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Human head and neck kinematics during autonomous and human braking in three initial head positions

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Abstract

Whiplash injuries resulting from vehicle collisions are still a significant socio-economic issue across the world. Years of research has resulted in the development of injury criteria, restraint systems and a deeper understanding of the injury mechanism. However, some grey areas remain and, in the context of the increasing automation of vehicles, one can wonder how the injury mechanisms may change due to changes in collision forces or directions. This paper presents an experiment with ten volunteers subjected to two braking modes, including automated braking preceded by an alarm warning or robot human braking, in three different initial head positions: forward facing, lateral rotation and flexion rotation. The volunteers were equipped with inertial measurement units to record their head and neck dynamics. Results show that the initial position of volunteers implies differences in the volunteer head dynamics. Also, the auditory alarm emitted prior to the emergency braking may have helped the volunteers to mitigate the mechanical stimulus and most likely the injury risk.

1. Introduction

Whiplash is a leading cause of injury in motor vehicle collisions and occurs frequently in rear end and frontal collisions [1]. It is caused by the sudden movement of the head forwards, backwards or sideways in the impact. This can result in symptoms such as soreness, neck muscle pain [2], and dizziness [3], or even more severe problems such as soft tissue lesions and tearing [4]. Whiplash injuries remain a significant socio-economic cost, according to the European Transport Safety Council the cost in Europe is 10 billion euros per year [5]. The injury mechanism, however, is still not fully understood [3], [6].

Various approaches have been proposed in the literature to better understand the human head-neck behaviour when submitted to external forces such as those during a vehicle collisions. Numerical analysis are proposed using mostly finite-element models to study the influence of geometrical and dimensional parameters on kinematics [7], [8], [9], but still lack validation data. Human volunteer experimental studies are used to explore the influence of cognitive parameters [10], [11] or to validate or test protection systems [1], [12]. Experiments are also carried out on dummies [13] or cadavers [14], [15]. When conducting experimental tests, the main protocol difference is usually the dynamic system used for the mechanical stimuli [1], [16] (mainly a car or a sled) and the choice of car crash dummies, cadavers or voluntary subjects. When car crash dummies are used, crash tests can be carried out at a high energy level and injury risk criteria computation is made possible using sensors fitted to the dummies. They have facilitated significant and invaluable progress in term of road safety by increasing the passive safety systems within vehicles. However, crash dummies will reach a limit in the advantages they provide because they are not validated for lower energy levels, represent only a narrow section of geometry variation, and cannot reflect the living component of human occupants such as muscular activity or specific behaviours influenced by cognitive factors. When volunteers are used, the level of energy is lower as volunteers must be protected from any risk of injury. The advantage of volunteers is that the dynamic neuromuscular response of each individual is accounted for [18]. However, injury criteria are then difficult to compute because of the nature of the parameters used as inputs (mostly precise mechanical strains). Thus, the experimental outcome is generally the volunteers' dynamic range of motion [19].

Many parameters have been identified as important factors for head stabilization. These parameters can be divided into passive and active, depending on the subject's behaviour. Alignment of the cervical spine [20], the dimensions of the head and neck, a subjects' relative size [7] and neck strength [21] have been identified as passive parameters. Anthropologic studies have exposed the variability in individuals in terms of size, weight, centre of gravity and the moment of inertia of various body segments [22]. These parameters explain the high inter-individual variability which can be observed in head stabilization among volunteers [19].

Active factors or cognitive factors are also responsible for a certain part of the human head/neck dynamic response to a sudden acceleration. Neck muscle activation, which is influenced by such cognitive factors, plays a key role in the dynamic response. Muscular tonus is, for example, higher among participants who have been exposed to stressful stimuli [23]. Part of the inter-individual differences observed can be explained by significatively different muscle activation timing [10]. A sudden loud sound prior to the acceleration event influences muscle activation and the dynamic response [10]. Awareness of the imminent acceleration also affects the dynamic response of subjects [11].

The automation of vehicles, from the addition of Advanced Driver-Assistance Systems (ADAS) (levels 1 and 2) to the fully driving automation (level 3 to 5) [24], is increasingly assigning the driving task to robotic systems. This leads to the car occupants' having to pay less attention to the road and the exterior environment and they are able to do other activities [25], [26]. This could mean the occupants are less likely to perceive an incoming emergency event and their head/neck dynamic response may differ, especially in the case of a whiplash-like loading where the preparation of people to the impact is a determining factor to the injury outcomes [11], [27]. Higher levels of automation in vehicles is likely to lead to a wider range of car occupants position [28]. Even though the automation of vehicles is supposed to make them safer, emergency events such as emergency braking or collision avoidance are still likely to happen, and the effectiveness of restraint systems may be reduced, since they are designed for in-position occupants and assessed using forward-facing crash-test dummies with both hands on the steering wheel [28]. As opposed to this precise seating position, all other positional set-ups will be referred as Out-Of-Positions (OOP). Little is known about the effects of OOP on the head and neck responses of vehicle occupants [29].

This study addresses the need for analysis of the kinematic motion of OOP occupants in vehicles undergoing emergency stops. The objective of this study was to quantify head dynamic motion response differences in passenger volunteers between human emergency braking and automatic emergency braking for one standard forward facing position and two OOP.

