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An investigation into neck injuries in simulated frontal impacts

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Abstract

Spinal injury is one of the most debilitating and costly injuries. It is a devastating condition that disrupts the life of the injured and their families. Vehicle crashes are the most common cause of fractures, soft tissue injuries and dislocations of the neck and cervical spine. Among vehicle crashes frontal impacts are the most frequent cause for head and neck injuries. Neck injuries are often caused due to unusual head motion. Improvement of the knowledge of the correlation between crash dynamics, human body behaviour and internal neck phenomena could contribute to the development of new protection systems for the neck. The most acceptable way to study the behaviour of the human body and internal interactions during car crashes is mathematical modeling as it is non-invasive and repeatable. In this work, computer simulations have been performed using a multibody dynamic model of the cervical and thoracic-lumbar spine, where rigid bodies are connected by articulated joints and spring-damper elements. The models were developed using the 'Working Model 2D' and were used to simulate frontal impacts in vehicles. The models were validated based on experimental data available in literature. The verified models were used to analyse the behaviour of the driver and vehicle kinematics and calculate the internal neck forces and motion. Principle virtual power of neck was applied at inter-vertebral levels for various impact speeds. Principle virtual power of neck was then correlated with real world crash data of neck injuries. It has been shown that principle virtual power of neck at each intervertebral level correlates well with the crash data and can be used as a predictor of neck injuries.

Keywords

Neck injury, spinal injuries, cervical, multibody simulations, risk, peak virtual power, frontal impact

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Introduction

Frontal impact is defined as a planar crash with impact location on the front of vehicle with the principal direction of force (PDoF between 11 and 1 o'clock).

Neck injuries are the most frequent disabling injuries among car occupants in road traffic accidents. One or all of these injuries may be the source of pain in whiplash casualties leading to the clinically observed symptoms of neck, shoulder, or back pain, headache, dizziness, paresthesias or vertigo.

Chronic neck injury results in a large economic burden of medical and insurance costs and loss of work.[1] In 2007, an estimated 66% of all insurance claimants under bodily injury liability coverage and 57% under personal injury protection coverage (two of the most important insurance injury coverage) reported minor neck injuries. For 43% and 34% of bodily injury liability and personal injury protection claimants, respectively, neck sprains or strains were the most serious injuries reported. The cost of the claims in which neck pain was the most serious injury was about $8.8 billion, representing approximately 25% of the total paid for all crash injuries combined.[2]

Improvement in the automotive safety systems in the past decades have steadily reduced the severe injuries among car occupants in general.[3] However, neck injury has increased substantially, both in terms of risk and

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in number. In England the incidence of neck injuries has been shown to have doubled over a 10-year period.[4]

Krafft[5] has shown that there is a 2.7 times higher risk of sustaining disability to the neck in cars launched at the end of the 1980s and in the beginning of the 1990s compared to those launched at the beginning of the 1980s. AIS1 neck injuries have become the major disability caused by collisions with modern vehicles. It has been shown that approx. 60% of the injuries causing disability in Sweden between 1990 and 95 were AIS1 neck injuries.[5]

Nygren[6] found that 10% of the occupants in rear impacts and 5% of the occupants in frontal impacts that reported initial whiplash symptoms suffered permanent disability at least 1 year after the collision.

Injuries to the neck are often regarded as a problem in rear-end impacts. However, soft-tissue injuries have been reported in 23–70% of individuals involved in frontal impacts.[7–12]

Despite this high injury prevalence, the underlying soft tissue injury mechanisms are not fully understood.[13–15] Different hypotheses exist concerning injury mechanisms in rear-end impacts, covering both flexion,[16] extension[17] and hyper-extension.[18] In frontal collisions, Larder et al.[19] found that even when no head contact with the interior of the vehicle compartment had been noticed, the forward flexion of the neck was assumed to be the injury-causing motion. Walz and Muser[20] also described the motion of the head relative the neck in a frontal collision with no head contact. For a restrained occupant in the initial phase of a collision a purely translational motion occurs[21] forming an S-shaped cervical spine. After that the cervical spine is forced into flexion. Walz and Muser[20] suggested that neck injury may occur in such acceleration mechanisms.

There are few biomechanical studies documenting injuries to the cervical spine ligaments due to flexion-type loading, the most likely injury mechanism during frontal impact.[14] Frontal impact causes middle C2–3 to C4–5 and lower C6–7 and C7–T1 injury.[13] Direct facial impact has shown a flexion motion of the upper or middle cervical spine, with extension of the lower cervical spine.[22] Panjabi et al.[13] found that ligaments had the highest peak strains occurring at C3–C4, suggesting that this intervertebral level may be at the greatest risk for injury during frontal collision.

Neck injury mechanisms have been the subject of research for many years. Typically two methods, multibody dynamics (MBD)[23–30] and finite element methods (FEM)[31–39] are being used to study human spine and soft tissues injury mechanisms in order to have a better understanding of the kinetics, kinematics, and clinical aspects.

