Occupant Kinematics and Biomechanics with Rotatable Seat in Autonomous Vehicle Collision: A Preliminary Concept and Strategy

Xin Jin, Haibin Hou, Ming Shen, Hequan Wu, King H. Yang

Abstract In a highly autonomous vehicle (HAV), the rotatable seat is likely to be designed to facilitate ease of communication between the occupants. We hypothesise that the protective effects of current restraint systems vary among different seating configurations and that by using the rotational seat to alter the occupant's orientation in accordance with the direction of impact, occupants will be better protected. Moreover, in a HAV it's likely that an imminent impact could be detected at a time of 200 ms, or even longer, prior to the initial contact. The availability of this additional time could be used strategically to actively position the occupants into a safer position for impact.

Finite element simulations were performed using the THUMS^{TM} model to test the hypothesis. The simulation results indicated that during a frontal impact, the backward-facing occupant is safer than occupants in other seating orientations. Moreover, 200 ms is sufficient to rotate the occupant by ±45^{o} without introducing additional injuries. Further studies are needed to optimise the rotating seat parameters in order to maintain occupant posture and improve crash safety in HAVs.

Keywords Highly autonomous vehicles, Seating orientation, Finite element method, Occupant kinematics, Injury risk.

I. INTRODUCTION

In the coming era of highly autonomous vehicles (HVAs), drivers will be released from driving tasks and no longer required to sit behind the steering wheel. Occupants will thus expect to relaxation and converse with their passengers, which will open up the need for new spatial configurations. The design concepts for HVA seating and interior design include flexible seat positioning and orientations, and fully reclining seats [1]. Jorlöv *et al.* [2] performed a volunteer study to identify the most commonly selected seating configurations during a long drive scenario. It was found that the most preferred seating configuration was rotating the front seats 90–180°, so that all occupants can face each other.

From safety perspectives, these new seating positions may raise some concerns. Conventional occupant restraint systems are designed for the upright, forward-facing position and their effective performance is generally sensitive to the initial posture of the occupant. Altering the seat recline angle, orientation, or even location will greatly affect the performance of the restraint system. Kitagawa *et al.* [3] studied the influence of seating position and orientation on occupant kinematics using THUMS^{TM} model. The results indicated that the combination of seatbelt and seat was essential for the control of occupant kinematics. However, injury risk evaluation was not equally emphasised in that study. It is likely that the conventional seatbelt and seat system cannot sufficiently protect occupants in certain seating configurations. In other words, the protective effects of the restraint system may vary under different seating configurations in a given impact scenario. As a result, it can be hypothesised that if the occupant can be actively and strategically "moved" to a safer seating configuration, then the occupant will be better protected.

In conventional passive safety concepts, this hypothesis is not true because there is insufficient time to translate or rotate the seat after an impact is detected. In a HAV, however, the camera and radar sensor system is likely to detect an imminent impact before it actually happens. In a typical active safety system [4], a forward

X. Jin (e-mail: xin.jin@wayne.edu: tel: +1 313 577 0422) is a Research Assistant Professor and M. Shen is a Graduate Research Assistant in Biomedical Engineering at Wayne State University, USA. K. H. Yang is a Professor and Director at the Bioengineering Center at Wayne State University. H. Hou and H. Wu are Visiting Scholars from the Department of Mechanical Engineering at Changsha University of Science and Technology, China.

collision warning system can recognise an impending impact and warn the driver to take action about 1.5 s before the collision happens. If no action is taken by the driver, the autonomous precrash braking will be activated about 0.45 s before the crash. In addition, an active safety system can recognise the inevitable collision states (ICS) about 0.1–0.35 second before collision at a variety of impact velocities and angles [5].

There are two objectives in this study. The first is to evaluate and compare the occupant injury risks in different seating orientations in a frontal crash. The second is to apply seat rotation to move the seat and occupant within the pre-collision timeframe. The load applied to the occupant and the corresponding injury risk will be assessed to examine if this active seat rotation is biomechanically feasible.

II. METHODS

The interaction between the occupant and the restraint system was simulated using a simplified vehicle cabin and a human body finite element (FE) model (THUMS™, Version 4.0, AM50 occupant model). Simulations were performed using LS-DYNA™ (Version 971, Livermore Software Technology Corporation).