2. Materials and method

2.1. Vehicle

A passenger vehicle with an automatic transmission was used for the experiment. The vehicle was natively equipped with an automated emergency braking (AEB) system (automation level 1). This system was comprised of two cameras which detect frontal obstacles. In the case of an imminent collision, a repetitive alarm sound is emitted inside the passenger compartment. If the driver does not press the brake pedal, the alarm sound becomes louder and more urgent. Eventually, the brake pedal is automatically applied and the vehicle brought to a stop. The alarm sound is emitted approximately one second before the vehicle initiates its automatic braking.

The car was also fitted with a pedal robot system (see Figure 1) which was able to manipulate the accelerator and brake pedals. This system was used to produce a repeatable and reproducible acceleration (see 3.2), travel speed, and braking profile for the car. This system was used to imitate a human emergency braking, designated “Reproduced Human Braking” (RHB) in this study. The RHB deceleration profile was designed and programmed using real-world recordings provided by [30]. No specific alarm sound was emitted during the RHB protocol. The car was also fitted with electronics controlling the robot and recording output data about the vehicle’s dynamics (position, speed and acceleration of the vehicle and both actuators).

Figure 1: Pedal Robot

2.2. Test track and environment.

The test track was an empty section of an outdoor track with a level bitumen surface. The test track was approximately 60m in length. The start was at one end with approximately 15 meters used for accelerating the vehicle to the desired constant speed. Once the vehicle was at the desired constant speed, RHB braking could be applied as any time within the RHB zone in order to reduce the possible learning effect of the volunteers. At the other end of the track a pedestrian dummy was positioned (ref: 4activePS 50% adult male, 4activeSystems, Austria) which automatically triggered the AEB once the vehicle travelled beyond the RHB zone. While the pedestrian dummy was not used to trigger the RHB system, it was always in place for all trials so that the volunteers would be unaware of what braking mode will be applied. The vehicle returned to the start after each braking event. Figure 2 shows the setup of the test track.

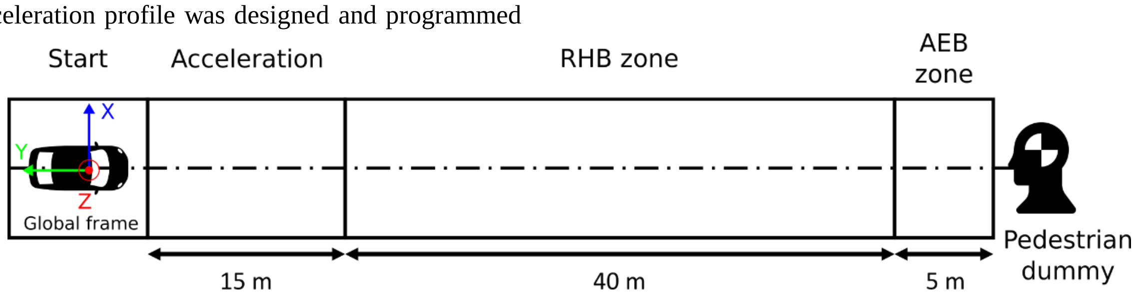

Figure 2: Schematic representation of the track. The RHB zone represents the area where the RHB can occur. The AEB zone represents the area where the AEB occurs.

2.3. Acquisition equipment

Subjects were equipped with two Inertial Measurement Units (IMUs) (MTx, Xsens, Enschede, the Netherlands) composed of three accelerometers, three gyroscopes and three magnetometers. The output data of interest were the 3-axis acceleration of the IMU in its own body-attached frame (see Figure 3) and the quaternions giving the orientation of the IMU in the global frame (3D rotation from the IMU-attached frame to the global one), with 100 Hz sampling. One IMU was placed on the top of the head (fixed on a headset) and the other on the T1 vertebra (see Figure 3). A third IMU was fixed rigidly to the centre console in order to record the acceleration of the vehicle. A fourth sensor was attached to the sacrum, but its data was not studied for this paper. Frames were realigned before every trial started using the IMU software utility.

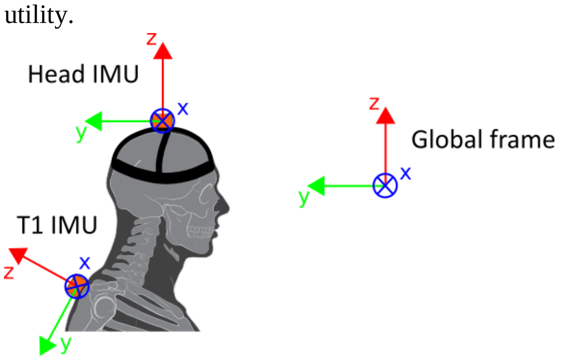

Figure 3: IMUs location (adapted from Patrick J. Lynch, C. Carl Jaffe, 2006)

2.4. Experimental conditions

The subjects were seated in the front passenger seat and did not have any control for the vehicle manoeuvre. In the vehicle with them were three investigators. A first investigator, seated at the driver seat, steered and piloted the vehicle and could override any robotic control in an emergency. Vehicle acceleration and braking were primarily robotically controlled. Two other investigators seated in the rear seats controlled the pedal robot system and the IMU equipment.

