Effect of Countermeasures on Adult Kinematics during Pre-Crash Evasive Swerving

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Abstract Low-acceleration time-extended (LATE) events, such as evasive swerving, often precede a crash event. The inertial forces during LATE events have the potential to cause changes to the occupant's initial state (initial posture, position, muscle tension). The objective of this study is to systematically quantify the kinematics of adult human volunteers during a simulated pre-crash evasive swerving manoeuvre and evaluate the effect of bracing and two vehicle-based countermeasures. A novel laboratory device was designed to expose subjects to non-injurious loading conditions that mimic real-world evasive swerving events. Adult subjects (n=19, age: 26.0 ± 6.8 years) were exposed to a series of test conditions (relaxed, braced, pre-pretensioned seat belt, sculpted vehicle seat with and without inflated torso bolsters) while their kinematics were captured using 3D motion-capture and muscle activity was recorded. Similar reductions in the head and lateral trunk displacement were achieved by actively bracing and by the implementation of a pre-pretensioned seatbelt. The second countermeasure – a sculpted seat with inflatable torso bolsters – did not show similar benefits in reducing maximum lateral displacement. Differences in kinematics existed across subsequent oscillations within a given test, with the first oscillation demonstrating the largest displacement, suggesting that active neuromuscular strategies are being employed to counteract motion.

Keywords EMG, Human Volunteers, Occupant Kinematics, Sled Testing, Swerving Manoeuvre.

I. INTRODUCTION

Safety countermeasures are designed to minimise contact between the occupant and the vehicle interior and to extend the time over which crash loads are applied, thereby significantly reducing the risk of injury. Vehicle safety restraint devices are optimised by performing laboratory-based tests using anthropomorphic test devices (ATDs) and/or via simulation with computational human body models (HBMs). Current testing protocols primarily evaluate restraint performance with an optimally positioned ATD or HBM (e.g. seated upright against the seat back, gaze forward and upper extremities at the side). However, these idealised testing positions may not reflect the occupant's pre-crash state in real-world scenarios. Previous research has shown that 60% of crashes involve some form of pre-crash manoeuvres [1]. Pre-crash manoeuvres are defined as low acceleration, time extended (LATE) events and comprise a spectrum of dynamic impact avoidance manoeuvres that are exhibited in critical driving situations [9-102-3]. Pre-crash manoeuvres can displace occupants away from idealised seating positions and may induce bracing or muscle tensing as reaction strategies. In turn, the occupant's pre-crash state (posture, position, muscle tension) may alter the restraint system's performance and potentially reduce occupant protection. For example, sub-optimal pre-crash occupant position had a significant effect on injury outcomes for passengers who were killed by airbag deployments in the early 1990s [4].

Alterations in occupant positioning resulting from pre-crash manoeuvres and the associated injury risks have not been studied extensively, therefore the robustness of restraint systems to accommodate changes in occupant size and state remains largely unknown. It is imperative to study occupant kinematics during pre-crash events because the optimisation of restraint systems requires performance in these conditions. Evasive swerving, defined as a vehicle generating a primarily lateral sinusoidal or oscillating path, is a LATE event that is currently understudied and is therefore the focus of this study.

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Head injuries that result from head contact with the vehicle interior are one of the most common serious injuries sustained by occupants in car crashes. The Centers for Disease Control and Prevention determined motor vehicle crashes were the third leading cause of traumatic brain injuries (TBI) among all age groups. [5]. The inertial forces during evasive swerving have the potential to displace the occupant and reduce the ability of the restraint system to retain the torso, thus increasing the likelihood of head contact with the vehicle interior. This problem exists due to a passenger's head having proximity to side structures such as the 'B/C' pillars, side window glass and side door [6]. Since the early 1980s researchers have agreed head injuries have been the most difficult to mitigate via vehicle design [7].

Little is known about the phasing and magnitude of occupant kinematics during evasive swerving and how it varies by age and key restraint parameters. Previous studies have investigated the kinematic response of ATDs in manoeuvres that include a substantial lateral component, such as evasive swerving. These studies concluded that ATD kinematic response time was faster than that of a human and that their lateral displacements were overestimated [8][9]. Studies then began to investigate human volunteer biomechanical data in response to pre-crash lateral manoeuvres. Many of these studies, whether using a closed track or a laboratory setting, used acceleration pulses derived from vehicle handling tests (e.g. Fishhook, VDA) to mimic a single or double lane change. A series of experiments evaluated the adult occupant kinematics on a closed track setting during an emergency lane-change manoeuvre [10][11][12]. This series of tests found distinct inter-subject variability of head and trunk displacements, and these displacements were significantly influenced by different levels of occupant awareness to the testing condition. It was determined that a swerving manoeuvre initially performed towards the right results in faster activation of right-side muscles compared to left-side muscles. Ejima *et al*. conducted laboratory sled tests that exposed human volunteers to 0.4 g and 0.6 g of pure lateral accelerations, and investigated the posture of volunteers in a driver position as they simulated an active steering response at the onset of the swerve. Results revealed that muscle activation, or bracing, significantly influenced occupant's behaviour during these low-speed lateral manoeuvres [13]. Adult kinematic behaviour and muscular activity were also evaluated using a robot vehicle that achieved peak lateral accelerations up to 0.7 g in a laboratory environment [14]. Bracing was once again determined to decrease the maximum displacement of a majority of body regions by 28–38% compared to a relaxed posture. Parenteau *et al*. compared occupant kinematics with and without shoulder-belt slack in dynamic lateral manoeuvres that had the potential to induce shoulder-belt slip-out. Results show lateral accelerations displaced passengers more than drivers [8], indicating the role of awareness in the impending vehicle dynamics.

