Head lateral bending for pulses 1 to 12 of the five volunteers under various acceleration shapes. Straight spine posture and relaxed muscles. SL: Sine low, SH: Sine High, PL: Plateau Low, PH: Plateau High, SD: Standard Deviation.

expected for this posture. Although there was little variation in the anthropometric characteristics of the volunteers, each participant had his own individual spinal curvature. Nonetheless, it is assumed that the effect of this difference under lateral loading is negligible, since no significant differences were found between the sagging and straight postures. Future work on the difference in initial muscle states between these two configurations will be conducted.

The test matrix cannot combine all the chosen configurations, as the experiment was limited to 30 pulses per volunteer for ethical reasons. In particular, only the PH pulse shape was used to compare sagging and straight spine positions. Likewise, the influence of the muscular state was assessed only during the PH and PL pulses.



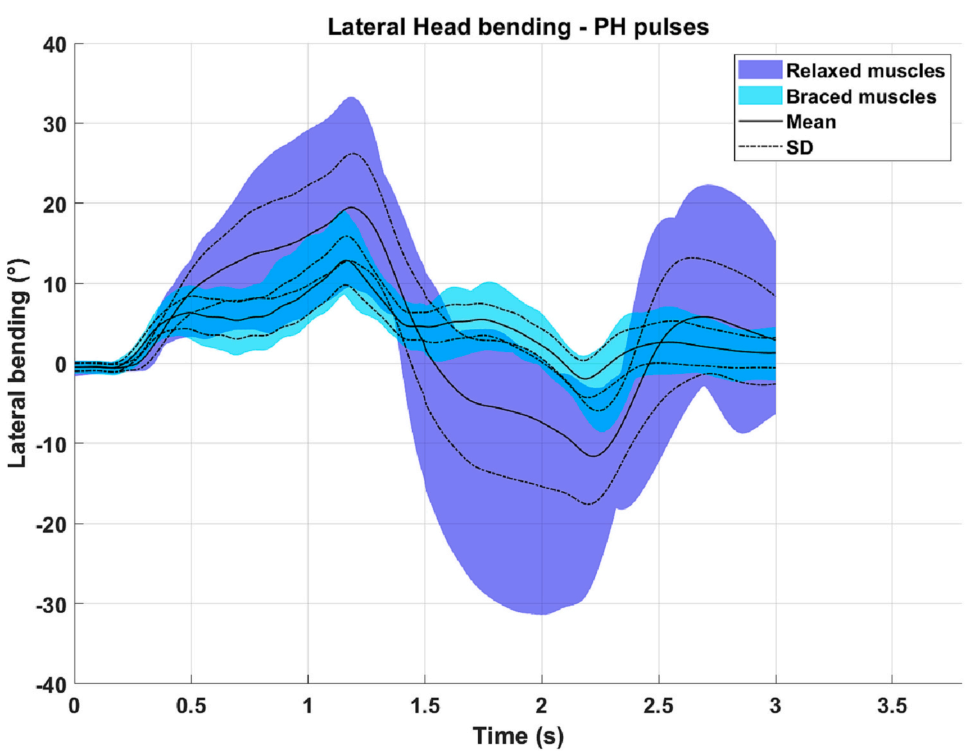


Fig. 4. Effect of the relaxed/ braced muscles condition for the Plateau Low (PL) and Plateau High (PH) acceleration shape inputs on the lateral head bending. Straight spine posture. SD: Standard Deviation.

Since the movement of the upper body was unrestricted in this experiment, the range of motion of the lower region of the body, starting at the pelvis, is a key factor that can influence overall lateral bending. Therefore, the pelvis of each volunteer was constrained using lateral brackets. Nonetheless, pelvic rotations could not be completely avoided (Appendix 5). In addition, the height of the lateral pelvic brackets was intentionally limited to avoid a potential interaction with the lower ribs during the lateral motion of the volunteers.

Finally, the whole-body braced muscle state was confirmed using EMG, but the level of contraction was not controlled in real time for each individual muscle. According to feedback by the volunteers, they braced all their muscles at once to their maximum extent during the corresponding pulses. In future work, recorded EMG data will be used to investigate different muscle contraction patterns in relation to the various configurations.

5. Conclusions

The results confirm the hypothesis that not only pulse amplitude but also shape influences human behaviour under low accelerations. In addition, upper body movement is reduced when muscles are braced during plateau pulses. However, spinal posture does not show a significant influence on lateral head bending during the PH pulses.

Overall, well-documented *in vivo* experimental data are of high interest for the validation of AHBMs. The present publication provides the necessary experimental data as supplementary material (SANDOZ, 2023).