The literature shows various investigations of chest, pelvis and neck injuries, together with injury mechanisms, tolerances, and comparison with impact dummies using post-mortem human subjects (PMHS).[40–43] Some of the literature shows studies of cervical spine injuries based on different anthropometric measurements on human volunteer test series.[44,45] It is impossible to carry out experiments on living volunteers to such a degree that injuries are produced. Therefore, there has been great emphasis on computational simulation.[46–55] FE models need a great deal of computational power, but can provide detailed information about tissue deformations and injury prediction. Multibody models can also include many anatomical details while being computationally efficient. This makes them suitable for parameter optimisation analyses.[56,57]

In many previous studies, numerical or physical models have been developed to investigate neck injuries and use different neck injury criterion for estimating the likelihood of neck injuries. However, there has been little work investigating the risk of neck injuries received by occupants in real accidents and comparison with the calculated injury criteria. A number of neck injury criterion like the N_{ij},[58,59] and NIC[60] have been investigated using these models. Sturgess[61] has proposed the peak virtual power (PVP) as a reliable neck injury criterion particularly for intervertebral soft tissue injuries. In order to compute power, intervertebral forces and motions play important roles.

This study further investigates the intervertebral forces and motion using a simplified model of the spine and demonstrate their efficacy in estimating risk of neck injuries.

A 2D multibody model of the whole spine has therefore been developed in Working Model software. The models were validated against the data of sled tests conducted at the Naval Biodynamics Laboratory and those available in literature.[62–65] The validated models were then used to analyse the behaviour of driver/vehicle kinematics and the neck forces and motions at intervertebral levels for various impact speeds.

The article first gives an overview of the severity of neck injury in frontal impact. This is followed by a description of the models developed and their validation. The simulations developed are then presented along with the results of the simulations and their analysis.

Severity of neck injury

Historically, vehicle change-of-velocity (Delta-V or ΔV) and the resultant responses of the human occupant have been the most commonly used index for reporting injury risk in various types of real-world motor vehicle collisions.

Some investigations have shown a correlation between the change of velocity and initial neck injury symptoms in rear impacts.[16,66] However, Krafft[5] found that given an initial neck injury, there was no difference between the change of velocity leading to long-term consequences, than those leading to short-term consequences. This indicates that the acceleration levels seem to influence the disability risk.[67]

Cervical spine fracture thresholds in real-world frontal and rear-end crashes do differ with Delta-Vs. Otte et al.[68] had reconstruction analysis of 117 crashes and compared medical and radiographic diagnosis of these occupants. Their study found that the lowest Delta-V for a fracture in a frontal crash was 17 km/h and was 10 km/h in a rear end crash.

Results from real-life rear-end impacts where the crash pulse has been measured with a crash pulse recorder indicate that acceleration levels seem to better explain the neck injury risk than does change of velocity.[5] In frontal impacts, little is known about how impact severity relates to neck injury risk. Norin et al.[69] found that the neck injury risk increased slowly with increased impact severity measured by the equivalent barrier speed (EBS). The link between neck injury risk in frontal impacts and impact severity parameters based on acceleration levels remains unknown.

In analyses of injury risks in real-life collisions, it is important to determine adequate, valid and reliable impact severity parameters which influence the injury risk.[70] The possibilities of using different impact severity parameters in retrospective reconstructions methods are limited. Most reconstruction methods use the equivalent energy speed (EES)[71] or change of velocity (ΔV) to describe impact severity. Attempts have also been made to estimate mean acceleration. However, other impact severity parameters may better relate to injury risk. A study by Kullgren[70] showed that mean and peak accelerations separately better explain the overall injury risk than does change of velocity. Since the injury risk may depend on several parameters in the crash phases of an impact, it is important to have better reconstruction methods, where adequate severity parameters can be measured or reconstructed with accuracy.

The severity of neck injury can be identified by using the MAIS-ΔV curves for both restrained and unrestrained occupants. Based on data collected from real word accidents in Saudi Arabia, the relationship between MAIS and ΔV has been found to vary between a square and cubic manner by Sturgess.[72] The results are shown in Figures 1 and 2.

It can be inferred from above graphs that ΔV^3 gives the best correlation for belted drivers, whereas for the unbelted drivers, the relationship between MAIS and ΔV is quadratic. On this basis Sturgess has showed that unrestrained occupants suffer higher injury levels at lower ΔV than do restrained occupants, because they are subject to higher PVP inputs for a given ΔV.[72]

As stated by Sturgess,[72] the high degree of correlation demonstrates that, by making very simple assumptions about idealised impact types, a simple theory can account for 85–90% of the injuries obtained from Co-operative Injury Study in the UK (CCIS – Phase 7) and NASS-CDC Databases. The fact that all injuries require an expenditure of energy means that energy methods are independent of injury mechanisms; therefore, PVP is a good candidate for a universal injury criterion which can be correlated with real-world injury

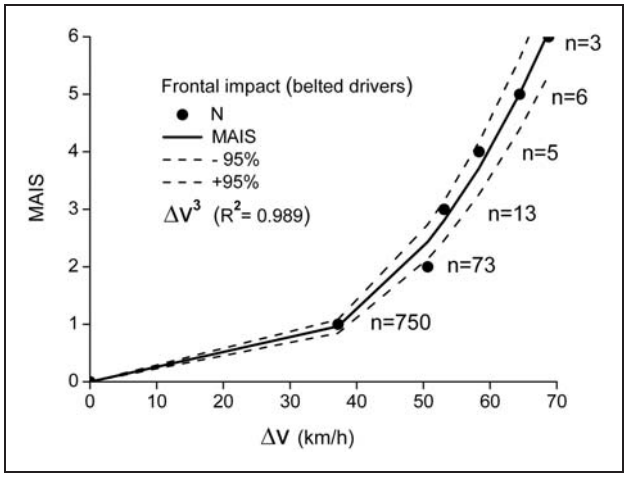

Figure 1. MAIS vs Delta-V for frontal impact, belted drivers, for neck injuries.
MAIS: maximum abbreviated injury scale.