This study seeks to contribute to the literature by investigating human volunteer kinematics as a result of several cycles of reversing lateral acceleration rather than a single unidirectional pulse. Morris *et al*. hypothesised that several cycles of reversing lateral acceleration would provide sufficient time for occupants to adapt to the loading condition and find a suitable structure to brace against to restrict motion [15]. This study broadly aims to systematically quantify the kinematic responses of restrained paediatric and adult human volunteers during a simulated evasive swerving manoeuvre and evaluate the effects of age, two safety countermeasures (e.g. pre-pretensioning and inflated torso bolsters) and intentional muscle response (e.g. bracing) on occupant kinematics. The analyses presented here will focus on the kinematic responses of adult volunteers in loading conditions that mimic evasive swerving and also the effect of bracing and specific restraint- and vehicle-based countermeasures.

II. METHODS

A. Volunteers and Informed Consent

Healthy male volunteers whose height, weight and BMI were within 5th and 95th percentile (based on CDC NHANES data) were included in the study. Adult subjects were between 18 and 40 years of age. Subjects with a proclivity to motion sickness, existing neurologic, orthopedic or neuromuscular conditions and any previous injury or abnormal pathology relating to the head, neck or spine were excluded from the study. A total of 19 male subjects within the 18–40 year age range were evaluated. Subject anthropometry is shown in Table I. The study protocol was reviewed and approved by the Institutional Review Board of the Children's Hospital of Philadelphia and informed consent was obtained from all volunteers.

TABLE I
AVERAGE VOLUNTEER DEMOGRAPHICS AND ANTHROPOMETRY
(N=19)

Age (years)	26.0 ± 6.8
Height (cm)	175.9 ± 6.3
Weight (kg)	75.7 ± 12.4
Seated Height (cm)	86.4 ± 4.0
BMI (kg/m²)	24.4 ± 3.0

B. Sled Apparatus for Evasive Swerving

Subjects were exposed to loading conditions that mimic evasive swerving via a safe and repeatable testing device. Figure 1 illustrates the evasive swerving simulator sled (hereafter 'the LATE device'.) A meta-analysis of previous laboratory, on-road studies, consumer information programs and safety standards was conducted to determine the appropriate oscillatory acceleration and magnitude that is safe for human subject testing and also representative of dynamic pre-crash field data. This analysis revealed that previous pre-crash swerving test conditions ranged from 0.5 g to 1.2 g in lateral acceleration [16]. Based on this data, the acceleration pulse for the LATE device was developed. The LATE device is capable of delivering up to 1 g of lateral sinusoidal oscillatory acceleration; the target acceleration for these tests was 0.75 g. This loading environment was determined to be safe for human volunteers and relevant to real-world swerving events.

Fig. 1. The LATE device that mimics a pre-crash swerving event.

A Scotch Yoke mechanism drives the LATE device, which involves two WEG W21 5 HP motors powering a sprocket. The gear system is coupled to a sliding yoke on the bottom of a 1.5 m x 0.9 m cart. In turn, the cart is actuated to glide along 3.7 m steel rails via near-frictionless Teflon shoes.

An occupant compartment was mounted to the cart such that the motion was perpendicular to the occupant. The seating compartment interior was designed to simulate a passenger environment relevant to either the first- or second-row and consisted of a production vehicle seat. The vehicle seats were locked to a stationary position. A 3-point seatbelt was affixed to an adjustable B-/C-pillar and D-ring structure. The B-/C-pillar maintained 3 degrees of freedom, which provided aft, fore-aft, vertical, and lateral adjustability.

An onboard motion-capture system, consisting of eight Optitrack Flex 13 infrared cameras (120 Hz NaturalPoint, Inc., Corvallis, OR), was located around the perimeter of the compartment. Two onboard GoPro HERO Session 4 cameras were oriented in the overhead and frontal perspective of the volunteers. High-speed 2D videos from these cameras were captured at 60 Hz. A triaxial force plate was mounted to the footrest to obtain loads exerted from the lower extremities. An accelerometer (Endevco 7290a-10), located on the bottom right of the seating compartment, measured the compartment acceleration. The acceleration data were filtered with a 4-pole Butterworth filter. The accelerometer and force plate data were sampled at 10,000 Hz using an onboard T-DAS data acquisition system (Model T-DAS Pro, DTS Inc, Seal Beach, CA).

C. Vehicle Countermeasures

The effect of bracing and two countermeasures – a seat belt pre-pretensioner and inflatable torso bolsters – were assessed. To assess bracing as a volunteer-induced countermeasure, a removable bracing structure was attached to the outboard (right) side of the compartment in immediate reach of the subject, similar to a vehicle interior door structure. The bracing structure was designed to accommodate a wide range of subject arm-forearm lengths and to allow the subject to brace comfortably.

An automotive 3-point belt system with three seat-belt load cells was used, along with an electromechanical motorised seatbelt retractor (MSB), known as a pre-pretensioner. The reversible pre-tensioned belt was integrated into the shoulder belt (Takata Corporation, Japan). The purpose of this countermeasure was to remove slack from the shoulder belt and improve coupling of the occupant to the vehicle once the vehicle detected an imminent crash. This is achieved by the seatbelt retractor deploying a pre-pretensioning load of approximately 200 N. The pre-pretensioner was powered by a 12V-20A electrical output and was integrated into the sled trigger, enabling it to be activated simultaneously at test start. The restraint system was incorporated into the LATE device such that the pre-pretensioner was only active during selected tests.