Each trial followed the same baseline sequence: instructions were given to the subject about how he/she should position him/herself on the seat. The vehicle pilot released the manual brake and the pedal robot accelerated the vehicle to the desired trial speed. When the speed was stabilised for a few seconds, the braking would occur.

Three position conditions (see Figure 4) were studied: a 'forward' position (defined as the standard position), a 'discussion' position and a 'phone' position. These positions were chosen to represent the standard position and two OOP selected in the literature [25], [26]. In the forward position, subjects were asked to look forward. In the discussion position, the subjects were asked to look towards the driver as if having a conversation. In the phone position, subjects were asked to look down towards their lap as if using a mobile phone or reading a book. The 'discussion' and 'phone' positions were measured and quantified for each participant, relative to the 'forward' position.

Figure 4: Position modalities: forward (left), discussion (middle) and phone (right)

Two speeds were used: 8 and 15 km/h. These speeds were achieved during the first few meters of the track. These speeds were low enough that subjects were not under risk of any kind of possible neck injury.

Two braking modes were used: either using the natively equipped AEB system or the RHB. The AEB system was triggered by the detection of a pedestrian dummy on the track (last 5 meters labelled as AEB zone on Figure 2). RHB was triggered by one operator in the rear seat at some point prior to the stereovision system detecting the dummy. The triggering of the RHB was randomized to occur at any point between when the vehicle speed stabilised and prior to AEB activation (labelled as RHB zone on Figure 2).

Twelve conditions (2 speeds x 2 braking modes x 3 positions) were tested (see Table 1). Each condition was tested 3 times, for a total of 36 trials per subject. The condition order was randomized during the testing. After volunteers were asked to position their head prior to each trial, no indication was given regarding the speed or the braking mode they were about to be submitted to.

Table 1: Test matrix of all conditions, each was randomly repeated 3 times.

Condition number	Speed	Position	Braking
1	8 km/h	Forward	RHB
2	8 km/h	Forward	AEB
3	8 km/h	Discussion	RHB
4	8 km/h	Discussion	AEB
5	8 km/h	Phone	RHB
6	8 km/h	Phone	AEB
7	15 km/h	Forward	RHB
8	15 km/h	Forward	AEB
9	15 km/h	Discussion	RHB
10	15 km/h	Discussion	AEB
11	15 km/h	Phone	RHB
12	15 km/h	Phone	AEB

2.5. Experimental data post-processing and statistical analysis

All data acquired during the experimental tests were processed with a Savitzky-Golay filter using a polynomial order of 3 and a window length of 21 samples [31]. The peaks were preserved and the signal was not distorted.

Data were post-processed using Python. Acceleration was computed in the global frame (with z-axis being aligned with the gravity) and the gravity-induced acceleration removed. Acceleration peaks were calculated in resultant. Rotation matrices were computed from the quaternion data output. Relative rotation matrices were computed through the matrix product. Euler angles were computed from the rotation matrix given the 'xyz' sequence. Relative head/T1 angle was thus computed as the relative head/T1 matrix. Range of Motion (ROM) is defined as the difference between the maximal and the minimal relative head/T1 angle during the braking and corresponds to the total angular displacement of the head relative to the T1 vertebrae. Data from each trial was synchronised with each other so that the maximum rate of change of acceleration (peak jerk) was coincident. Peak jerk approximately corresponded to when the vehicle had come to a complete stop and was stabilising on its suspension system. It will be referred as time = 0 in the figures.

For each dependant variable, a three-way repeated analysis of variance (ANOVA) was performed to assess differences related to the position (forward, discussion and phone), the speed (8 or 15 km/h) or the braking mode (AEB or RHB), with a 0.05 significance level.

3. Results

3.1. Subjects

After approval by the Adelaide University Human Research Ethics Committee (University of Adelaide – Ethics approval no - H-2018-241), ten healthy adult subjects volunteered to participate in this study (Table 2).

Volunteers were recruited thanks to an email campaign. Only volunteers who did not match our criteria were excluded. Exclusion criteria were: younger than 18 years old, had experienced a car crash during the past five years, had experienced a whiplash or head injury at any time in his/her life, currently had restricted head or neck movements, was pregnant, or currently under the influence of alcohol or drugs. All volunteers were male. Their height, weight and age (mean ± standard deviation SD) were respectively 180 ± 6 cm, 77 ± 4 kg and 37 ± 15 years. All subjects had a sedentary professional activity and little to no sporting activity. No subjects reported neck stiffness during the trials or at a two-week follow up.

Table 2: Subjects anthropometry

Subject number	Size (cm)	Weight (kg)	Age (years)
1	179	84	23
2	178	75	20
3	178	74	48
4	178	75	36
5	165	69	69
6	182	82	54
7	181	74	23
8	191	78	22
9	178	80	40
10	186	78	38

3.2. Vehicle braking behaviour

The AEB system was always automatically activated by the detection of the pedestrian dummy, as expected, and the vehicle came to a complete stop less than one metre before any impact. The RHB system always worked as expected for all the corresponding trials. Accelerations induced by the AEB and RHB at the two trial speeds were computed to compare the braking mechanical stimuli. Figure 5 shows the mean acceleration measured by IMU in the track axis (AEB vs RHB) at each speed.

