

## WHIPLASH INJURY MECHANISMS OF CAR REAR OCCUPANTS: A REVIEW

Akmal Nur Haniffah<sup>1, 2\*</sup>, Siti Zawiah Dawal<sup>1</sup> and Sabariah Julaihi<sup>1</sup>

<sup>1</sup>Department of Mechanical Engineering, Faculty of Engineering, University of Malaya, 50603 Kuala Lumpur, Malaysia.

<sup>2</sup>Department of Mechanical Engineering, Faculty of Engineering, Universiti Pertahanan Nasional Malaysia, 57000 Kuala Lumpur, Malaysia.

\* Corresponding author: Akmal Nur Haniffah

Email: [akmal\\_sonic@yahoo.com](mailto:akmal_sonic@yahoo.com)

### ABSTRACT

*Whiplash injury due to low severity vehicles crash is a global problem. The injury has long-term clinical and biomechanical implications. Since the mid-1960s, injury statistics have continuously revealed that females face a higher risk of suffering the injury category compare to males. Besides, in a frontal crash, the injury measures from the adult rear dummies were mainly higher than the same size dummies located in driver and front occupant seat. However, most regulations and user crash tests have focused on vehicle drivers and front-seat passenger due to high occupancy and mortality rates in the front seat. In this paper, mechanisms of whiplash injury were reviewed to contribute a further inclusive understanding of human impact reaction, variability quantification, validation, and prevention. The objective of this study is to develop a new design of head restraint (HR) for car rear occupants. In order to raise consideration whiplash injury and prevention mechanisms, impacts are simulated with computer modelling (Ls-Dyna simulation) and validated using Matlab. Therefore, a review of these injury mechanisms indicates the development of new anti-whiplash technology in the automotive safety area is necessary.*

**Keywords:** Whiplash; impact; head; neck; head restraint (HR)

### INTRODUCTION

Whiplash Associated Disorders (WADs), which related to whiplash injuries, are the most common category of automobile injuries<sup>1</sup> for car rear occupants. Whiplash injury is an acceleration-deceleration mechanism of energy transmitted to the neck<sup>2</sup>. It can rise from a front or rear-end impact motor vehicle accident but can also happen while diving or other mishaps. The impact can occur in bony or soft-tissue injuries, which can lead to a sort of clinical manifestations. Besides, whiplash injuries resulting from low-velocity automotive impacts (typically <25 km/h)<sup>3</sup>.

As of 2018, whiplash injuries are estimated to affect 0.3% of people a year in the US sustained during sporting injuries, falls, and most commonly, motor vehicle accidents<sup>4</sup>. While estimation made for the European Union (EU) indicates that 300,000 EU residents undergo whiplash injuries yearly with 15,000 results in lasting sustaining and a combined socioeconomic impact of nearly EUR 10 billion every year, besides, WADs arise from injuries with structures of the soft tissue neck which cause most of the lasting disability due to vehicle impacts, i.e. up to 70% in Sweden<sup>5</sup>.

The majority sufferers facing beginning neck sign of illness, heal in several weeks or months of the accident; but, 5-10% of persons suffer various medical levels classified permanent disabilities. Whiplash injuries continue to be a significant neck pain cause and disability in the general population<sup>6</sup>. In specific, the cervical spine S-

shaped deformation at the time of rear collision is recognised as necessary for the mechanism affecting WADs<sup>7</sup>. Despite considerable research effort, the incidence of acute injury<sup>8</sup> and the rate of individuals transitioning to chronic pain symptoms have not remarkably changed over the past 30 years<sup>9</sup>.

Despite most of the whiplash signs are not, i.e. rated as AIS1, a lot of vehicles passengers facing WAD will have a social issue concerning medical and economic point of view. Besides, it is frequently reported that a minor whiplash injury is hard to discover biomechanically even with great techs medical devices like MRIs and CT scanners. Hence, physicians usually cannot treat a sufferer accurately and also some false WAD insurance claims cannot be filtered<sup>10</sup>.