Fig. 2. Standard (A) and Sculpted seat (B) used during testing.

The LATE device was also designed such that two different vehicle seats could be accommodated: a second-row captain's chair from a recent model year minivan (standard seat); and a front-row, more sculpted seat from a recent model year sports sedan (Fig. 2). Both seats were manufactured with leather upholstery, and the head restraints were removed prior to testing for better visibility of the head markers. The seating configuration was selected to match the interior of a common US passenger vehicle. The seat pan surface for the standard and sculpted seat (width: 50 cm, 54 cm) was inclined 21° and 14°, respectively. Specific mounting requirements for each seat contributed to the differences in the selected seat pan angles. The recline angle, between the seat back and the horizontal, was set to 115° for both seats. The seatback angles were selected to mimic the rear seat geometry of a standard, passenger vehicle. The sculpted seat was equipped with a pneumatic lateral support safety countermeasure (inflatable torso bolsters) designed to restrict torso motion during dynamic lateral manoeuvres. For the relevant tests, the left and right torso bolsters were inflated to 13.8 kPa, using a regulated air compressor, before the test start.

Fig. 3. Initial position of adult volunteer.

D. Experimental Conditions

Volunteers were exposed to a series of non-injurious low-speed tests. The test matrix is shown in Table II. Subjects were positioned as a right-side passenger, therefore the shoulder belt spanned across the subject's right shoulder (mid- clavicle) and buckled at their left hip. The initial position of the lap belt was below the anterior superior iliac spine. To ensure similar seatbelt fit among the full range of subject sizes, a rigorous custom seat belt donning procedure was followed where anatomical landmarks were used as the primary reference points to establish optimal subject specific D-Ring positioning and ultimately belt fit. The initial position for each subject was gaze forward, feet flat and centered on the footrest, and hands placed on the lap (except for the braced condition) (Fig. 3).

Each trial was composed of four cycles of oscillation, with a frequency of 0.5 Hz. One complete cycle of oscillation required the seating compartment to travel 1.8 m laterally, in the +y direction (towards the volunteer's right) and then return 1.8 m to the starting position.

Before testing, each volunteer performed a series of exercises to collect a Maximum Voluntary Isometric Contraction (MVIC). Exercises were carried out on an apparatus custom designed to target each muscle examined in the study. Volunteers were encouraged to maintain a maximum contraction for 10 seconds.

In total, each subject participated in five testing conditions, each repeated once for a total of 10 trials. Test conditions were randomised within seat type (e.g. within the standard seat, Baseline, Braced and MSB were randomised). Repeat trials were not randomised. Subjects did not receive any visual or auditory cues to indicate the exact time of test start. This was done to capture the response of an unsuspecting occupant.

TABLE II
TEST MATRIX

Baseline	**MSB**	**Braced**	**Uninflated**	**Inflated**
Relaxed	Relaxed	Structure to hold	Relaxed	Relaxed
Standard restraint	Motorised seat belt	Standard restraint	Standard restraint	Standard restraint
Standard seat	Standard seat	Standard seat	Sculpted seat	Sculpted seat
	Motorised seat belt		Inflatable seat torso bolsters – unfilled	Inflatable seat torso bolsters - filled

E. Occupant Instrumentation

Volunteers were provided with an athletic compression shirt and a pair of athletic shorts. Once in proper attire, photo-reflective markers were attached to the head, torso and extremities. For the head markers, volunteers wore a tightly fitted headpiece with five head markers (top, bilaterally on the front and back). In total, eight markers were placed on the head (*bilateral:* front of the head, back of the head, external auditory meatus; *central:* top of the head, fourth cervical vertebrae, nasion). In total, 13 markers were placed on the torso and extremities (*bilateral:* acromion, humeral epicondyle, ulnar styloid process, femoral epicondyle, lateral malleolus; *central:* suprasternal notch, first thoracic vertebra, xiphoid process) (Fig. 4).