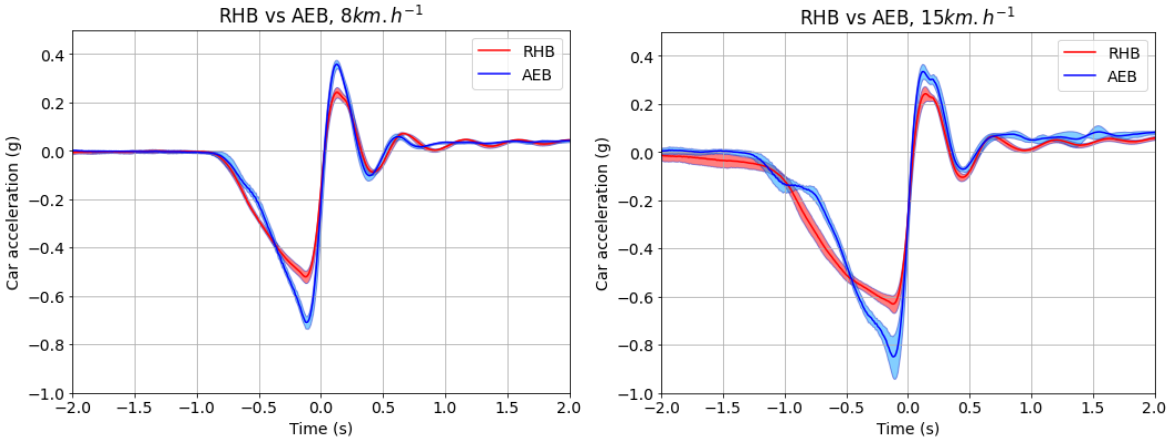

Figure 5: Mean car acceleration measured by IMU in the track axis for AEB (blue) and RHB (red) at given speed (8 km.h^{-1} at the left side, 15km.h^{-1} at the right side) and their respective corridor (±SD). The time 0 corresponds to the stop of the car.

Using the data recorded by the pedal robot system (which uses a dual GPS system), the braking time was computed using threshold detection. Table 3 shows the braking mean duration and the mean car deceleration peak for each trial speed and braking mode. AEB brought the vehicle to a stop in a shorter time compared with RHB for each speed. The performance differences were greater at the higher trial speed.

3.3. Head acceleration peaks

With the small number of volunteers, no significant difference was observed in the head acceleration due to height or age. No learning effect was observed in the head acceleration peaks. The three-way repeated ANOVA revealed that the head acceleration peak is significantly higher for the AEB compared to the RHB braking mode ($p<0.0001$).

Figure 6 shows an example of measured and processed accelerations of the car, head, T1and S1 of subject 1 in "Forward" position during AEB at 8 km.h^{-1}. The upper body acting like an inverse pendulum, it can be observe that the order or peaks are in a relevant order: car, S1, T1 and eventually the head.

Table 3: Mean braking duration (s) and mean car deceleration peak (g) for each braking mode and trial speed

Trial speed / Braking mode	8 km.h^{-1}	15 km.h^{-1}
Braking duration and car deceleration for RHB	0.56 s / - 0.5 g	0.80 s / - 0.61 g
Braking duration and car deceleration for AEB	0.52 s / - 0.72 g	0.61 s / - 0.86 g

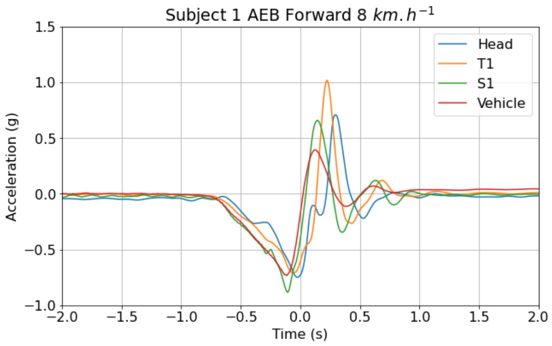

Figure 6: Accelerations of vehicle, head, T1 and S1 of subject 1 in "Forward" position during AEB at 8 km.h-1. The time 0 corresponds to the stop of the car.

Table 4 regroups the head acceleration peaks to be expressed as a ratio of head acceleration peak to the car deceleration peak. The results of the three-way repeated ANOVA on the ratio show that there were significant differences on this ratio computed among the different conditions. Significant differences were found for the braking modes (p=0.01). No significant difference was found for the different trial positions (p=0.33) and the trial speeds (p=0.10). No significant interaction effect was found between modes. Average head acceleration peaks were also computed. Average head acceleration peaks were higher for the AEB mode compared to RHB (p < 0.001).

Table 4: Mean ratio (± SD) of head deceleration to car deceleration peaks for each condition

	Trial position					
	Forward	**Discussion**	**Phone**	**Forward**	**Discussion**	**Phone**
Trial speed / Braking mode	**8 km.h^{-1}**			**15 km.h^{-1}**		
Head/car acceleration ratio for RHB	1.94 ± 0.32	1.68 ± 0.34	1.85 ± 0.33	1.62 ± 0.25	1.63 ± 0.35	1.80 ± 0.47
Head/car acceleration ratio for AEB	1.51 ± 0.15	1.61 ± 0.23	1.66 ± 0.22	1.61 ± 0.21	1.50 ± 0.22	1.51 ± 0.28

3.4. Head/T1 angles

With the small number of volunteers, no significant difference was observed in the head/T1 angle due to height or age. No learning effect was observed in the head/T1 angles. The maximal physiological flexion/extension ROM of all the volunteers were measured before the experiment and they presented an average (±SD) of 113.2° (± 32.3). For the ‘discussion position’, passengers turned their head to the driver's seat by an average (±SD) rotation of 43° (± 8.3). For the ‘phone position’ they angled their head down by 22° (± 4.5).