The objective of this study is to develop a new design of head restraint (HR) for car rear occupants. In this paper, the head-neck movements during impact and the dynamic response between females and males are presented. Besides, the whiplash mechanism is clarified based on various type of fields, such as medical, engineering, and mathematics perspectives.

### HEAD-NECK MOVEMENTS DURING IMPACT

In the primary phase, the subject is located on the seat in a regular location. While the car is struck, the structure acceleration is transferred to the seat by its anchorages, moving forward about the passenger. This subject's first zone, which is the

pelvis and the lumbar area, followed by the thorax, are receiving the seating force. When the spine is pushed forward, originally curved by its physiological shape, it tends to bend, move the neck base (vertebra T1) upward, and generate few compressions on it. The incident can be amplified by the entity thorax motion upwards caused by the seat angle and the base acceleration. The phenomenon is frequently named "ramping up." Even though the thorax starts to move, the head at this position remains in its initial situation. The T1 vertebra, positioned initially behind the head's centre of gravity, moves to be in front of it, and the previous throat compression becomes traction with the thorax pulling on the head. The T1 movement leads the cervical vertebrae to function as a chain, moving the motion from the lower end upwards, while the inertia of the head produces resistance to the action at the upper end. The combination of these reactions produces a transitory biphasic condition known as "S-shape" in that there is a very pronounced expansion of the reduced portion of the neck (vertebrae C5-C7) while the upper part is bending. The movement of the rear head, referred to as T1, is called retraction.

When the head starts spinning the whole neck reaches in extension phase with the head being pulled on by the thorax. As the base acceleration reduces, the elastic energy remains on the seat, and the passenger starts to be released, resulting from a rebound movement with a rotation forward of the torso around the pelvis of all the subject. When the body returns to its original place, normally the seatbelt begins to tense over the pelvis and thorax, creating a strong neck flexion. Lastly, the body is stopped owing to the tension of the belt and returns to the backrest. The motion phases are shown in Fig 1<sup>11</sup>.

#### THE DYNAMIC RESPONSE BETWEEN FEMALES AND MALES

Since the mid-1960s, injury statistics have continuously revealed that females face greater risk in suffering this kind of injury compared to males, around 1.5 to 3 times greater<sup>3,12,13</sup>. By comparing to males, females had a smaller head-to-HR distance<sup>14,15,16,17</sup> and earlier head-to-HR contact<sup>18</sup>. Females also had a smaller horizontal rear head displacement and larger cervical

vertebral angle with further severe S-shaped deformation<sup>19</sup>. Greater and earlier peak head horizontal accelerations<sup>20,18</sup> and higher (or similar) T1 accelerations<sup>15,16</sup> were recorded for females than males. Besides, smaller (or same) and earlier neck injury criterion (NIC) values were discovered for females<sup>17, 18</sup>. Females have faster seatback interaction and HR<sup>20</sup> and a greater pronounced rebound motion<sup>21</sup>. Unfortunately, the whiplash protection seat strategy has verified that males are more helpful than females<sup>22</sup>. These outcomes indicate that when both genders are seated, the safety performance of different seat concepts can differ. It is necessary to assess more and to consider the cause of such diversity to reduce the greater injury risk for female and to contribute further protection to both sexes<sup>23</sup>.

#### MODELS FOR CRASH SIMULATION

In the field of biomechanics, a large amount of literature has been published on the testing and understanding the whiplash mechanism, which is associated with rear-end impacts. To better consideration of whiplash injury and prevention mechanisms, collisions are simulated using human volunteer (in vivo studies), crash test dummies, cadavers (ex vivo studies), computer modelling and mathematical models. Every technique has particular advantages and limitations<sup>24</sup>.

##### Human volunteer

The human volunteer investigations contribute to passenger load and movement reaction corridors information that is the gold standard from other model reactions that can be contrasted. Dehner et al. simulated a rear-end impact with 6.3 km/h speed change ( $\Delta V$ ) in a sled test, with 8 female subjects who have no prior injury record or cervical spine pain. A high-speed camera was up to record movement input. Accelerometers were used for logging the acceleration data.