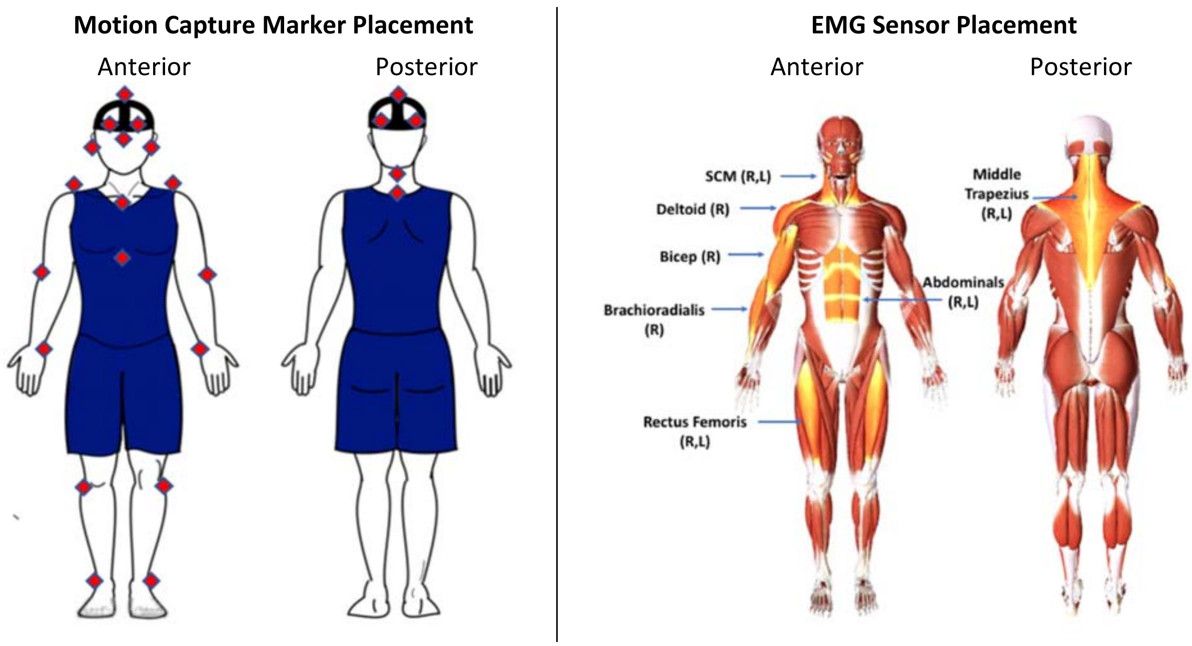


Fig. 4. Volunteer marker placements and EMG sensor locations.

Surface electromyography (sEMG) electrodes (EMG Delsys, Inc., 2000 Hz) were then placed bilaterally on the subject's: neck (sternocleidomastoid (SCM), middle trapezius); torso (abdominals); lower extremities (rectus femoris); and unilaterally on the right upper extremity (deltoid, bicep, brachioradialis).

F. Data Processing

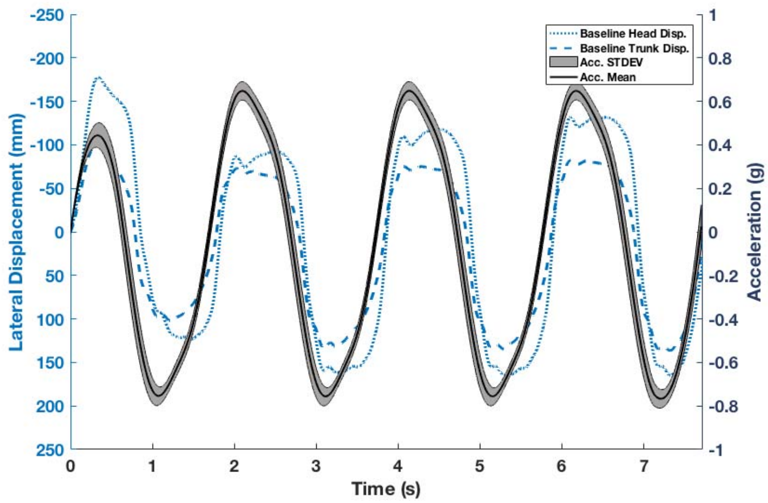
The motion analysis data were processed using Motive: Tracker software (NaturalPoint, Inc., Corvallis, OR). The kinematic measures of interest include the maximum head and torso lateral displacement. The photo-reflective marker located on the top of the headpiece and the marker on the supra-sternal notch (SSN) defined the head top and torso midline, respectively. Displacement was measured by quantifying the motion of the marker in the lateral (y) direction, relative to initial position at event onset. Thus, +y represents the volunteer moving toward the upper anchorage (into the belt) and –y is away from the upper anchorage (out of the belt). For this analysis, only maximum displacements out of the belt (as volunteers moved away from the upper anchorage) were reported. Shoulder-belt slip-out was determined using methods provided by Seacrist *et al*. [9].

Muscle activation during each trial was measured using sEMG electrodes and analysed after systematic processing. The raw EMG signals were filtered with a Band-pass filter (20–500 Hz, filter order: 558) based on the Finite Impulse Response (Kaiser Window method) filter [17][18][19]. A root-mean-square (RMS) method with a 200 ms moving average smoothing window was then applied. For each subject, the processed EMG signals were normalised by their respective MVIC. Mean EMG responses were analysed with respect to key event time points, derived from the acceleration pulse, such as the onset and offset for each direction (directions include the +y and –y of the seating compartment) of the cycle (into belt and out of belt). Custom MATLAB (Mathworks, Inc., Natick, MA) algorithms were used to analyse the motion capture and EMG data. This evaluation included data from four muscles: the trapezius bilaterally (right and left) was chosen due to its relationship to head and trunk stabilisation, and two muscles of the right extremity (bicep and brachioradialis) were selected as they are the primary muscles involved in bracing.

Four separate one-way repeated measure ANOVAs were performed on the maximum lateral head top displacement, maximum lateral trunk displacement, and mean EMG of the right bicep, brachioradialis and trapezius bilaterally to determine differences between test conditions (5 levels) and cycles (4 levels). Post-hoc tests were performed using Tukey's HSD, with significance level set to p=0.05.

III. RESULTS

The following represents head and trunk kinematic data from 19 adult subjects (age: 26.0 ± 6.8 years, height: 175.9± 6.3 cm, weight: 75.7 kg ± 12.4 kg). The acceleration profile is presented in Fig. 5. Figure 6 depicts the time series of the head (left) and trunk (right) lateral motion for all five conditions. The subjects' motion is relative to the moving compartment. As the cart moves to the subjects' right, the initial movement is to the subjects' left, out of the shoulder belt. Subjects overall moved less in the Braced and MSB conditions than in the Baseline, Uninflated and Inflated conditions. Greater motion was observed in the head than the trunk.



N=490

Fig. 5. The acceleration profile of the LATE device. The mean of all subject tests and trials is represented by the black solid line, while the standard deviation is depicted by the shaded area. The dotted and dashed lines represent the average head top and trunk displacement for the baseline condition respectively.