Figure 7 shows an example of the head/T1 angle during trials at 8 km.h^{-1} and illustrates various response patterns. Positive values refer to extension of the head while negative refer to its flexion. For visual representation, angles were zeroed 2s before the car acceleration peak.

The results of the three-way repeated ANOVA show that there were significant differences on the ROM measurements among the different conditions. Differences were found between the different trial positions (p=0.014). No significant difference was found for either the trial speeds (p=0.97) or for the braking modes (p=0.96). No significant interaction effect was found between conditions. Average head/T1 ROM (± SD) are shown in Table 5. ROM was found to be lowest for the ‘Forward’ position and highest for the ‘Phone’ position.

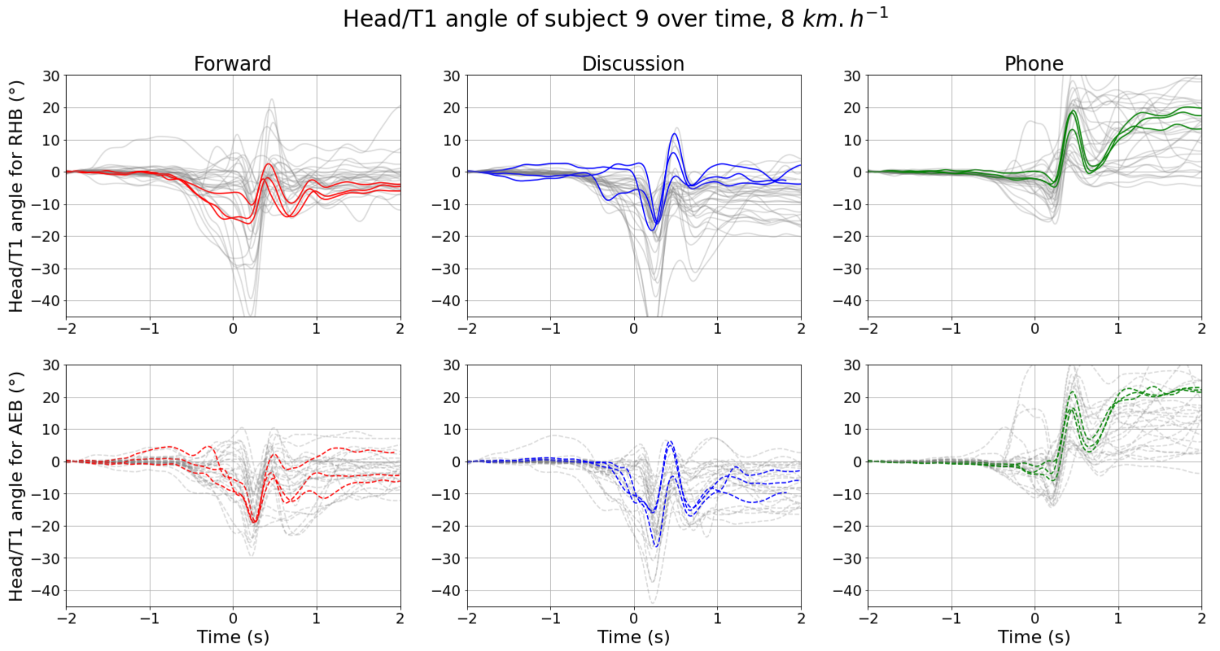

Figure 7: Head/T1 angles at 8 km.h^{-1}. Colored lines represent trials for subject 9 as an example, above all the 10 subjects' trials in grey, at 8 km.h^{-1}. The time 0 corresponds to the stop of the car.

Table 5: Head/T1 mean ROM (± SD) for each condition (°)

	Trial position					
	Forward	**Discussion**	**Phone**	**Forward**	**Discussion**	**Phone**
Trial speed / **Braking mode**	**8 km.h^{-1}**			**15 km.h^{-1}**		
Head/T1 ROM for RHB	21.1 ± 10.4 °	22.3 ± 10.1 °	26.1 ± 11.4 °	21.3 ± 11.3 °	22.3 ± 8.0 °	26.1 ± 10.0 °
Head/T1 ROM for AEB	20.8 ± 6.8 °	22.2 ± 7.4 °	25.5 ± 6.5 °	22.8 ± 8.2 °	20.8 ±6.0 °	27.2 ± 7.8 °

4. Discussion

The current study presents a total of 360 controlled braking events performed on 10 volunteers, aimed at investigating the head kinematics of vehicle occupants in response to two braking modes combined with three trial positions at two different trial speeds. The positions were chosen to reflect some recurrent situations: a standard position where the subjects were asked to look forward, an OOP where the subjects were asked to look to their right as if they were speaking with the driver, and a second OOP where subjects were asked to look down as if reading a mobile phone or book. The two initial trial speeds were tested in order to explore whether this had an effect on the outcomes. Finally, two braking modes were utilized: a natively equipped AEB system which was triggered by the detection of an obstacle and was preceded by an alarm sound approximately one second before the braking, as well as a robotic braking called Reproduced Human Braking (RHB) which was designed to emulate real driver behaviour in an emergency braking situation. No alarm sound was emitted before the RHB braking.