Further experiments will be performed with female volunteers to evaluate potential differences in balance responses depending on sex.

Furthermore, the sled apparatus will be upgraded with additional sensors, such as pressure plates to record the interaction of the volunteers with the footrest. In addition, future experiments will use virtual reality technologies to control the visual and auditory environment of future volunteers to study the influence of these senses on stabilization strategies.

Declaration of Competing Interest

No competing financial interests exist.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2023.105952and https://doi.org/10.57745/I8KEDC.

References

Beeman, S.M., Kemper, A.R., Madigan, M.L., Duma, S.M., 2011. Effects of bracing on human kinematics in low-speed frontal sled tests. Ann. Biomed. Eng. 39 (12), 2998–3010. https://doi.org/10.1007/s10439-011-0379-1.

Beeman, S.M., Kemper, A.R., Madigan, M.L., Franck, C.T., Loftus, S.C., 2012. Occupant kinematics in low-speed frontal sled tests: human volunteers, hybrid III ATD, and PMHS. Accid. Anal. Prev. 47, 128–139. https://doi.org/10.1016/j.aap.2012.01.016.

Blouin, J.S., Descarreaux, M., Bélanger-Gravel, A., Simoneau, M., Teasdale, N., 2003. Self-initiating a seated perturbation modifies the neck postural responses in humans. Neurosci. Lett. 347 (1), 1–4. https://doi.org/10.1016/S0304-3940(03)00632-3.

Caneiro, J.P., O'Sullivan, P., Burnett, A., Barach, A., O'Neil, D., Tveit, O., Olafsdottir, K., 2010. The influence of different sitting postures on head/neck posture and muscle activity. Man. Ther. 15 (1), 54–60. https://doi.org/10.1016/j.math.2009.06.002.

Dehner, C., Elbel, M., Schick, S., Walz, F., Hell, W., Kramer, M., 2007. Risk of injury of the cervical spine in sled tests in female volunteers. Clin. Biomech. 22 (6), 615–622. https://doi.org/10.1016/j.clinbiomech.2007.02.004.

Dehner, C., Schick, S., Arand, M., Elbel, M., Hell, W., Kramer, M., 2008. Influence of anthropometry on the kinematics of the cervical spine and the risk of injury in sled tests in female volunteers. Accid. Anal. Prev. 40 (4), 1305–1312. https://doi.org/10.1016/j.aap.2008.02.002.

Dehner, C., Schick, S., Hell, W., Richter, P., Kraus, M., Kramer, M., 2013. In-vivo kinematics of the cervical spine in frontal sled tests. Global J. Health Sci. 5 (3), 115–126. https://doi.org/10.5539/gjhs.v5n3p115.

Ejima, Susumu, Ono, K., Holcombe, S., Kaneoka, K., Fukushima, M., 2007. A study on occupant kinematics behaviour and muscle activities during pre-impact braking based on volunteer tests. In: International Research Council on the Biomechanics of Injury - 2007 International IRCOBI Conference on the Biomechanics of Injury, Proceedings, pp. 31–45.

Ejima, Susumu, Zama, Y., Satou, F., Holcombe, S., Ono, K., Kaneoka, K., Shiina, I., 2008. Prediction of the physical motion of the human body based on muscle activity during pre-impact braking. In: International Research Council on the Biomechanics of Injury - 2008 International IRCOBI Conference on the Biomechanics of Injury, Proceedings, pp. 163–175.

Ejima, S., Ito, D., Satou, F., Mikami, K., Ono, K., Kaneoka, K., Shiina, I., 2012. Effects of pre-impact swerving/steering on physical motion of the volunteer in the low-speed side-impact sled test. In: 2012 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury, pp. 352–366.

Ewing, C.L., Thomas, D.J., Lustick, L., Muzzy, W.H., Willems, G., Majewski, P.L., 1976, February 1. The effect of duration, rate of onset, and peak sled acceleration on the dynamic response of the human head and neck. In: 20th Stapp Car Crash Conference. https://doi.org/10.4271/760800.

Ghaffari, G., Brolin, K., Brase, D., Pipkorn, B., Svanberg, B., Jakobsson, L., Davidsson, J., 2018. Passenger kinematics in lane change with brakingmanoeuvres using two belt configurations: standard and reversiblepre-pretensioner. In: International Research Council on the Biomechanics of Injury - 2018 International IRCOBI Conference on the Biomechanics of Injury, Proceedings, pp. 493–511.