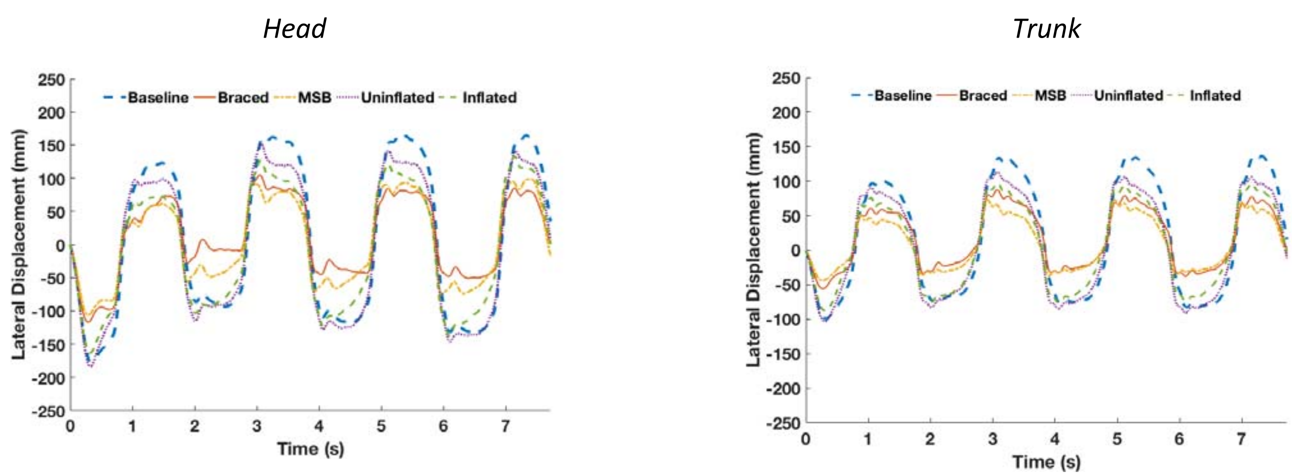


Fig. 6. Average lateral displacement for the head (left) and trunk (right) for each condition.

Maximum head and trunk kinematics varied across test conditions (p<0.001) (Fig. 7). Post-hoc comparisons revealed that maximum displacements for the head and trunk in both the MSB and Braced conditions were smaller than in Baseline, Uninflated and Inflated (head: p<0.001; trunk: p<0.001). No differences were detected between the following relevant comparisons: baseline and uninflated (head: p=0.99; trunk: p=0.99); baseline and inflated (head: p=0.86; trunk: p=0.48); and MSB and Braced (head: p=0.91; trunk: p=0.94).

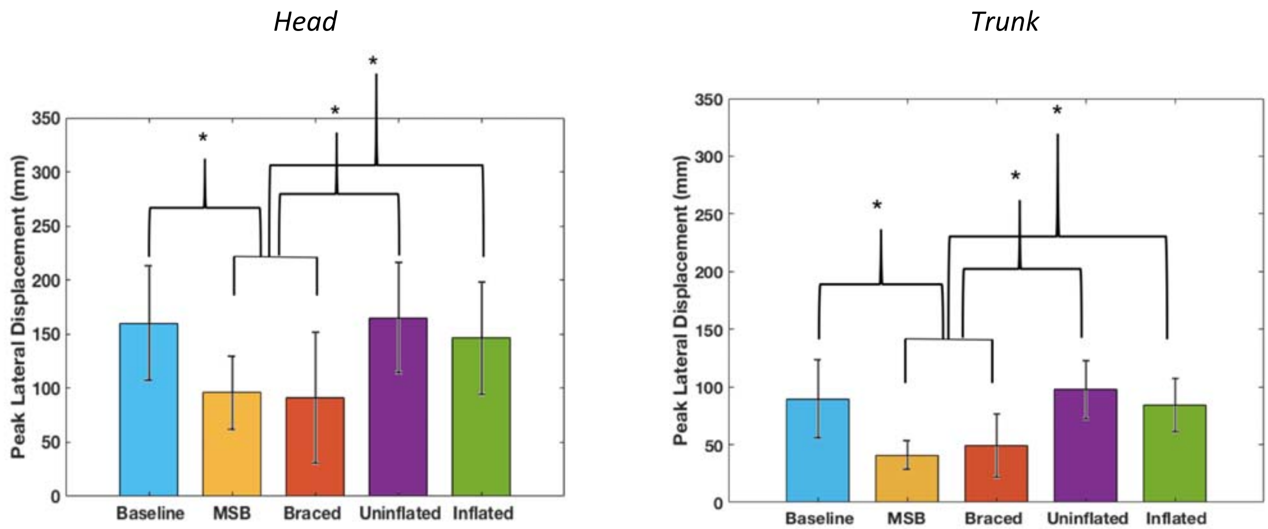


Fig. 7. Influence of test condition on the average peak lateral displacement for the head (left) and trunk (right). Error bars represent the standard deviation. (Averaged across all cycles) (*P<0.001).

Regarding cycles, the first cycle produced the greatest displacement for both the head and trunk (Fig. 8). For the head, each cycle was significantly different from one another ($p<0.03$). Concerning the trunk motion, cycle one was significantly greater than the remaining cycles ($p<0.001$) and cycle 2 was significantly less cycle 4 (p=002). There were no significant differences found between cycles 2 and 3 and cycles 3 and 4.