The AEB was found to be a more severe braking event with mean braking time and distance lower than that for the RHB. This was an expected result and is representative of what happens in the real world. This is also why the emergency brake assist (or brake assist system) was developed: drivers fail to brake with enough force when faced with an emergency. This can be observed in Figure 5Figure where acceleration inside the car measured higher for the AEB compared to the RHB. As no significant ROM differences were found between braking modes, it may indicate that the auditory signal from the AEB allowed the subjects to maintain a similar ROM output (and most likely a similar injury risk) despite the stronger braking [17]. The alarm sound emitted by the AEB system is not natively designed to produce a startle response or specific high intensity muscle contraction. It is a progressive warning to alert the driver that he/she has to act on the brake pedal before an automatic action will occur. No specific guidelines were provided to the volunteer passengers. When the alarm sound was emitted, all the volunteers kept their head position constant, depending on the current trial condition. No volunteer looked up at the road or shifted their head to the forward position. This behaviour might be explained by their confidence in the AEB system and the safe environment of the experiment, even while testing several emergency brakings. Moreover, this behaviour in "discussion" and "phone" positions could be compared to future studies focused on the possible lower awareness of incoming rear end collision events.

Plots of the head/T1 angle show that there is a different flexion/extension balance (i.e., the relative flexion/extension distribution) during the movement of subjects according to their initial position. In addition, statistical analysis shows that only the initial head positions had a significant effect on the ROM. The head/T1 ROM was lowest for the “Forward” condition and highest for the “Phone” position, with the “Discussion” position in between. Several hypotheses are proposed. In the OOP, muscles are not in their neutral position and so their ability to restrain the head may have been reduced [32]. In the "Phone" position, the head is already close to the maximal flexion prior to braking. Even if the whole upper body is moving forward due to the braking, the relative angle between head and torso is not able to increase to a greater flexion (Figure 7). The same circumstances could also occur in the "Discussion" position while the subject’s head is close to a maximal lateral rotation. Another explanation might result from the subject’s field of view. In the “Discussion” position, and particularly the “Phone” position, subjects have a restricted view of the track, so their knowledge about the incoming event would have been reduced. As reported by Kumar et al. [11], expectation of an incoming braking event reduces head movement by approximately 30% compared to an unexpected braking event. Thus, head movement of the subjects may have been increased by their reduced expectation of the braking event.

The differences found in terms of kinematics in the three simple head positions investigated in this study show the need to further investigate the effect of OOP on whiplash injury risk, especially in the context of the increasing automation of vehicles. Mathematical and mechanical models, as well as crash test dummies, should be developed or adapted to assess injury risk in various OOP as they become more common.

When considering the head acceleration peak, statistical analysis showed that only the braking mode had a significant influence on the results. Head acceleration peak was significantly higher in the case of AEB compared to RHB. This can be explained by the fact that the vehicle deceleration peak was higher in the AEB case, given that acceleration of the head is correlated with the acceleration of the car, as reported by Siegmund et al. [33]. Additionally, the ratio of head acceleration peak to car deceleration peak was lower in the case of AEB. As reported earlier, no significant effect from the braking mode was found on the ROM. This is a notable result as ROM changes were expected. According to Siegmund et al. [33], car acceleration also correlates positively with the head angle. The non-difference in terms of ROM, in light of the different levels of braking may be explained cognitively. Despite no specific indications being given to the subjects regarding speed and braking mode prior to each trial, they are likely to have sensed the acceleration of the car and perhaps deduced the car's speed. The two braking modes were somewhat expected by the subjects as it was clear the vehicle had to stop before the end of the track. The most unexpected was the RHB which occurred at a random time while the vehicle was into the RHB zone. The timing of AEB was clearer to the subjects as a motionless dummy, positioned at the end of the track, was used to trigger the automatic braking system. Subjects were thus aware of the AEB timing in the 'Forward' position as they could see the vehicle approaching the dummy prior to the triggering. Moreover, an auditory alarm was emitted approximately one second prior to the event. As such, the timing of AEB triggering was also deduced by the subjects when in the 'Discussion' and 'Phone' position, even if they could not necessarily see the exterior environment. Kumar et al. [34] reported that the earliest contraction of neck stabilization muscles (levator scapulae, sternocleidomastoid and trapezius) occurs before the head acceleration peak, and plays a significant role in the dynamic response. We can hypothesize that the alarm prior to AEB may have resulted in an earlier muscle reaction process which helped the subjects to mitigate the higher braking power of the AEB. This also could explain why the ration of head acceleration peak to car deceleration peak is lower in the AEB case. Unfortunately, the present experiment did not include electromyography (EMG) sensors and the muscle activity was therefore not acquired. Further investigation using EMG would be required to confirm this hypothesis. In addition to potentially triggering early muscle contractions, the alarm signal prior to AEB might have caused a sense of 'urgency' [35]. However, despite a lack of EMG recordings, the recorded videos did not reveal any evidence of a startle response from any of the subjects; mainly because it is not the objective of the AEB system alarm sound. Further work on the alarm design is suggested, especially with regard to whiplash injury risk mitigation. Indeed, some research teams have already shown correlations between alarm parameters (e.g. fundamental frequency, wave amplitude, or harmonics) and objective data (EEG, EMG) [36], [37].