Acceleration load to the cervical spine was determined by a concurrent assessment of head angle and angular head acceleration, together with the concurrent evaluation of relative motion and relative acceleration between head and T1, to reflect intervals of enhanced danger of cervical spine injury during rear-end collision. The authors

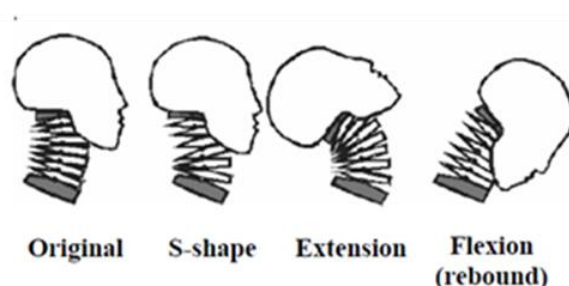


Figure 1. Various stages of the motion of the head in an impact<sup>11</sup>

reported that the motion sequence was characterised by the similar states which for male volunteers have already been defined. Increasing angular head acceleration may be explained by facet joint injuries during the expansion motion (100 - 120 ms) and therefore occurs approximately 50 ms later than that has been demonstrated in cadaver models. The merge of maximum ventral head acceleration and head motion in the late rebound is underestimated and may be accountable for soft tissue injuries<sup>25</sup>.

In 2019, 8 volunteers between the ages of 20 and 29 (mean 26.5 years, SD 3.34 years) were subjected to rear-impact sub-injury. Surface electromyography (EMG) was used before, during and after effect to record cervical muscle activity. Muscle response time and amplitude of the EMG signal have been analysed. Data on the acceleration of head, pelvis, and T1 were recorded. As a result, cervical muscle activity was discovered to be important. After the effect stimulus, respectively before maximum head speed (113 ms), sternocleidomastoideus, trapezius and erector spines were activated on average 59 ms, 73 ms and 84 ms<sup>26</sup>. Overall, these humans in vivo studies could not provide a better consideration of the basic biomechanics of structural elements potentially related to injury and/or pain generating processes. It is impractical to measure the forces or strains on intervertebral discs and individual ligaments in a living subject, and it would be unethical to expose the human volunteers to higher accelerations<sup>27</sup>.

### Crash Test Dummies

Crash test dummies (also referred to as anthropomorphic models) are full-body replicas that are used to simulate the effects of vehicle impact on humans. Population sampling has allowed for the creation of models suited explicitly for the study of rear-impact collisions. The anthropomorphic dummies (ATD) are made to represent the body mass and properties of the average human to best replicate the musculoskeletal response during whiplash. Due to these advantages, the ATDs are widely used in the automobile industry for various safety evaluations.

Crash test dummies are used in vehicle regulatory tests such as ECE R16, R94 and R95 (UNECE 2017) and consumer data tests such as NCAP tests (Euro NCAP 2017), to develop and evaluate occupant safety efficiency of a vehicle<sup>28</sup>. The average model of male crash test dummies represents the entire adult community. The male crash test dummies of the 50th percentile (the Biofidelic Rear Impact Dummy, BioRID; and the RID3D) are currently used to evaluate vehicles whiplash injury protection in low severity test. This crash test dummy approximately regarding mass and stature of a 90th to 95th percentile woman<sup>29</sup>. Hence, current seats and whiplash protection

devices are tailored primarily to the male 50th percentile without concern for female features, despite the increased danger of whiplash injury in women. Therefore, existing concepts of whiplash protection are more helpful to men than to women, with a 45% reduction in the danger of ongoing medical deficiency for women and 60% for men, according to insurance claim documents<sup>30</sup>. EvaRID (Eva – female / RID – Rear Impact Dummy), the world's first numerical collision test dummy of an average woman, was created as a first step in addressing this limitation<sup>31</sup>.

Even though ATDs have trouble simulating the complex motions of the spine during vehicle collisions, their usage has led to various safety improvements since their inception. In summary, similar tests of human volunteers on crash dummies might provide useful kinematic data<sup>32</sup>.