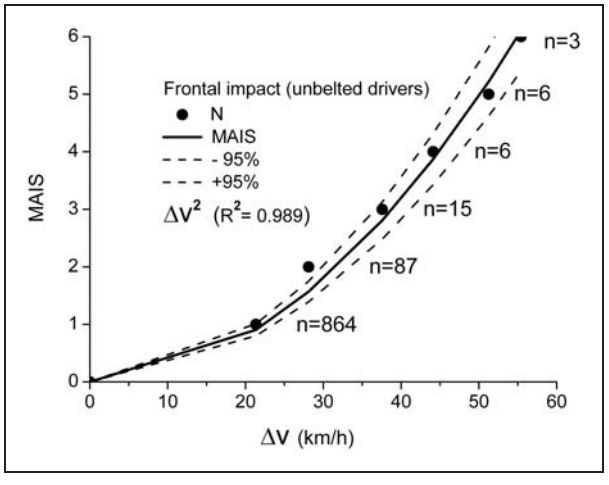

Figure 2. MAIS vs Delta-V for frontal impact, unbelted drivers, for neck injuries.
MAIS: maximum abbreviated injury scale.

experiences. Furthermore, energy is the only physical quantity that remains unchanged at all scales, and so PVP can be applied at the micro, meso and macro scale.

Currently there is very little correlation between neck injuries received by occupants in frontal impactss and the severity of these crashes. In a recent study, Al-Shammari[73] has derived risk curves using ΔV for a group of 3035 occupants who sustained MAIS $\geqslant 1$ neck injuries. In this study the severity of neck injury has been predicted using the multidimensional logistic regression statistical technique.[74,75]

The probability of neck injury (MAIS $\geqslant 3$ or MAIS 1-2) can be modeled by equation (1)

$$P(\text{MAIS} \geqslant 3) = \frac{e^{g(x)}}{1 + e^{g(x)}} \quad (1)$$

where g(x) is a function of the independent variables.

The probability of neck injury for frontal impact as a function of change in speed has been plotted in Figure 3. These curves were plotted for both belted and non-belted occupants. It can be seen from the graphs that the probability of neck injuries with severity MAIS $\geqslant 3$ is the highest for the non-belted occupants as compared to belted occupants at intermediate values of Delta-V.

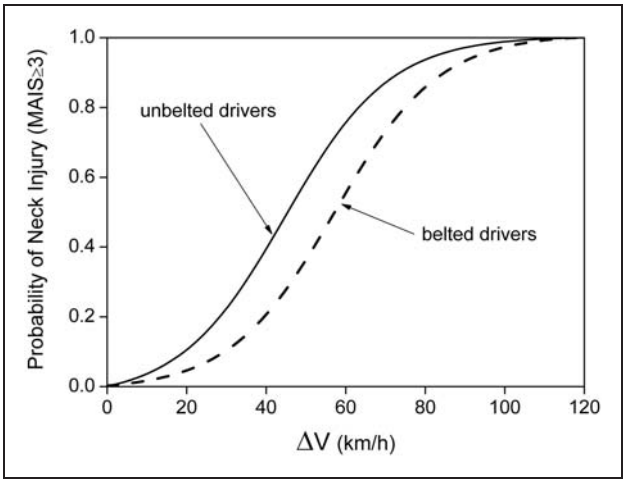

Figure 3. Risk of neck injury for belted and un-belted drivers in frontal impacts.

Frontal impact simulation

Model construction

In this study, models have been developed to study neck injuries in frontal impact simulation according to the Insurance Institute for Highway Safety (IIHS) frontal offset crash test.[76] In the Institute's test, frontal impact takes place at 64 km/h (40 mph). Forty percent of the total width of the vehicle strikes the barrier on the driver side. Each car tested is subjected to an offset impact into an immovable block fitted with a deformable face. This impact is intended to represent the most frequent type of road crash, resulting in serious or fatal injury. It simulates one car having a frontal impact with another car of similar mass.

A Hybrid III dummy representing an average-size (50^{th} percentile) male is positioned in the drivers seat.

Figure 4 shows the initial configuration of the model. The modelling of frontal impact consisted of three systems combined in one MBD model: the car; the deformable front barrier (DFB); and the driver dummy in the sagittal plane.

The front barrier is composed of three elements: the base unit, extension and deformable units. The base unit is 184 cm high, 366 cm wide and 542 cm deep.

Table 1. Mechanical properties of the DFB.[76]

Mechanical parameter	Unit	Value
Base mass (m_1)	kg	5×10^5
Deformable face mass (m_2)	kg	100
Extension mass (m_3)	kg	100
Spring stiffness, k_1	N/m	1508.50
Damping coefficient, b_1	N.s/m	1000
Damping coefficient, b_2	N.s/m	8.5×10^5

DFB: deformable front barrier.

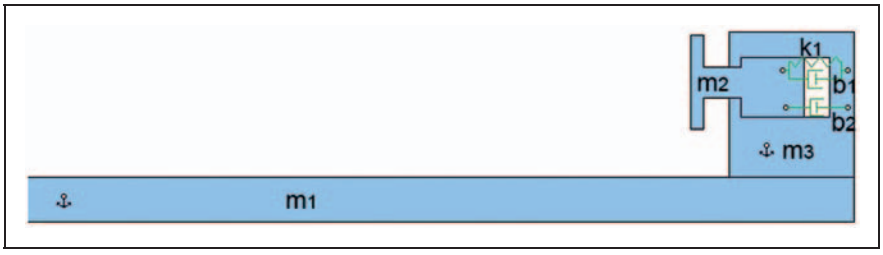

Figure 4. Modelling of the deformable front barrier.