The main limitations of this study are the lack of physiological data, particularly with EMG, and none of the conditions enabled a situation where the subject was completely surprised by the braking event. However, the results of this study suggest that the alarm sound specifically helped volunteers mitigating the emergency braking and further work can be considered. In particular, it would have been interesting to design a study in which a subject has to carry out an activity requiring a certain mental load while the braking takes place unexpectedly, particularly in a traffic situation [38].This could be a future objective by integrating virtual reality such that this type of scenario can then be performed and compared in a driving simulator.

5. Conclusion

One of the main challenges for the research on head and neck stabilization~~,~~ is the multidimensionality of the relevant parameters. Only ten subjects took part in the preliminary study reported here and the statistical power is not high, particularly as repeated measures had to be used to overcome the inter-individual differences observed while studying the effects of whiplash-like loading on volunteers. However, some results can be outlined. It seems that the level of braking, the presence of an auditory signal and the position of the subjects may have all influenced the output in some way. As reported in the introduction, the on-going automation of vehicles is likely to dramatically change accident characteristics [28], leading to, for example, more OOP. With the increasing automation of vehicles comes the rise of greater on-board computational resources. Some thoughts on future work can be given. It could be imagined that future automated braking could consider the position and cognition of passengers, thanks to cameras and artificial intelligence classification algorithms, and provide a more forgiving deceleration when necessary and permitted by the impending emergency. The position and cognition of the vehicle occupants could be considered as parameters for the applied braking curve. Models of the subject's dynamics and injury criteria could even be incorporated in a decisional algorithm which could help to find the balance between the power of the braking, protection of the passengers and the space/time available prior to an impact with another vehicle, pedestrian or other object.

References

[1] L. Jakobsson, B. Lundell, H. Norin, and I. Isaksson-Hellman, "WHIPS - Volvo's whiplash protection study," *Accid. Anal. Prev.*, vol. 32, no. 2, pp. 307–319, 2000.

[2] G. Bono, F. Antonaci, S. Ghirmai, F. D'Angelo, M. Berger, and G. Nappi, "Whiplash injuries: clinical picture and diagnostic work-up," *Clin Exp Rheumatol*, vol. 18, no. 2 Suppl 19, pp. S23-8, 2000.

[3] H. Chen, K. H. YANG, and Z. Wang, "Biomechanics of Whiplash injury," *Chinese J. Traumatol.*, vol. 12, no. 5, pp. 305–314, 2009.

[4] M. Curatolo, N. Bogduk, P. C. Ivancic, S. A. McLean, G. P. Siegmund, and B. A. Winkelstein, "The role of tissue damage in whiplash-associated disorders," *Spine (Phila. Pa. 1976).*, vol. 36, no. 25, pp. S309–S315, 2011.

[5] T. Janitzek, "Reining in Whiplash – Better Protection for Europe's Car Occupants," *ETSC - Eur. Transp. Saf. Counc.*, 2017.

[6] S. Laporte *et al.*, "An attempt of early detection of poor outcome after whiplash," *Front. Neurol.*, vol. 7, no. OCT, 2016.

[7] A. N. Vasavada, J. Danaraj, and G. P. Siegmund, "Head and neck anthropometry, vertebral geometry and neck strength in height-matched men and women," *J. Biomech.*, vol. 41, no. 1, pp. 114–121, 2008.

[8] B. Bazrgari, A. Shirazi-Adl, and M. Parnianpour, "Transient analysis of trunk response in sudden release loading using kinematics-driven finite element model," *Clin. Biomech.*, vol. 24, no. 4, pp. 341–347, 2009.

[9] P. Tropiano *et al.*, "Using a finite element model to evaluate human injuries application to the HUMOS model in whiplash situation," *Spine (Phila. Pa. 1976).*, vol. 29, no. 16, pp. 1709–1716, 2004.

[10] J. S. Blouin, J. T. Inglis, and G. P. Siegmund, "Auditory startle alters the response of human subjects exposed to a single whiplash-like perturbation," *Spine*, vol. 31, no. 2. pp. 146–154, 2006.

[11] S. Kumar, Y. Narayan, and T. Amell, "Role of awareness in head-neck acceleration in low velocity rear-end impacts," *Accid. Anal. Prev.*, vol. 32, no. 2, pp. 233–241, 2000.

[12] D. C. Viano and M. F. Gargan, "Headrest position during normal driving: Implication to neck injury risk in rear crashes," *Accid. Anal. Prev.*, vol. 28, no. 6, pp. 665–674, 1996.

[13] H. Yoshida and S. Tsutsumi, "Experimental analysis of a new flexible neck model for low-speed rear-end collisions," *Accid. Anal. Prev.*, vol. 33, no. 3, pp. 305–312, 2001.

[14] J. Cholewicki, M. M. Panjabi, K. Nibu, L. B. Babat, J. N. Grauer, and J. Dvorak, "Head kinematics during in vitro whiplash simulation," *Accid. Anal. Prev.*, vol. 30, no. 4, pp. 469–479, 1998.