### Human cadaver (ex vivo studies)

Ex vivo studies were adequate to find out the biomechanics from sub injury to injury-generating load or acceleration levels in the past decade. All the isolated head-neck rear-end impact tests have provided invaluable information in understanding the whiplash injury mechanisms<sup>27</sup>.

Panjabi et al. conducted several studies to figure out the whiplash injury mechanism<sup>33</sup>. They estimated the potential risk of injury based on the damage caused by the impacts to the spinal structures (such as facet joints, capsular ligaments, and intervertebral discs) compared to the tissue damage by a physiological range of motion. They used isolated cervical spines and attached the surrogate head to simulate the whole cervical spine model.

Later, Ivancic et al. performed a study with 12 cervical spines (6 rear end impact-exposed and six control) and prepared 66 capsular ligament specimens (C2/3 to C7/T1) to determine if rear-end impact caused the increasement in capsular ligament laxity. The study was done by employing quasi-static loading to capsular ligaments exposed to whiplash and regulated. They reported a significant increase in capsular ligament laxity (0.9 mm at 5 N tensile load) compared to the control ligament<sup>34</sup>.

All these isolated head-neck tests have provided invaluable information into the complex. However, there are several problems remain. Firstly, cervical spine ligaments have been associated with whiplash injury, especially in the upper cervical region. The head of the specimen replaced by surrogate head destroys vital ligaments in the C0-C2 complex. Secondly, the postero-anterior loading of the rear end impacts does not consider the effects of thoracic ramping on the isolated head-neck complex. These separate head-neck specimens do not account for

the influence of compression due to the straightening of the thoracic spine<sup>35</sup>.

Studies with cadavers have shown the whiplash injury is the formation of the S-shaped curvature of the cervical spine which induced hyperextension on the lower end of the spine and flexion of the upper levels, which exceeds the physiologic limits of spinal mobility<sup>36</sup>.

Few investigators used full postmortem human subjects (PMHS) to reproduce whiplash injuries accurately. Later, the same researchers conducted tests of rear-end impacts using PMHS and captured the cervical motion with high-speed x-ray images. They reported the absence of S-curvature and supported the hypothesis that a significant facet joint stretch can be the neck pain source<sup>37</sup>. However, these studies could not elucidate the damage caused in the critical spinal structures (parameters like intervertebral disc stresses and upper cervical ligament strains).

### Computer modelling

Finite Element (FE) Human Body Models (HBMs) are the most commonly used injury assessment computer devices and essential devices for studying the human reaction to impact loads. Models of FE allowed for a further accurate and practical representation of the neck geometry and material behaviour. By using this model, more precise neck injury mechanisms can be investigated<sup>38</sup>.

Stemper et al. investigated the cervical spine issue by a rear-impact lordotic, straight or kyphotic neutral curvature<sup>7</sup>. The model is one of the 50th percentile male's different cervical spine models<sup>39,40,41, 42,43,44,45</sup>. Currently, the same attempt has not been created for the 50th percentile female<sup>46</sup> that limit the development of vehicle protection systems focuses on neck injuries for the incidence is affected by the gender of the accident sufferer<sup>47</sup>.

Using a parameterized head-neck FE model, John et al. used a variety of time-history responses and examined the effect of parameters on the S-curve shape. Most affected by the anteroposterior location of the head centre of mass and segment size was the time to maximum S-curve. The maximum formation of the S-curve (peak flexion in the segment C2-C3) was most affected by the depth of the vertebral body, disc height, followed by the anteroposterior position of the mass head centre. The impact of gender-dependent vertebral body depth and segment size may show that the vertebral column's 'female-like' morphological differences result in enhanced segmental rotations and may predispose females to whiplash injuries.

The limitations of the PMHS sled test data include the lack of active musculature and the boundary

condition at the inferior end of the head-neck complex. In the models, the muscles have not been activated, and their impacts on the reactions have not been assessed. The mini-sled experimental setup applied a linear acceleration to the T1 and therefore did not consider the torso and seat interaction rotational movement of T1. The model was also not validated with any local tissue response as it was not obtainable from the simulated experiments conducted in this research<sup>48</sup>.