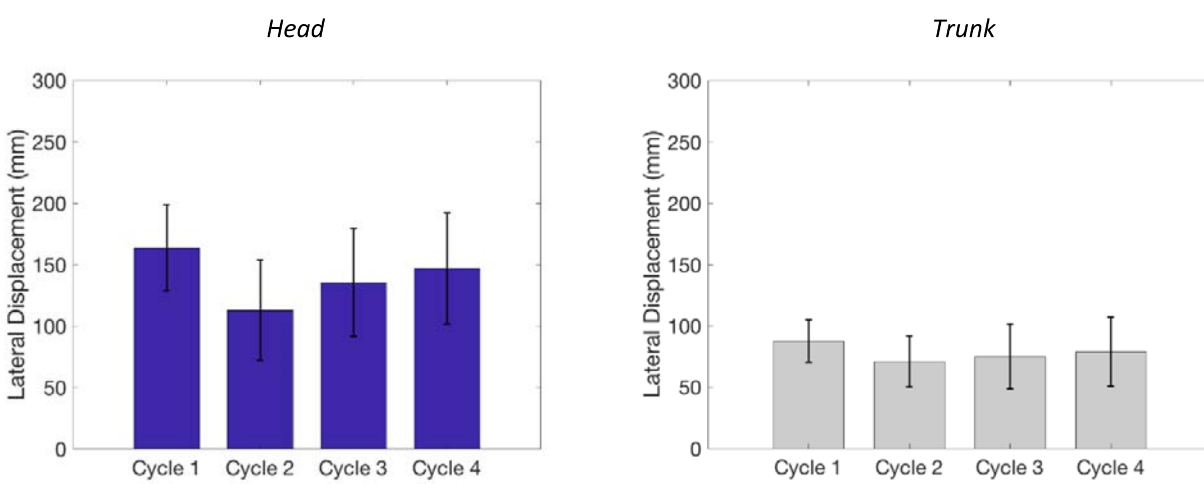


Fig. 8. Average peak lateral displacement and the standard deviations for the head (left) and trunk (right) per cycle.

For the right bicep, trapezius and brachioradialis (BRD), no significant differences in muscle activity were found between Baseline and Uninflated, Baseline and Inflated, and Uninflated and Inflated (R. bicep: $p>0.17$; R. trap: $p>0.77$; R. BRD: $p>0.76$) (Fig. 9). Despite similar kinematics, greater activity in both muscles was found in the Braced condition compared to MSB ($p<0.001$). No significant differences were found the between the test conditions for the left trapezius muscle ($p=0.117$).

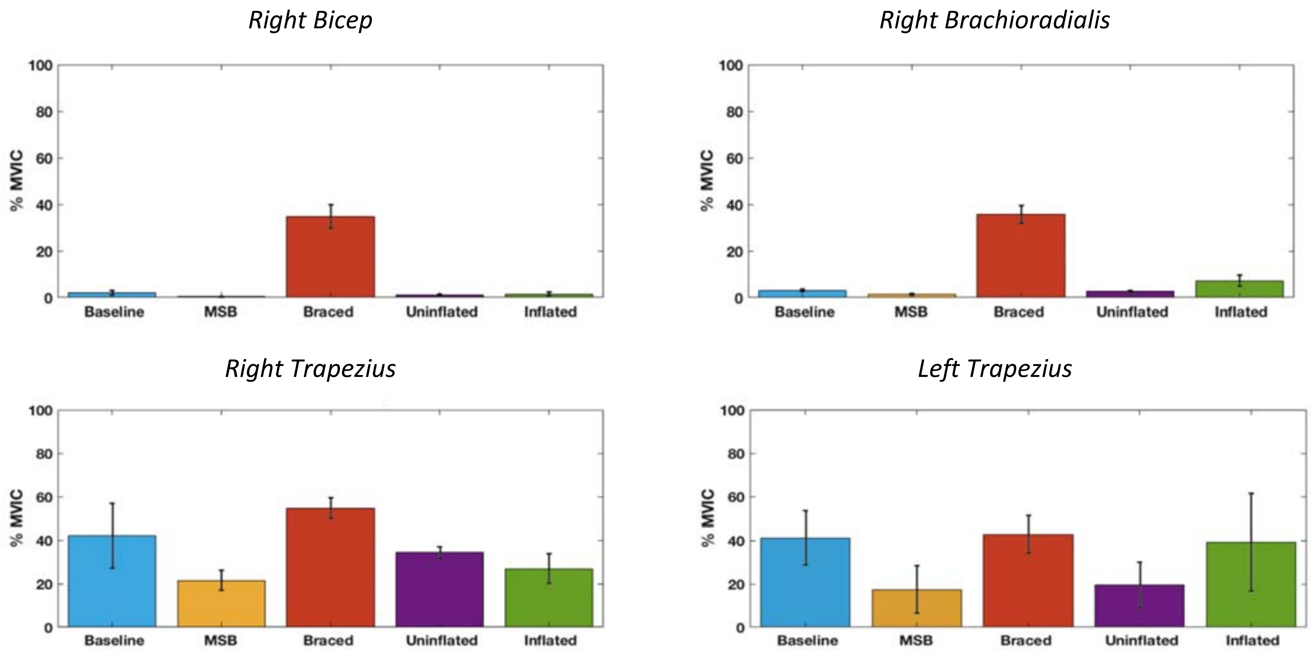


Fig. 9. Mean EMG response expressed as a percentage of MVIC for Bicep, Brachioradialis and Trapezius (L, R) muscles by condition (averaged over all cycles). Error bars represent the standard deviation.