Frontal Collision Simulation Model

In the first part of this study, a frontal vehicle collision with a delta-V of 56 km/h was simulated. The deceleration pulse was defined as input and applied to the vehicle body (Fig. 1). A single-seat model was constructed. The steering wheel and airbags were not included and the occupant was only restrained by a 3-point seatbelt. This setup ensured that the occupant kinematics and injury outcomes were fully controlled by the seatbelt during the simulated collision. The force limit of the seatbelt retractor was set to be 4 kN, and a 2 kN limit was set for pretension. The seat model represented the geometry of a general front seat, with rigid seat frames and deformable foam covers. Rigid seat structures were fixed to the rigid floor. Seat angles were set to a normal driving position. The lower legs of the THUMS model were slightly pulled back from the original driving posture. Figure 1 shows the model setup of the frontal collision simulation. Four initial seating orientations – 0°, 90°, 135° and 180° – were used, to simulate the driver-preferred orientations reported by [2], as shown in Fig. 2.

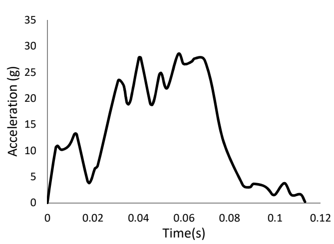


Fig. 1. Deceleration pulse.

Fig. 2. Frontal collision model setup.

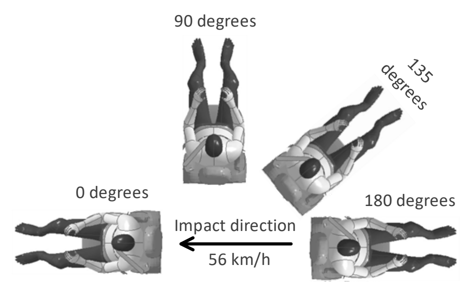


Fig. 3. Definition of initial seating orientations.

Seat Rotation Simulation Model

In the second part of this study, pre-collision seat rotation was simulated. The aimed rotation angle was set at ±45° and ±90°, as shown in Fig. 4. The rotation center was defined as the center of the seat frame structure under the cushion. The rotation velocity was defined as a sine curve with 200 ms duration (Fig. 5). Simulations terminated at 400 ms. During the first 100 ms, the seatbelt was allowed to pretension and fasten the occupant upper torso to the seat. From 100 to 300 ms, seat rotated to the desired angle. The last 100 ms allowed the lagged human body to catch up. 200 ms was chosen as the rotation duration as the current active safety system can provide a maximum of 0.35 s of pre-collision time before crash [5]. To assist seat rotation, a baffle structure was created, covering thigh, lower leg and foot, and attached to the rigid seat frame. A foam layer with the same material properties as the seatback and cushion was created to cover the baffle. The model setup of the seat rotation simulation is shown in Fig. 6.

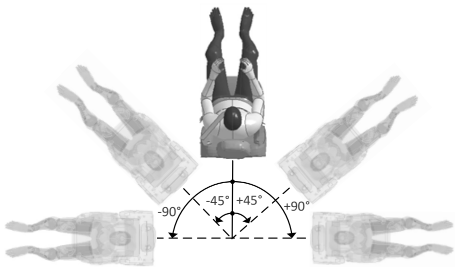


Fig. 4. Seat rotation angle definition.

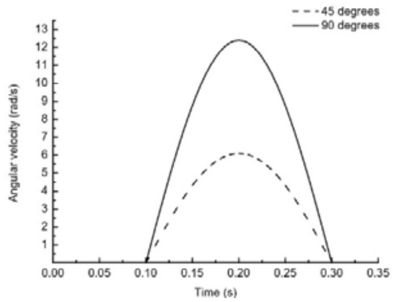
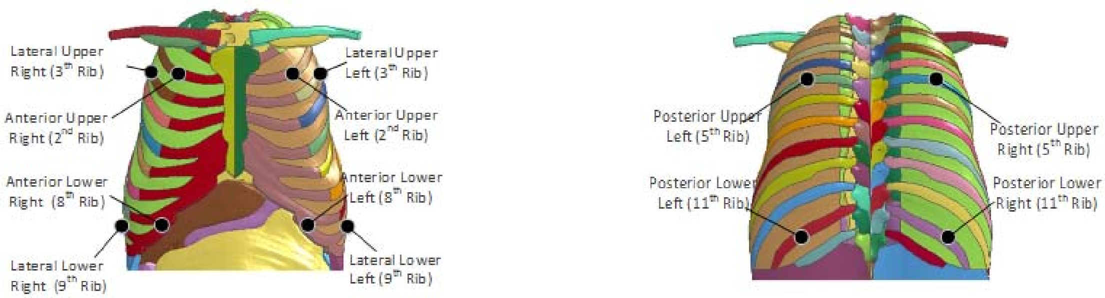


Fig. 5. Rotation velocity vs time.