Table 2. Mechanical properties of vehicle.[86]

Element	Parameter	Value
Body of vehicle	Mass	800 kg
Front bumper	Mass	150 kg
Rear bumper	Mass	150 kg
Armchair assembly	Mass	22 kg
St. wheel/pedals	Mass	28 kg
Wheels	Mass	50 kg
Crash zone	Stiffness, k_t	1500 N/m
	Damper coefficient, b_t	7×10^5 N s/m

The extension is 91 cm high, 183 cm wide and 125 cm deep. The front barrier has been simulated by a combination of rigid parts, connected through a set of springs and dampers.[76] The mechanical properties of the front barrier are given in Table 1.

The car model used for simulation of frontal impact was developed using the data for a Toyota Corolla car. The vehicle has been modelled from regular geometrical shapes representing rigid bodies, which are joined in an appropriate way to consider the effect of the energy dissipation at the moment of collision as well as characteristics of body deformation. The pedals and

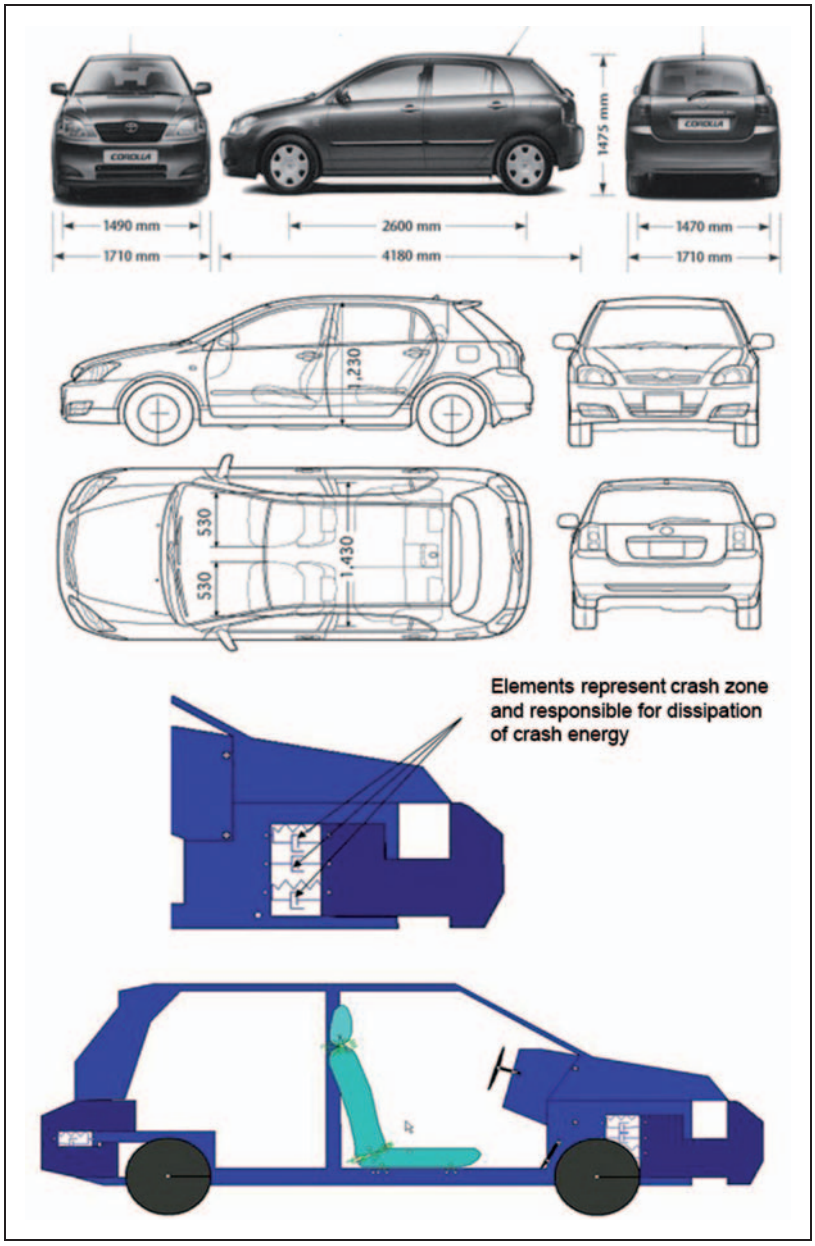

Figure 5. Main components of the simulated car.

the wheel were modelled as rigid elements joined to the body through bracket joints.

The details of the main car parts used in the simulations are provided in Table 2 and Figure 5 gives the details of the armchair and the car body contacts including the seatbelts.

The kinematics of each rigid body element used in the simulation is described after fixing the origin of the local coordinate system at the centre of mass of the body from which the displacement, velocity and acceleration are defined in accordance to the global coordinate system as shown in Figure 6. The movement of the human body in frontal impact has been analysed in the sagittal plane (XY) of the coordinate system. The geometry of the whole spine for a 75-kg male was preliminary downloaded from an internet resource (www.3dcafe.com). The model of the human spine was then rescaled by using Rhinoceros Nurbs Modelling v3.0 SR5 according to the dimensions of a real adult male spinal column.[41] The inertial parameters of each spinal vertebra were calculated separately about the X, Y and Z axes by using Solid Works Office Pro 2007.