Most computer simulation models provide poor accuracy for cervical and head motions during whiplash due to the difficulty of capturing the movement and properties of the intervertebral discs, facet joints, ligaments, muscles, etc. throughout the dynamic whiplash motion. Complex finite element models have been able to approximate the intervertebral motion at the tremendous computational expense. This model must also provide an accurate response for both the head and intervertebral movements to enable a precise assessment of injury given our current understanding of whiplash trauma.

### Mathematical models

Nowadays, some criteria of injury are applied in the analysis and whiplash risk evaluation in automotive rear impacts. The Neck Injury Criterion (NIC)<sup>49</sup> was calculated applying the relative horizontal acceleration ( $a_x$ ) and velocity ( $v_x$ ) between the head and T1 centres of mass, where 0.2 is the length parameter. In NIC measurement, change of hydrodynamic force in the spinal canal is recognised to influence the head relative acceleration to the first thoracic vertebra during the initial state or S-shape deformation state, generally within 150 ms after the crash. NIC is calculated using the following Equation (1).

In 2002, Neck Protection Criterion ( $N_{km}$ )<sup>50</sup> was suggested based on the linear combination of shear forces ( $F_x$ ) and sagittal bending moments corrected to the occipital condyle ( $M_{y_{oc}}$ ), measured with the upper neck load cell. This criterion distinguishes equation among four possible conditions depending on the sign of and, as shown in Table 1. The Neck Protection Criterion was determined as Equation (2).

Besides, criterion Lower Neck Load Index (LNL) was suggested, shown in Equation (3) associates that three force elements and two of the moment elements measured at the neck base<sup>51</sup>.

Mainly, the LNL value depends on the usage of the dummy. The suggested LNL-index with the RID2 dummy has demonstrated excellent correlation to the real-world claim data in the investigation. The LNL for the rebound was not assessed as realistic seatbelts were not used in the testing. However,

$$NIC(t) = 0.2 * a_x^{Headrel.T1}(t) + [v_x^{Headrel.T1}(t)]^2 [m^2 / s^2] \tag{Equation (1)}$$

$$N_{km} = \frac{|F_x|}{F_{int}} + \frac{|M_y \text{ OC}|}{M_{int}} \tag{Equation (2)}$$

$$LNL-index(t) = \left| \frac{\sqrt{M_{y_{lower}}(t)^2 + M_{x_{lower}}(t)^2}}{C_{moment}} \right| + \left| \frac{\sqrt{F_{x_{lower}}(t)^2 + F_{y_{lower}}(t)^2}}{C_{shear}} \right| + \left| \frac{F_{z_{lower}}(t)}{C_{tension}} \right| \tag{Equation (3)}$$

$$E = l_s - l_i \tag{Equation (4)}$$

Table 1: Cases of  $N_{km}$ <sup>50</sup>

Case	$M_y$	$F_x$
$N_{fa}$ (Flexion Anterior)	>0	>0
$N_{fp}$ (Flexion Posterior)	>0	<0
$N_{ea}$ (Extension Anterior)	<0	>0
$N_{ep}$ (Extension Posterior)	<0	<0

theoretically, an application of the LNL to the rebound state should be possible.

Facet joint capsular ligament elongations were verified as a function of cervical posture, spinal level, and anatomic region. Ligament elongations (E) were described as the length increment during S-curvature from the primary length, as shown in Equation (4)<sup>11</sup>.

Mathematical simulations presented the greater head and T1 horizontal accelerations for females compared to males<sup>52</sup>. The mathematical model output is dependent upon the accuracy of complex input data, containing detailed spinal anatomy, muscle, and ligament mechanical properties, and muscle activation styles<sup>24</sup>.