Fig. 6. Seat rotation model setup.

Post-processing of Simulation Results

For both frontal collision and seat rotation simulations, nodal time history data at head CG location was output to calculate brain injury criteria (BrIC). A cross-section was created at C7. The output of neck axial loading and bending moment was used to calculate normalised neck injury criteria (N_{ij}). Anterior-posterior and lateral chest deflections were measured at multiple points on the ribcage, as reported by [6] (Fig.7). Maximum principal strain values on the anterior longitudinal ligaments (ALL), posterior longitudinal ligaments (PLL), capsular ligaments (CL), ligament flavum (LF) and interspinous ligaments (IL) were recorded and compared with the cervical spine ligament injury threshold reported by Yoganandan *et al.* [7]. Rib fractures and stress distribution patterns on the ribcage were observed.



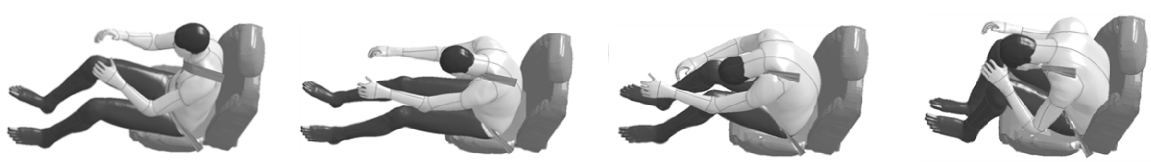
(a) Anterior view (b) Posterior view

Fig. 7. Measurement of chest deflection along anterior-posterior and lateral directions.

III. RESULTS

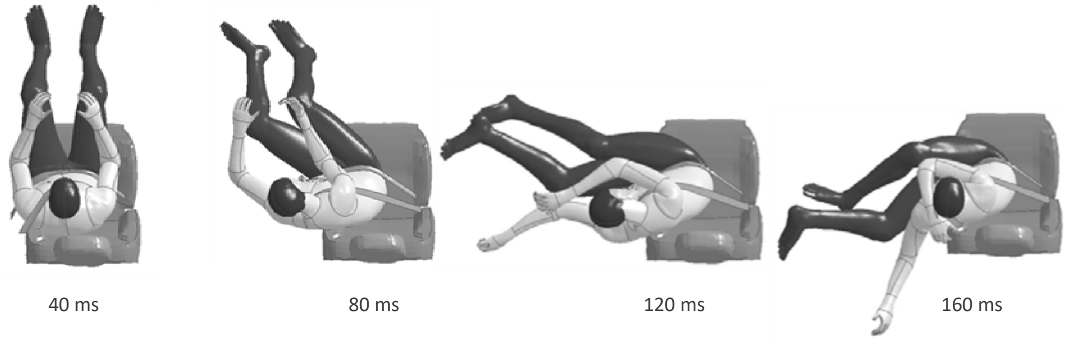
Frontal Collision Simulation

Figure 8 shows the occupant motion sequence at 40 ms intervals. The 0^{o} seating orientation indicates a forward-facing position. The shoulder belt restrained ribcage to top upper torso from moving forward. In the 90^{o} and 135^{o} configurations, although the upper torso was successfully stopped by the seatbelt, the restraint was applied through the neck engaging with the belt. The 180^{o} seating orientation indicates a rear-facing position. In this configuration, occupant was stopped by the seatback while minimum head rotation was observed.



40 ms 80 ms 120 ms 160 ms

(a) 0 degrees.



(b) 90 degrees.

(c) 135 degrees.

(d) 180 degrees.

Fig. 8. Occupant kinematics in frontal collision in different orientations.