Material properties were assumed to be homogenous bone structure with a density of 1.5 g/cm^3 which is the median value for cortical and trabecular tissues of vertebral bone. The mass and moment of inertias of the vertebral spine in the sagittal plane (I_z) are presented in Table 3.

In the model, the dynamic stiffness of the disc is assumed to be twice the static stiffness to approximately allow for strain-rate effects. Vertebral disc responses are obtained by subjecting a motion segment (vertebra-disc-vertebra) or a disc segment (body-disc-body) to external loading. Disc stiffnesses reported by Moroney et al.[77] and Yoganandan et al.[40] were used for the cervical spine.

Material properties for cervical spine discs were obtained from Yoganandan et al.[40] For thoracic spine, the stiffness values given by White and Panjabi[78] were employed in the modelling. For lumbar spine, the motion segment stiffness matrix results of the study of Gardner-Morse and Stokes[79,87] was utilised in the modelling.

The determination of the compression strength of the human vertebrae has been the subject of research from the early days of biomechanics. The results of some studies, in the form of strength vs. vertebral level, are shown in Figure 7.

The main parts of the human body modelled in are the head, spine, rib cage and the upper and lower limbs. The elements of the model were joined by articulated joints as open kinematic chains and additionally joined with spring-damper elements. Each of the muscles and ligaments have been modelled using appropriate spring and damper elements.

The upper and lower limbs were connected by a pivot with a rotational spring-damper systems. The vertebral spine model includes 24 solid vertebrae, muscles, ligaments, inter-vertebral joints and discs as shown in Figure 8.

The driver was then inserted in the vehicle model and joined by spring-damper elements representing the flexible connections between the human body with the armchair, pedals, seat belts and steering wheel as shown in Figure 9.

The stiffness and damaping coefficients of these elements are given in Table 4.

Validation of the models

The car model has been validated against available IIHS tests data for the same vehicle in frontal barrier tests.[65] Figures 10 and 11 show a comparison between the acceleration and velocity of the vehicle based on the simulation and the IIHS tests. Since the peak accelerations as well as the acceleration pulse for both are similar, this was considered a good match and the car model was considered acceptable for further use.

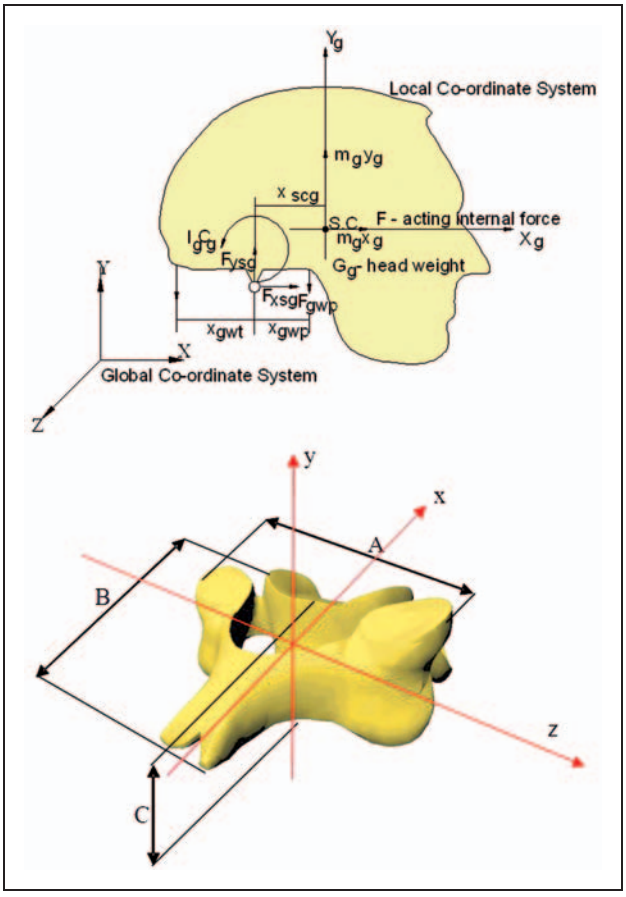

Figure 6. Global/local coordinate system in the model.

The neck model for frontal crash was validated against the data taken from volunteer studies subjected to sled tests accelerations ranging from 3 to 15 g. These sled tests were performed at the Naval Biodynamics Laboratory.[62] Response corridors from these tests are presented by Wismans et al.[63] and Thunnissen et al.[64]

A detailed description of the instrumentation and test methods is provided in Yoganandan et al.[42] The subjects are restrained by shoulder straps, a lap belt and an inverted V-pelvic strap tied to the lap belt. Upper arm and wrist restraints are used to prevent flailing. In addition a lightly padded wooden board is placed against the right shoulder of the subject to limit the upper torso motion. The volunteers were asked to take a normal automotive posture. The initial head angle was 0°, where the head angle was defined as the angle between the Frankfort plane and the horizontal plane. The Frankfort plane is defined as the imaginary plane passing through the external ear canals and across the top of the lower bone of the eye sockets. The test conditions are summarised in Table 5.[64] Mean values of the sled acceleration-time histories are shown in Figure 12.[64]

The validations are considered to be good if the response was within the corridor of the experimental data, it was considered to be reasonable if the response was within a 25% deviation, while it was considered to be poor if the deviation was more than 25%.