Graci, V., Douglas, E., Seacrist, T., Kerrigan, J., Mansfield, J., Bolte, J., Sherony, R., Hallman, J., Arbogast, K., 2019. Effect of automated versus manual emergency braking on rear seat adult and pediatric occupant precrash motion. Traffic Injury Prevent. 20 (sup1), S106–S111. https://doi.org/10.1080/15389588.2019.1630734.

Graci, V., Douglas, E., Seacrist, T., Kerrigan, J., Mansfield, J., Bolte, J., Sherony, R., Hallman, J., Arbogast, K., 2020. Age differences in occupant motion during simulated in-vehicle swerving maneuvers. Int. J. Environ. Res. Public Health 17, 1834. https://doi.org/10.3390/ijerph17061834.

Hernández, I.A., Fyfe, K.R., Heo, G., Major, P.W., 2005. Kinematics of head movement in simulated low velocity rear-end impacts. Clin. Biomech. 20 (10), 1011–1018. https://doi.org/10.1016/j.clinbiomech.2005.07.002.

Higuchi, K., Forman, J.L., Kent, R.W., Arbogast, K.B., 2020. Behavior of ATD, PMHS and human volunteer in crash test II. Int. J. Automot. Eng. 11 (2), 49–56. https://doi.org/10.20485/jsaeijae.11.2_49.

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. In: 2018 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury.

Holt, C., Seacrist, T., Douglas, E., Graci, V., Kerrigan, J., Kent, R., Balasubramanian, S., Arbogast, K., 2020. The effect of vehicle countermeasures and age on human volunteer kinematics during evasive swerving events. Traffic Injury Prevent. 21 (1), 48–54. https://doi.org/10.1080/15389588.2019.1679798.

Huber, P., Kirschbichler, S., Prüggler, A., Steidl, T., 2014. Three-dimensional occupant kinematics during frontal, lateral and combined emergency maneuvers. In: 2014 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury, pp. 668–669.

Huber, P., Kirschbichler, S., Prüggler, A., Steidl, T., 2015. Passenger kinematics in braking, lane change and oblique driving maneuvers | enhanced reader. In: 2015 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury.

Kemper, A.R., Beeman, S.M., Madigan, M.L., Duma, S.M., 2014. Human occupants in low-speed frontal sled tests: effects of pre-impact bracing on chest compression, reaction forces, and subject acceleration. Traffic Injury Prevent. 15 (sup1), S141–S150. https://doi.org/10.1080/15389588.2014.938323.

Kirschbichler, S., Prüggler, A., Steiner, K., 2011. Detailed analysis of 3D occupant kinematics and muscle activity during the pre-crash phase as basis for human modeling based on sled tests. In: 22nd International Conference on the Enhanced Safety of Automobiles Conference, pp. 1–8.

Kirschbichler, S., Huber, P., Prüggler, A., Steidl, T., Sinz, W., Mayer, C., D'Addetta, G.A., 2014. Factors influencing occupant kinematics during braking and lane change maneuvers in a passenger vehicle. In: 2014 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury, pp. 614–625.

Kumar, S., Narayan, Y., Amell, T., 2000. Role of awareness in head-neck acceleration in low velocity rear-end impacts. Accid. Anal. Prev. 32 (2), 233–241. https://doi.org/10.1016/S0001-4575(99)00114-1.

Kumar, S., Narayan, Y., Amell, T., 2003. Analysis of low velocity frontal impacts. Clin. Biomech. 18 (8), 694–703. https://doi.org/10.1016/S0268-0033(03)00137-2.

Lopez-Valdes, F.J., Juste, O., Lorente, A.I., Piqueras, A., Maza, M., Muehlbauer, J., Schick, S., Pipkorn, B., Mroz, K., Peldschus, S., 2017. Comparison of the kinematics and dynamics of the THOR-50M dummy and elderly volunteers in low-speed frontal decelerations. In: Conference Proceedings International Research Council on the Biomechanics of Injury, IRCOBI, 2017-Septe, pp. 272–273.

Mackenzie, J., Dutschke, J., Di Loreto, C., Forrest, M., Van Den Berg, A., Merienne, F., Chardonnet, J.R., Sandoz, B., 2022. Human head and neck kinematics during autonomous and human braking in three initial head positions. SAE Int. J. Trans. Safety 10 (2), 121–133. https://doi.org/10.4271/09-10-02-0007.

Morris, R., Cross, G., Bingley, L., 2005. Improved understanding of passenger behaviour during pre-impact events to aid smart restraint development. In: In *Proceedings of the 19th International Technical Conference on the Enhanced Safety of Vehilces (ESV)*. Washington DC, USA, pp. 6–9.