[15] A. F. Tencer, P. Huber, and S. K. Mirza, "A comparison of biomechanical mechanisms of whiplash injury from rear impacts.," *Annu. Proc. Assoc. Adv. Automot. Med.*, vol. 47, pp. 383–98, 2003.

[16] S. Kumar, R. Ferrari, and Y. Narayan, "Effect of head rotation in whiplash-type rear impacts," *Spine J.*, vol. 5, no. 2, pp. 130–139, 2005.

[17] D. C. Viano and J. Davidsson, "Neck displacements of volunteers, BioRID P3 and Hybrid III in rear impacts: Implications to whiplash assessment by a Neck Displacement Criterion (NDC)," *Traffic Inj. Prev.*, vol. 3, no. 2, pp. 105–116, 2002.

[18] G. P. Siegmund, J.-S. Blouin, M. G. Carpenter, J. R. Brault, and J. T. Inglis, "Cervical Multifidus Muscle Activity during Whiplash and Startle Responses," pp. 133–144.

[19] N. Vibert *et al.*, "Psychophysiological correlates of the inter-individual variability of head movement control in seated humans," *Gait Posture*, vol. 23, no. 3, pp. 355–363, 2006.

[20] K. Ono, S. Ejima, Y. Suzuki, K. Kaneoka, M. Fukushima, and S. Ujihashi, "Prediction of Neck Injury Risk Based on the Analysis of Localized Cervical Vertebral Motion of Human Volunteers During Low-Speed Rear Impacts," *IRCOBI Conf. Proc.*, no. September, pp. 103–113, 2006.

[21] M. C. Hillier, "Head Acceleration And Associated Pain Felt In The Neck Region During a Simulated Automobile Low Velocity Rear-End Collision As a Function Of Seat Back Angle," University of Florida, 2008.

[22] J. T. McConville and T. D. Churchill, "Anthropometric Relationships of Body and Body Segment Moments of Inertia," 1981.

[23] R. L. Hazlett, D. R. McLeod, and R. Hoehn-Saric, "Muscle tension in generalized anxiety disorder:

Elevated muscle tonus or agitated movement?," *Psychophysiology*, vol. 31, pp. 189–195, 1994.
[24] SAE International, "Surface Vehicle," *SAE Int.*, vol. 4970, no. 724, pp. 1–5, 2018.
[25] M. Kyriakidis, R. Happee, and J. De Winter, "Public opinion on automated driving: Results of an international questionnaire among 5000 respondents," *Transp. Res. Part F Traffic Psychol. Behav.*, vol. 32, pp. 127–140, 2015.
[26] Z. Wadud and F. Y. Huda, " Fully automated vehicles: the use of travel time and its association with intention to use ," *Proc. Inst. Civ. Eng. - Transp.*, pp. 1–15, 2019.
[27] L. Barnsley, S. Lord, and N. Bogduk, "Clinical Review Whiplash injury," *Pain*, vol. 58, no. 3, pp. 283–307, 1994.
[28] D. Subit, P. Vezin, S. Laporte, and B. Sandoz, "Will automated driving technologies make today's effective restraint systems obsolete?," *Am. J. Public Health*, vol. 107, no. 10, pp. 1590–1592, 2017.
[29] D. C. Alpini, *Whiplash Injuries Diagnosis and Treatment*. 2014.
[30] B. Sandoz, K. Bucsuházy, A. van den Berg, J. Dutsche, and J. Mackenzie, "Acceleration of a car passenger during automatic emergency braking," 2018. World Congr Biomech.
[31] A. Savitzky and M. J. E. Golay, "Smoothing and Differentiation of Data by Simplified Least Squares Procedures.," *Anal. Chem.*, vol. 36, no. 8, pp. 1627–1639, May 1964.
[32] K. R. Kaufman, K. N. An, and E. Y. S. Chao, "Incorporation of muscle architecture into the muscle length-tension relationship," *J. Biomech.*, vol. 22, no. 8–9, pp. 943–948, 1989.
[33] G. P. Siegmund and J. S. Blouin, "Head and neck control varies with perturbation acceleration but not jerk: Implications for whiplash injuries," *J. Physiol.*, vol. 587, no. 8, pp. 1829–1842, 2009.
[34] S. Kumar, R. Ferrari, and Y. Narayan, "Kinematic and electromyographic response to whiplash loading in low-velocity whiplash impacts - A review," *Clin. Biomech.*, vol. 20, no. 4, pp. 343–356, 2005.
[35] J. Singer, N. Lerner, D. Kellman, and E. Robinson, "Crash Warning Interface Metrics : Warning and Message Perception Under Ambient Noise Conditions Laboratory Experiments," no. November, 2015.
[36] J. Edworthy, "The design and implementation of non-verbal auditory warnings," *Appl. Ergon.*, vol. 25, no. 4, pp. 202–210, 1994.
[37] J. L. Burt, D. S. Bartolome, D. W. Burdette, and J. R. Comstock, "A psychophysiological evaluation of the perceived urgency of auditory warning signals," *Ergonomics*, vol. 38, no. 11, pp. 2327–2340, 1995.
[38] G. Andersson, J. Hagman, R. Talianzadeh, A. Svedberg, and H. C. Larsen, "Effect of cognitive load on postural control," *Brain Res. Bull.*, vol. 58, no. 1, pp. 135–139, 2002.