The overall response of the front model is shown in Figure 13 in which the maximum heal flexion was noted at approximately 80 ms. Figure 14 shows that the head CG acceleration follows the same double peak as that observed in the experiments. The head forward rotation in the simulations is very similar to that in the experimental data. The overall correlation was better than 25% which was taken as reasonable according to the validation criteria defined above.

Simulations developed

Frontal impact simulations were conducted at impact speeds of 27, 80, 120 and 160 km/h using the car and the

Table 3. Moments of inertia, mass and dimensions of vertebras.[73]

		Mass moments of inertia			Dimensions		
Vertebrae	Mass (g)	I_x (g mm^2)	I_y (g mm^2)	I_z (g mm^2)	*A* (mm)	*B* (mm)	*C* (mm)
C1	21.1	8093.4	4248.45	11,924.4	85.1	59.7	18.6
C2	23.5	3436.8	4901.4	7257	57.8	60.3	29.7
C3	22.6	3222.45	4871.85	7067.55	56.2	61.2	14.6
C4	24.3	3990	5199.45	8079.9	58.3	61.2	14.6
C5	28.7	4940.25	8316.75	12,170.1	63.3	68.9	14.6
C6	29.4	5626.95	8392.65	12,822.15	71.1	68.9	14.4
C7	40.4	8143.8	9644.1	14,731.2	73.6	72	19.4
T1	40.3	9093.9	9937.95	14,545.2	94.9	70.1	19.3
T2	40.0	7128.15	9721.95	12,623.55	73.2	72.6	20
T3	48.5	9570.75	13,664.55	17,698.5	77.8	78	21.2
T4	55.9	11,912.25	17,352	22,233	76.4	82.9	22.3
T5	56.3	11,804.1	17,302.5	22,215	76.4	82.3	22.3
T6	60.8	12,942	22,192.5	27,441	76.4	87.1	22.4
T7	62.4	13,263.6	22,480.5	27,634.5	79.9	79.7	23.4
T8	68.4	15,633	22,884	29,808	79.9	74.7	23.4
T9	73.3	16,186.5	24,618	31,246.5	66.8	76.8	23.1
T10	88.0	20,677.5	32,013	37,747.5	56.3	79	27.5
T11	89.3	20,241	34,623	40,461	56.3	82.3	28
T12	89.3	20,220	34,683	40,542	56.3	82.4	28
L1	89.7	21,642	33,895.5	40,257	75.1	77.4	29.4
L2	86.4	20,587.5	33,579	39,921	75.1	77.3	33.1
L3	95.9	24,291	38,938.5	46,333.5	77.4	79.4	34.2
L4	96.8	24,697.5	39,570	46,929	77.5	80.7	34
L5	84.7	21,019.5	28,614	37,368	77.5	77.9	32.8
Sacrum	250.1	299,895	170,580	207,240	113.9	80.7	121.9

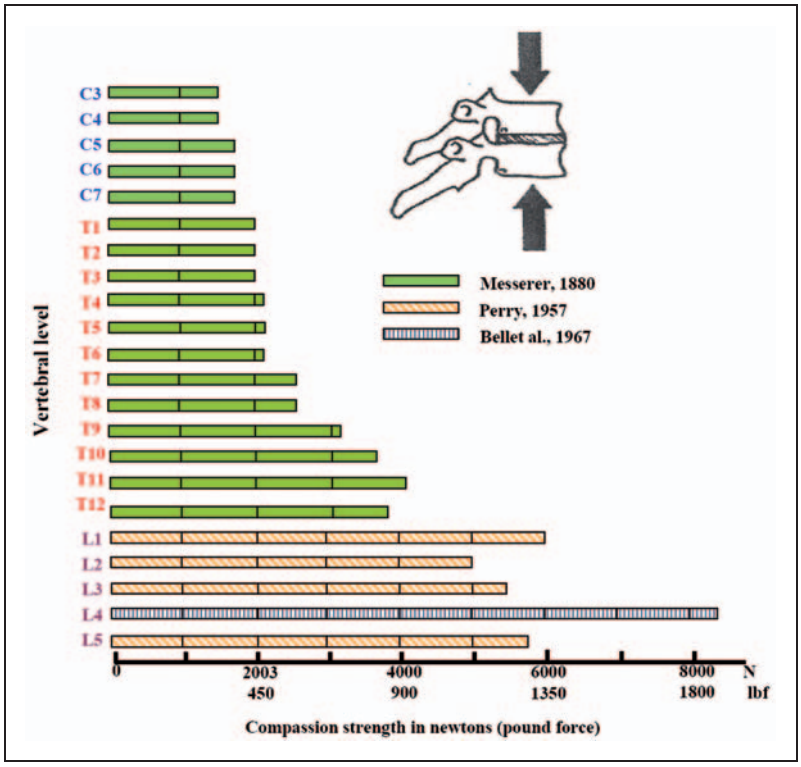

Figure 7. Vertebral static compression strength.