Parenteau, C., 2006. A comparison of volunteers and dummy upper torso kinematics with and without Shoulder Belt slack in a low speed side/pre-roll environment. Traffic Injury Prevent. 7 (2), 155–163. https://doi.org/10.1080/15389580500481973.

Parenteau, C., Shah, M., Steffan, H., Hofinger, M., 2002. Volunteer and dummy head kinematics in low-speed lateral sled tests. Traffic Injury Prevent. 3 (3), 233–240. https://doi.org/10.1080/15389580213642.
Park, G., Kim, T., Crandall, J.R., Arregui-dalmases, C., Luzon-narro, J., 2013. Comparison of kinematics of GHBMC to PMHS on the side impact condition. In: 2013 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury, pp. 368–379.
Reed, M.P., Ebert, S., Jones, M.H., Park, B.D., Hallman, J., Sherony, R., 2018a. Passenger head kinematics in abrupt braking and lane change events. Traffic Injury Prevent. 19 (sup2), S70–S77. https://doi.org/10.1080/15389588.2018.1481957.
Reed, M.P., Park, B.K., Ebert, S., Hallman, J.J., Sherony, R., 2018b. Marker-less tracking of head motions in abrupt vehicle manoeuvres. In: Conference Proceedings International Research Council on the Biomechanics of Injury, IRCOBI, 2018-Septe, pp. 561–562.
Reed, M.P., Ebert, S., Jones, M.H., Park, B.D., 2021. Occupant Dynamics during Crash Avoidance Maneuvers. Édité par University of Michigan. Transportation Research Institute, no DOT HS 812 997. https://rosap.ntl.bts.gov/view/dot/54737.
Sandoz, B., Simonin, A., Saletti, D., Laporte, S., 2014. Sled acceleration control for low-speed impact testing and transient response studies. Comp. Methods Biomech. Biomed. Eng. 17 (sup1), 48–49. https://doi.org/10.1080/10255842.2014.931105.
Sandoz, B., Rodrigues, L., Pitman, J., Jem, N., Subit, D., Vidal, P.-P., Laporte, S., 2016. Vision effect on the head stabilization during controlled linear acceleration of seated volunteers. ESB Congress 4 (1).
Sandoz, B., Sugiyama, T., Weber, W., 2019. Upper body in vivo kinematics during moderate lateral accelerations – towards active finite element human body model validation. In: IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury, IRC-19-28.
SANDOZ, Baptiste, 2023. Lateral sled experiments on human seated volunteers. https://doi.org/10.57745/I8KEDC. Recherche Data Gouv, V2, UNF:6: wU3EJ2vkkQyWLaygS4flcQ== [fileUNF].
Serre, T., Masson, C., Perrin, C., Chalandon, S., Llari, M., Py, M., Cavallero, C., Cesari, D., 2007. Real accidents involving vulnerable road users: in-depth investigation, numerical simulation and experimental reconstitution with PMHS. Int. J. Crashworthiness 12 (3), 227–234. https://doi.org/10.1080/13588260701441050.
Siegmund, G.P., Sanderson, D.J., Myers, B.S., Timothy Inglis, J., 2003. Rapid neck muscle adaptation alters the head kinematics of aware and unaware subjects undergoing multiple whiplash-like perturbations. J. Biomech. 36 (4), 473–482. https://doi.org/10.1016/S0021-9290(02)00458-X.
van Rooij, L., Elrofai, H., Philippens, M.M., Daanen, H.A., 2013a. Volunteer kinematics and reaction in lateral emergency maneuver tests. Stapp Car Crash J. 57, 313–342.
van Rooij, L., Elrofai, H., Philippens, M.M.G.M., Daanen, H.A.M., 2013b. Corrigendum to volunteer kinematics and reaction in lateral emergency maneuver tests [Stapp Car crash journal 57 (2013) 313-342]. Stapp Car Crash J. 57, 313–342. https://doi.org/10.4271/2013-22-0013.01.
Vibert, N., MacDougall, H.G., de Waele, C., Gilchrist, D.P.D., Burgess, A.M., Sidis, A., Migliaccio, A., Curthoys, I.S., Vidal, P.P., 2001. Variability in the control of head movements in seated humans: a link with whiplash injuries? J. Physiol. 532 (3), 851–868. https://doi.org/10.1111/j.1469-7793.2001.0851e.x.
Vibert, N., Hoang, T., Gilchrist, D.P.D., MacDougall, H.G., Burgess, A.M., Roberts, R.D., Vidal, P.P., Curthoys, I.S., 2006. Psychophysiological correlates of the inter-individual variability of head movement control in seated humans. Gait Posture 23 (3), 355–363. https://doi.org/10.1016/j.gaitpost.2005.04.006.