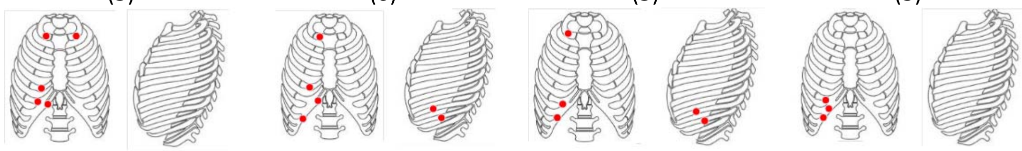
Table I summarises the injury criteria from simulation output. The seating configuration at 180^o shows the least injury criteria values among all four seating orientations. Although the predicted maximum chest deflection is greater at certain locations, the 180^o orientation has the least number of rib fractures. A similar finding was obtained from the ligament stretching results (Table II).

TABLE I

INJURY CRITERIA VALUES CALCULATED FOR FOUR SEATING ORIENTATIONS

Orientation (degree)		**0**	**90**	**135**	**180**
BrIC		0.90	1.15	1.57	0.62
N_{ij}		0.242	0.263	0.342	0.306
Chest Deflection* (mm)	Upper Left	-12.88	-24.85	-17.55	-16.97
	Upper Right	-16.04	-24.58	-14.82	-18.05
	Lower Left	-20.96	-60.44	-34.13	-44.31
	Lower Right	-32.34	-14.81	-14.03	-34.67
	Upper Lateral	-7.75	-8.84	-5.95	+5.03
	Lower Lateral	+20.43	+26.42	-28.82	+37.39
Rib Fractures		R1, R5-7, L1	L9-10, R1, R6-8	L9-10, R1, R7-8	R6-8

	(5)	(6)	(5)	(3)



*: Polarity indicates an increased (+) or decreased (-) distance between the two measurement points.

TABLE II
SUMMARY OF NECK LIGAMENT STRETCHING DURING FRONTAL COLLISION

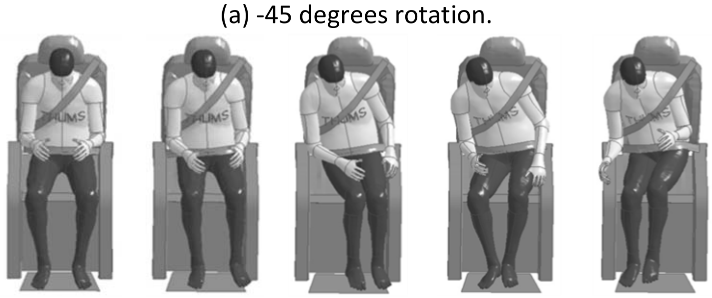
Ligament (C2-C7)	Injury Threshold (Yoganandan, 2000 [7])	0°	90°	135°	180°
ALL	0.35	0.12	0.33	0.43	0.37
PLL	0.34	0.25	0.66	0.12	0.31
CL	1.48	0.87	1.73	2.61	0.77
LF	0.88	0.21	0.36	0.44	0.35
IL	0.68	1.38	1.60	0.42	0.44

Seat Rotation Simulation

Figure 9 shows the seat rotation sequence at 80 ms interval. When the seat rotation was at negative angles (-45^{o} and -90^{o}), occupant's neck engaged with the shoulder belt. When the seat rotation was at positive angles (45^{o} and 90^{o}), the shoulder slipped off while the upper arm was still engaged with the seatbelt. The posture of lower extremities was well maintained at all rotation cases with the help of the baffle structure.

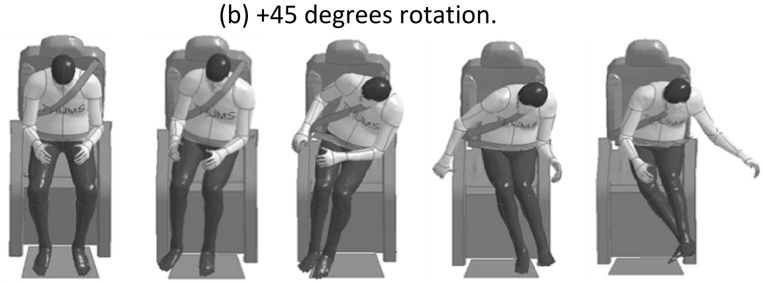
80 ms 160 ms 240 ms 320 ms 400 ms

(a) -45 degrees rotation.