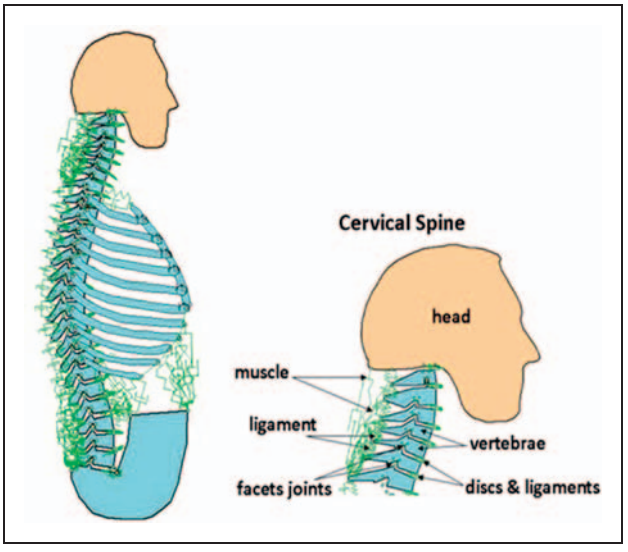

Figure 8. Modelling of the head and the whole spine in the sagittal plane.

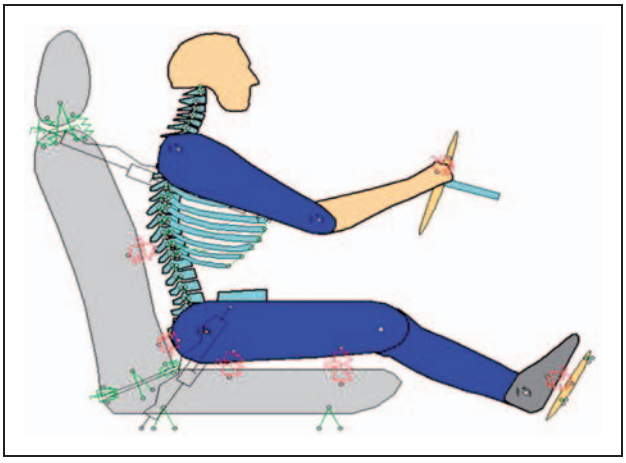

Figure 9. The connections between the driver and car in the sagittal plane.

dummy models described above. The simulations were run for the belted as well as the unbelted driver. A total of 8 simulations have thus been performed and the kinematics of the occupant and of the spine have been analysed. Intervertebral forces and velocities have then been used to compute the PVP in the intevertebral and levels and have been compared with incidences of intervertebral injuries in frontal impacts obtained from real world accidents.

The N_{ij}, forces and moments at the occipital condyles have been calculated according to equation (2)

$$N_{ij} = \frac{F_Y}{F_{YC}} + \frac{M_{OCZ}}{M_{ZC}} \quad (2)$$

Table 4. Stiffness and damping values for driver/car contact points.[86]

Contact point	Stiffness (N/m)	Damping (N s/m)
Hand-steering wheel	5000	50
Foot-pedal	500	50
Leg-seat	5000	1000
Sacrum-seat	3000	1000
Sacrum-backrest	8000	1000
Thorax-backrest	1500	1000
Lower seatbelt	3×10^5	5000
Upper seatbelt	1×10^5	500

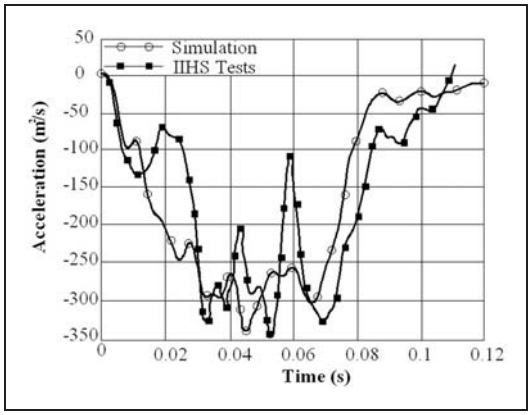

Figure 10. Comparison of vehicle acceleration obtained from simulation results and IIHS test data.
IIHS: Insurance Institute for Highway Safety.

The critical values of $F_{YC} = 6160$ N and, $M_{ZC} = 310$ Nm were used.[80]

In order to estimate the risk of neck injuries, Sturgess has argued that principle virtual power (PVP) could be a better predictor of the likelihood of injuries. This article investigates the PVP in the crashes and how well it predicts the likelihood of neck injuries. For the following neck injury evaluation, the PVPn of neck is based on the acceleration and velocity change of the neck.[61]

The PVP at each intervertebral level can be defined in equation (3)

$$PVPn = F.V + M.\omega \quad (3)$$

where F is the resultant force (N), V is the resultant linear velocity (m/s), M is the bending moment (Nm) and ω is the angular velocity (rad/s). Equation (3) represents the power obtained in transitional and rotational motions at each intervertebral level. The severity and probability of neck injury in frontal

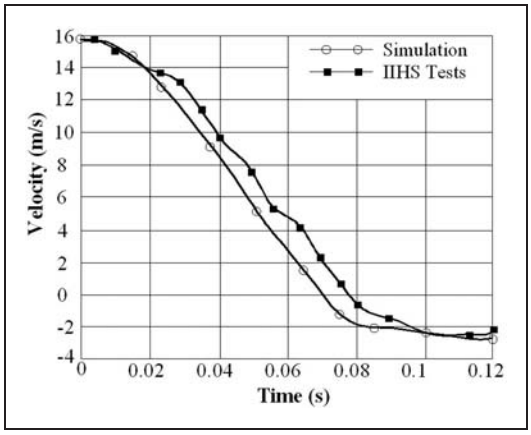

Figure 11. Comparison of vehicle velocity obtained from simulation results and IIHS test data.
IIHS: Insurance Institute for Highway Safety.