IV. DISCUSSION

This study explored the lateral motion of human volunteers via a safe and repeatable testing device in a controlled laboratory setting. Unlike previous laboratory protocols, this study quantified the average baseline head and trunk displacement in a multi-swerving event where the change in acceleration direction altered kinematics and potentially influenced the effectiveness of countermeasures compared to a single-lane-change manoeuvre typical of previous studies. On average, across all conditions, the lateral movement of the head was 45% greater than the trunk. This may be largely attributed to shoulder belt restricting the motion of the trunk, whereas the head was not restrained and therefore had a greater range of motion. The Baseline condition resulted in the second largest displacement for the head (160.2 ± 52.7 mm) and the trunk (89.8 ± 34.0 mm). For some subjects, this was sufficient movement to cause shoulder-belt slip-out, which was detected in the motion capture data and a 2D video analysis.

The effect of a braced posture (a subject-induced countermeasure) was similar to that of the MSB (a vehicle-induced countermeasure) and both significantly reduced head and trunk displacement by approximately 45%, compared to the baseline case. In the bracing condition, reduced displacements were achieved through activation of the R. bicep, R. brachioradialis, and L, R trapezius muscles, where volunteers' mean EMG reached 40–50% of their MVIC. Although both the implementation of the MSB and a braced posture had similar results for displacement, the MSB is a passive intervention, requiring no action from the occupant. Bracing may be less likely for rear-seated passengers as they may have no visual warning of an impending crash, resulting in insufficient time for them to find and grasp a vehicle structure. In addition, a few subjects experienced shoulder-belt slip-out due to their braced posture as they positioned themselves out of the belt by pushing against the bracing structure. This suggests the pre-pretensioner may be an advantageous countermeasure during pre-crash emergency manoeuvres.

The second vehicle-based countermeasure was the sculpted seat, and it did not show similar benefits. In comparison to Baseline, there were no significant reductions in head and trunk displacement in either the Uninflated or Inflated condition. In fact, the Uninflated condition resulted in the greatest displacement for the head (165.1 ± 50.8 mm) and trunk (97.7 ± 24.9 mm). The inflated torso bolsters only reduced peak lateral head and trunk motion by 9% and 6% compared to Baseline, and by 11% and 14% compared to the Uninflated condition. For these size occupants, qualitative analysis of 2D videos demonstrated that their torso girth was wider than intended for the contours of the sculpted seat, thus the bolsters possibly shifted their initial position forward and away from the confines of the bolsters. It is important to note that 74% of the subjects were deemed to have an average BMI, according to the CDC.

The acceleration pulse, across all 490 trials, proved that the LATE device is repeatable and safe for oscillatory loading human volunteer tests. The first cycle produced a mean peak lateral acceleration of 0.53 ± 0.04 g, slightly lower than the subsequent cycles due to the device ramp up. Although this cycle had the lowest peak lateral acceleration, it resulted in the highest peak lateral displacement for both the head and trunk (Fig. 8). Due to subject unawareness of event onset, the maximum head and trunk displacement in the first cycle may be governed primarily by inertia and not include any voluntary muscle response to counteract movement. Phase differences were observed when comparing the head and trunk displacement to the sled motion where both the head and trunk lagged behind sled motion. Future analysis will determine whether the lag is affected by muscle activity across cycles. To the knowledge of the authors, this is the only study that has quantified human motion over several cycles of reversing acceleration in a laboratory setting. The peak lateral displacement for the head and trunk was significantly less in the latter cycles. Subsequent cycles may have given subjects the opportunity to employ neuromuscular strategies to resist motion. There is limited data to pinpoint the average number of swerves in a real-world evasive swerving scenario, yet it is likely that multiple swerve scenarios do exist. Multiple cycles may affect countermeasure performance. If a pre-pretensioner is deployed too late in a LATE event (e.g. during or after the first cycle), it might not efficiently reduce lateral occupant displacement [20]. Triggering while an occupant is displaced during a cycle may have an effect on injury during the crash phase, therefore it is vital to understand the kinematics that occur during several cycles of swerving in order to determine the proper phasing and magnitude of restraint deployment.

There were several limitations associated with this study. First, although the seating compartment was representative of a vehicle environment, it did not include specific vehicle interior structures, such as a door or instrument panel or front seat back (for rear-seat occupants), which may have influenced subject behaviour. A more realistic environment was balanced with the benefit of obtaining a clear visualisation of the subject's kinematics and to avoid providing a potential contact surface that may cause injury to the human volunteers. Second, error may arise in the placement of 3D motion capture markers. Volunteers were provided with a tightly fitted sleeveless shirt to allow easy palpation of anatomical landmarks. Error may exist in assuming the markers, which are placed directly onto the volunteer's skin, match precisely the movement of the skeletal structures they represent. Third, the subjects were limited to an all-male volunteer population, which is not a holistic representation of the entire vehicle population. Gender differences arise in neck flexibility, which has been observed in the passive cervical range of motion in male and female children and adults [21]. To limit variability in the results caused by gender, the subject population was restricted to healthy males. Lastly, the sled provided a purely lateral loading condition; no pitch, yaw or roll or forward momentum was provided to simplify the testing environment.

Future research should consider a more realistic vehicle dynamic environment and should also explore the effects of age on occupant kinematics, as similar data were collected on paediatric volunteers aged 9–17 years. Developmental differences that occur between adolescence and adulthood may cause age-specific kinematic responses in the pre-crash phase. Examining the effect of age will reveal how occupant kinematics are affected by size and neuromuscular development. It is anticipated that the implementation of the safety countermeasures studied herein will reduce lateral displacement independent of age, but the magnitude of reduced kinematics will vary across age groups. Further analysis of additional muscle groups is also necessary as muscle activity from the various muscle groups may be modulated with age.