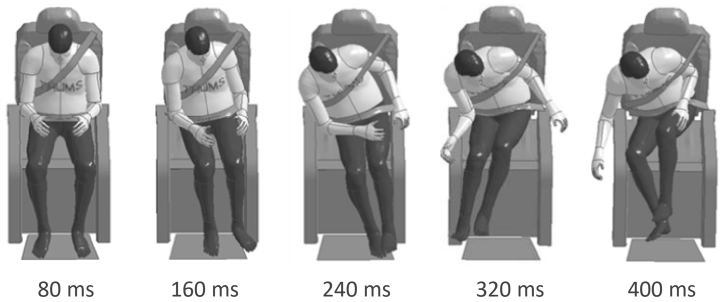
80 ms 160 ms 240 ms 320 ms 400 ms

(b) +45 degrees rotation.



80 ms 160 ms 240 ms 320 ms 400 ms

(c) -90 degrees rotation.



(d) +90 degrees rotation.

Fig. 9. Occupant kinematics in seat rotation at different angular displacement and velocities.

Table III summarised the injury criteria from seat rotation simulation output. The only rib fracture was found on R1 during -90^o rotation. In general, the higher the rotation velocity, the higher the injury criteria predicted from the simulations. The ligament stretching (Table II) also indicated an increased neck loading during the ±90^o seat rotation. During ±45^o seat rotation, the predicted injury criteria were less than the injury thresholds at different body regions. Thus, a 45^o seat rotation within 200 ms is biomechanically feasible.

TABLE III

INJURY CRITERIA VALUES CALCULATED FOR FOUR SEAT ROTATION SIMULATIONS

Seat Rotation (degree)		-45	-90	+45	+90
BrIC		0.10	0.25	0.14	0.22
N_{ij}		0.087	0.248	0.080	0.140
Chest Deflection* (mm)	Upper Left	-5.55	-11.66	-6.29	-6.50
	Upper Right	-10.09	-14.21	-3.01	-2.94
	Lower Left	-4.39	-7.86	-5.77	-17.95
	Lower Right	-5.94	-16.83	-6.97	-21.60
	Upper Lateral	+1.41	+1.67	+1.66	+3.63
	Lower Lateral	+2.91	+6.46	+2.99	+11.80
Rib Fractures		None	R1 (1)	None	None

*: Polarity indicates an increased (+) or decreased (-) distance between the two measurement points.

TABLE IV

SUMMARY OF NECK LIGAMENT STRETCHING DURING SEAT ROTATION

Ligament (C2-C7)	Injury Threshold (Yoganandan, 2000 [7])	-45 degrees	-90 degrees	+45 degrees	+90 degrees
ALL	0.35	0.12	0.34	0.12	0.26
PLL	0.34	0.24	0.60	0.13	0.25
CL	1.48	1.38	2.71	0.93	2.24
LF	0.88	0.21	0.06	0.17	0.27
IL	0.68	0.66	1.45	0.29	0.54

IV. DISCUSSION

In the first part of this study, we found the 180^o seating configuration was the safest among the four selected seating orientations. This finding is easy to explain in that during a frontal collision, the rear-facing position benefits the chest region by distributing the impact loading on a large area of seatback. Headrest helps to limit head rotation and thus decrease the BrIC and neck loads. On the other hand, 90^o and 135^o configurations have shown higher injury risks in the head and neck. This is because the current 3-point seatbelt design is based on forward-facing occupants. The excessive neck load due to the neck-seatbelt engagement will not happen in a conventional seating configuration. This type of seating orientation should be avoided during vehicle collision. As a result, the first hypothesis of this study has been tested: that the protective effects of the current restraint systems vary for different seating configurations. By simply “switching” the seating orientation, we can maximise the protection potential of the restraint system.

Multiple rib fractures were commonly observed in the lower right region (R6-8) while the maximum anterior-posterior chest deflection was mostly observed in the lower left. From the rib cage deformation sequence, we found that the maximum chest deflection happened near the end of the simulation (200 ms). However, the R6-8 fractured at about 60 ms. These fractures undermined the integrity of the rib cage and were followed with a large deformation on the left half of rib cage. But this deformation was distributed to the whole rib cage structure and no additional fracture was caused by it.

In the second part of this study, seat rotation was simulated within 300 ms pre-collision time (100 ms pretension followed by 200 ms rotation). This amount of pre-collision time is available with the aid of active safety systems. In the ±45^o rotation cases, all the predicted injury criteria were well below the injury threshold. In the ±90^o rotation cases, an increased neck ligament stretching and one rib fracture were observed, indicating that the rotation velocity was too high. Future studies should seek to establish the biomechanical limit under such seat rotation loading. To the best of authors’ knowledge, this study is the first to propose using a rotating seat to change the impact direction and improve crash safety for HAVs.