**METHODOLOGY**

In this study, the design of head restraint (HR) for car rear seat occupant is improved. Firstly, the characteristics of HR is identified. A vehicle seat HR assembly, consisting of a portion of the head support, a part of the neck support and a pair of connection members. The connection members are adapted to connect the HR assembly to the vehicle rear seat.

The HR assembly must be of sufficient size to provide support to the occupant without being so large as to impede the driver’s view from the car, including the ability to see other vehicles, barriers, pedestrians, cyclists, etc. The head support portion includes a frame structure that provides a rigid structure to the HR. The head support portion comprises at least a front face against which the rear of the occupant’s head is placed in use. The neck support portion consists

of any suitable size or shape of configuration. The neck support portion is associated with the head support portion. The head support portion and the neck support portion are fabricated separately to one another and adapted for fixed or temporary connection to it. Otherwise, the head support portion and the neck support portion are built as a single unit. The target is both support portions together form a body while the connection members comprise rods adapted for insertion in the bores of the rear seat – the diameter of the bore around 10-15mm.

The head and neck support portion are fabricated from natural materials (wool, cotton), synthetic materials (polyurethane) or polymeric foam (polyurethane foam, polyvinyl chloride foam, styrofoam). The head support portion is fabricated of a suitable thickness of cushion to absorb the forces experienced in the collision, thereby cushioning the occupant’s head and reducing the whiplash injury. The connection members are made from plastic or metal (steel, aluminium). Based on the size and the materials as the design parameters, the new HR is designed.

Then, the crash test simulation is run using Ls-Dyna simulation. The finite element simulations are run with a formerly developed model of the ligamentous cervical spine<sup>53,54</sup>, which is complete with a skull and soft tissues to form the head-neck model (Fig. 2). As described by Schneider, the model is created based on a 31-year-old female subject of 161.6cm height and 60.8kg weight, not over 0.1% and 2% from the aim of 50th percentile female<sup>55</sup>. The model is issued and shared under an Open Source license.

The common load scenario in a rear-end impact is as follows; (i)The automotive accelerates forward when struck (ii)The torso is pushed forward by the seat (iii)The spine starts straightening and the neck/torso joint rises. The anterior injury at C5-C6 level is observed, and the head-neck response from the simulation result is analysed. During the impact, all data of anterior T1 angle and neck extension (Fig. 3) are recorded, and the relationship between the item is determined.