V. CONCLUSION

The implementation of a pre-pretensioner or reversible motorised seat belt was an effective vehicle countermeasure during evasive swerving manoeuvres as it substantially reduced lateral head and trunk displacement. While a braced posture showed comparable results to the MSB, bracing may not be a practical strategy for unaware passengers during critical driving situations. The lateral support from the sculpted seat and inflatable torso bolsters were not a beneficial countermeasure for the adult size occupant as their torso geometry caused them to sit forward of the torso bolsters. Kinematics differences were observed across the subsequent cycles of the test; the first cycle, likely representing the most unaware occupant, experienced the largest lateral displacement, despite having the lowest lateral acceleration. This suggests that the occupant employs various neuromuscular strategies to counteract motion as he becomes more aware of the loading condition. These data provide critical human volunteer responses to guide the implementation of safety countermeasures relevant for the pre-crash phase.

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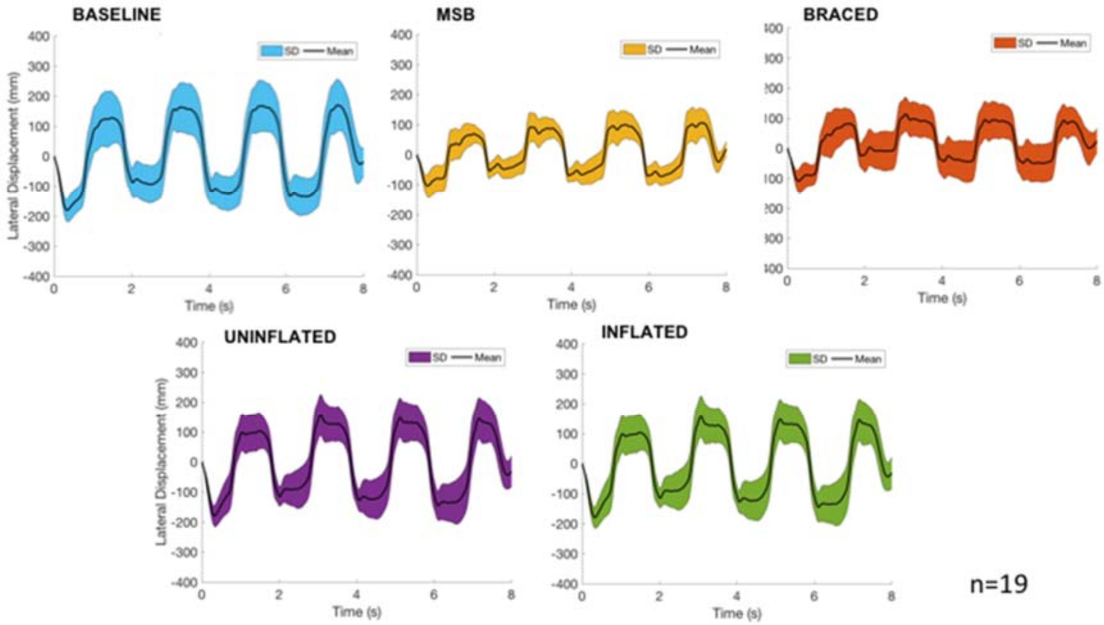
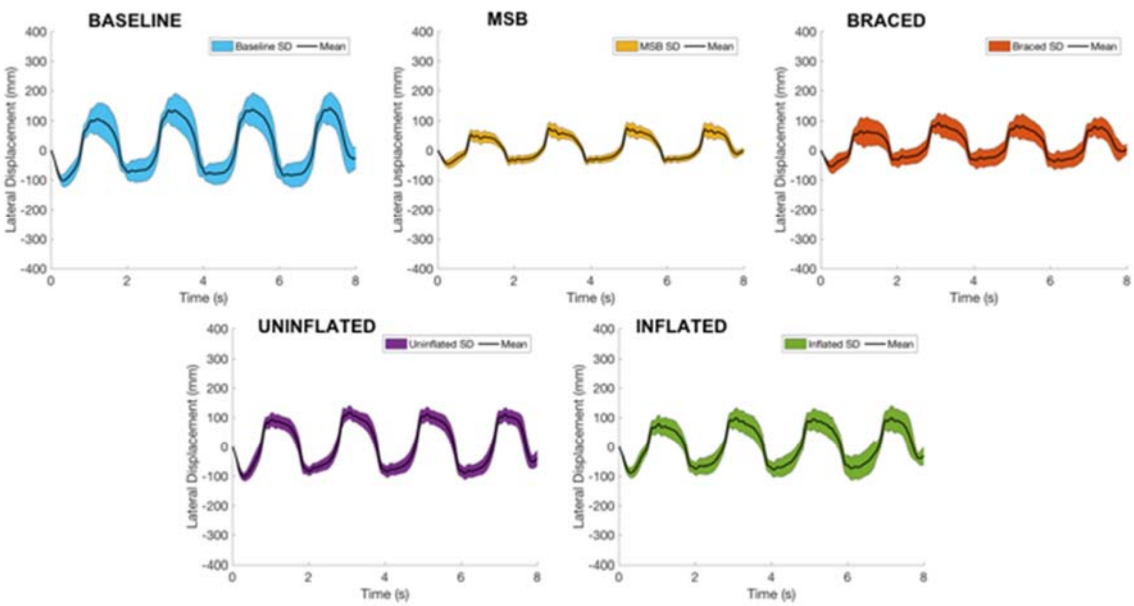
VIII. APPENDIX

Head Top Lateral Displacement Time Series



n=19

Trunk Lateral Displacement Time Series



n=19