The current study only examined the biomechanical feasibility of the concept, which is, with a certain rotational velocity, a rotating seat will not increase injury risks. But from engineering perspectives, it is not a well-developed solution. Quite a lot of engineering questions, such as the mechanism of rotation actuator, space available inside the car to carry out the maneuver, the pre-collision time determination, etc., need to be studied and resolved before applying this strategy in HAVs.

Kitagawa’s study [3] indicated that the seatbelt and seat were essential to control occupant kinematics. In this study, it was found that in some seating configurations (e.g. 90^o and 135^o) excessive excursion of upper and lower extremities was observed, indicating that the seatbelt alone is not sufficient protection in these varying seating configurations.

Limitations

This study has the following limitations. First, the occupant model under the frontal collision condition with the four initial seating orientations was not quantitatively validated against experimental data. Although the biofidelity of the THUMS^{TM} model has been examined in a verity of literature, the performance of the 3-point seatbelt used in the current study was not carefully tuned to match any existing product. Secondly, the number of selected seating configurations is small. Only four seating orientations were simulated. It’s very likely there is an even safer orientation with the corresponding restraint system. However, the scope of this study is to demonstrate that in a given system, some seating configurations are relatively safer than the others. The goal of the proposed active safety system is to identify the protective effects of different seating configurations and integrate that information into the active safety strategy. Thirdly, the seat structure was assumed to be rigid in the simulations. Deformable seat structures can help to absorb more energy and thereby alter occupant kinematics and injury outcomes. In addition, the effects of occupant posture were not discussed in this study. The current restraint system is generally very sensitive to occupant posture. The same will hold true for HAVs, so a system that can detect occupant posture will be a crucial feature in designing safe HAVs.

V. CONCLUSIONS

This study discussed the concept of an active seat rotation strategy that changes seat orientation during the pre-collision timeframe. This strategy consists of two parts: identifying the relatively safer seating orientation; and rotating the seat to this direction. Four selected seating orientations, 0^o, 90^o, 135^o and 180^o, were studied during a frontal collision. Simulation results indicate that the 180^o seating configuration had the least injury risk among the four tested. In the second part of the study, the seat was rotated within 200 ms of pre-collision time.

Two rotation levels were selected: 45° and 90°. Simulation results showed that it was safe to rotate the seat 45° within the pre-collision time. There was an increased injury risk when increasing rotational velocity.

VI. REFERENCES

[1] Adient. (2018) "Adient's AI18 concept vehicle showcases the future of automotive seating". Internet: https://www.prnewswire.com/news-releases/adients-ai18-concept-vehicle-showcases-the-future-of-automotive-seating-300582208.html. Accessed on 28 March 2018.

[2] Jorlöv, S., Bohman, K., Larsson, A. (2017) Seating Positions and Activities in Highly Automated Cars – A Qualitative Study of Future Automated Driving Scenarios. *Proceedings of IRCOBI Conference*, 2017, Belgium.

[3] Kitagawa, Y., Hayashi, S., Yamada, K., Gotoh, M. (2017) Occupant Kinematics in Simulated Autonomous Driving Vehicle Collisions: Influence of Seating Position, Direction and Angle. *Stapp Car Crash Journal,* **61**: pp.101–55.

[4] Kusano, K. D., Gabler, H. C. (2012) Safety benefits of forward collision warning brake assist and autonomous braking systems in rear-end collisions. *IEEE Trans. Intell. Transp. Syst.,* **13**(4): pp. 1546–55.

[5] Savino, G., Giovannini, F., Fitzharris, M., Pierini, M. (2016) Inevitable Collision States for Motorcycle-to-Car Collision Scenarios. *IEEE Trans. Intell. Transp. Syst.,* **17**: pp.2563–73.

[6] Kitagawa, Y., Yasuki, T. (2013) Correlation among Seatbelt Load, Chest Deflection, Rib Fracture and Internal Organ Strain in Frontal Collisions with Human Body Finite Element Models. *Proceedings of IRCOBI Conference*, pp. 282–316, 2013, Gothenburg, Sweden.

[7] Yoganandan, N., Kumaresan, S., Pintar, F. A. (2000) Geometric and Mechanical Properties of Human Cervical Spine Ligaments. *ASME. Journal of Biomechanical Engineering,* **122**(6): pp.623–9.