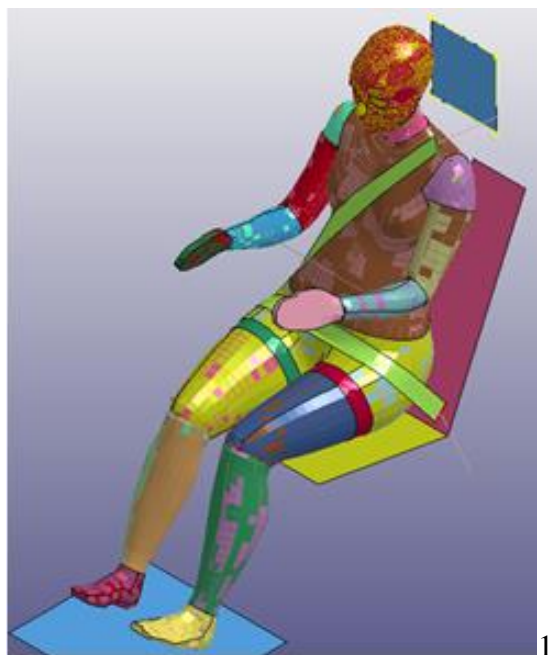


Figure 2: FE model<sup>47</sup>

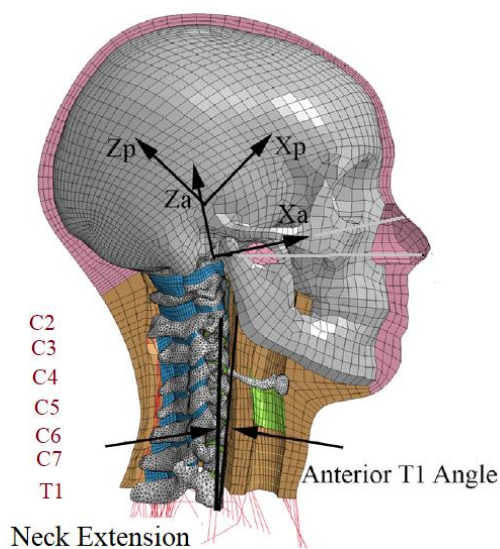


Figure 3: Model of head and neck for female<sup>47</sup>

The new design is repeated many times until correlation agrees with real data from sled-test. The accepted correlation equation from the new design is validated using Matlab simulation in order to minimise the NIC values (Equation (1)). Value 15 [m<sup>2</sup>/s<sup>2</sup>] is recommended as the threshold value for whiplash prevention. The target is to develop a model for optimising HR.

## WHIPLASH PREVENTION MECHANISM

HR can be effective against whiplash injuries in crashes, but for this to happen, the restraint itself must start with specific attributes. At a minimum, the precise geometry is required before an HR even has the potential to provide adequate protection in rear-end crashes. Current improvements in HR geometry and energy-absorption seat capability have motivated the decrement, however not total removal, of whiplash injuries<sup>56,57</sup>.

Many different design strategies have been proposed to reduce whiplash injury. The procedure for lowering whiplash injury in the event of a collision can be lumped into three main categories: (i) minimise energy transferred to the occupant, (ii) restrict occupant head movement, and (iii) alert the occupant to look forward and engage neck muscles. Opportunities for reducing the energy transferred to the occupant are limited because of high-speed collision safety requirements and space restrictions within the vehicle.

Volvo has designed a system that allows the seat to swivel backwards when the torso is pushed back into the seat during a rear-end collision. The design absorbs some of the energy of the crash and encourages a flexed cervical spine during the crash. This design is considered an active system because it is activated during an accident to protect the occupant from injury. Different active systems have focused on restricting the occupant head movement to prevent damage. For these systems, the HR will move forward to close the gap between the head and the HR to provide support early in the collision. It should be noted that the HR cannot be designed to have any initial gap with the passengers' head because it must not interfere with passengers' head with different driving postures and during movements to check mirrors etc.

The Saab Active Head Restraint (SAHR) and Mercedes Neck-Pro are two examples of active HR systems. These types of active systems have been found to reduce whiplash claims by 31-75% compared with conventional designs. There is potential for further gains as well. These active HR designs are only semi-active in that they are only activated once in the collision sequence (moving forward to close the gap between the passenger and the head). During normal driving, the conditions the active HR would move to maintain the desired top seat and backseat between the head and head restraint without interfering with passenger motions. Upon sensing a collision, the active head restraint would activate to close the gap between the head restraint and head and then move optimally to minimise injury. Further safety improvements to provide the optimal head restraint characteristics

(for standard and semi-active designs) and response (for active designs) have been limited by the development of robust human dynamic simulation models and the understanding of whiplash injury.

## CONCLUSION

The mechanisms of the whiplash injury were reviewed to achieve a further inclusive understanding of human impact reaction, variability quantification, validation, and prevention. Each has unique advantages and limitations. A review of the mechanisms indicates the head restraint (HR) design improvement for whiplash protection in the vehicle safety area is necessary because of the high incident rate of rear crashes. By employing the mechanism and the necessary parameters to some model, the optimised HR characteristics for whiplash prevention can be obtained.

Typically, conventional HR provides negligible support for the neck of an occupant, which means that injuries such as whiplash can happen in the case of a vehicle impact. There would, therefore, be a benefit if it were feasible to provide an HR for a car seat that offered an enhanced comfort for the occupant of the car, as well as better support for the occupant's neck, especially in the event of an impact. Also, the finite element modelling could be extended by using different body types of vehicles like compact cars, utility vehicles, etc. Different types of vehicles may have different types of mountings of HR on the rear seat and can provide additional factors for understanding rear-seat occupant protection.

## COMPETING INTERESTS

There is no conflict of interest.

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