Table 5. Test conditions of the NBDL experiments.[64]

Parameter	Unit	Frontal
Reference		Thunnissen et al.[64]
Subject		Volunteer
Total number of subjects		5
Total number of tests		9
Average mass	kg	68.6
Average height	m	1.69
Seat type		Rigid
Max sled pulse	g	15

NBDL: Naval Biodynamics Laboratory.

impacts was estimated based on the injury risk curves developed in Figures 1 to 3.[73]

Analysis of results

Kinematics of occupant

The verified models were used to conduct numerical simulations for several cases corresponding to the real-world vehicle crashes situations. In each of these situations the forces on the neck have been analysed and an attempt has been made to predict the likelihood of injuries and subsequently correlate the same with field data. To capture the motion of the driver head position, a reference frame is chosen whose orientation coincides with the CG location of the head just prior to the impact of the vehicle with the barrier. All positions are then measured from this reference frame.

The kinematics of the occupant was analysed for belted driver at an impact speeds of 27, 80, 120 and 160 km/h. The displacement, velocity and acceleration response for head (CG) and thorax (CG) were observed

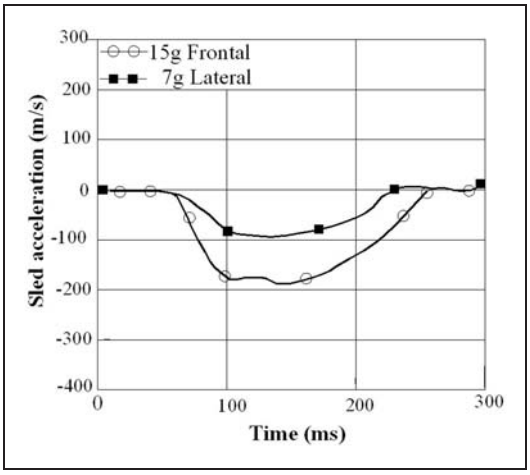

Figure 12. Mean sled acceleration used in NBDL experiments.
NBDL: Naval Biodynamics Laboratory.

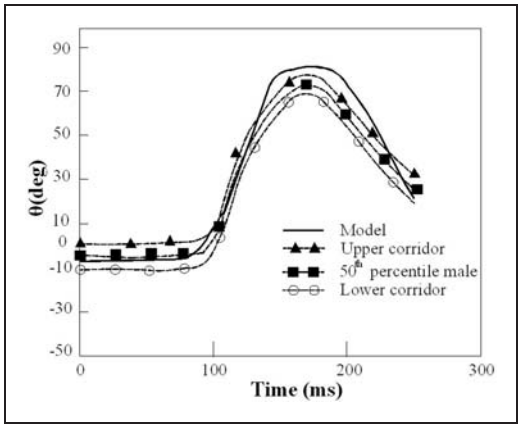

Figure 13. Comparison between the rotation of head CG in the simulations and NBDL experiments at 15 g.
CG: centre of gravity; NBDL: Naval Biodynamics Laboratory.

for belted drivers as shown in Figures 15 and 16. In these figures the highest peaks are observed for impact speeds of 160 km/h. However, the overall shapes of the responses are similar, with higher amplitude at higher speeds. This shows that the head and thorax of the occupant, and consequently the neck, are more likely to get severe injuries at higher speeds.[81]

Kinematics of cervical vertebrae

The kinematics of the cervical vertebrae were analysed for frontal impact for a belted driver at impact speeds of 27, 80, 120 and 160 km/h. For brevity results are presented only for an impact speed of 80 km/h. The pattern is similar at other impact speeds.

Figure 17 shows the kinematic response of the cervical vertebrae. Figure 17(a) shows that almost all cervical vertebrae show a similar displacement trend with

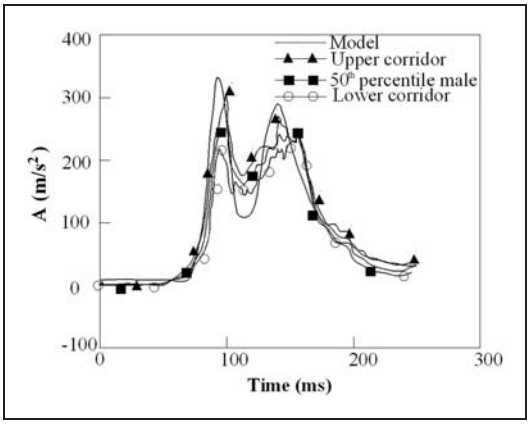

Figure 14. Comparison between the resultant acceleration of head for the simulations and NBDL experiments at 15 g.
NBDL: Naval Biodynamics Laboratory.

almost the same magnitude. During the simulations, the head translates posteriorly and inferiorly with respect to T1 as the spine extends. Around 75 ms time period, the development of the characteristic S-shaped curvature of the cervical spine is observed. The vertebral displacements depict that during this time period the upper levels of the spine (C0–C3) are flexed while the lower levels (C5–T1) are extended. In the 130–160 ms time period, the upper vertebrae of the model change from flexion to extension as the whole model becomes more and more extended into a C-shaped curvature. Maximum extension of the head and neck is reached at approximately 80 ms. In the later stages of trauma the head rebounds almost completely to its initial starting configuration.

Figure 17(b) shows that the velocities of C1–C7 have a sharp dip at 0.06 s, with C1 showing the maximum change in velocity and C7 the least. At around 0.15 s velocities have another dip with C1 and C2 showing the maximum change in velocity and C6 and C7 the least. Between these two time points the velocities vary gradually.

From Figure 17(c), it is seen that the maximum resultant translational acceleration of 50 g is observed at about 50 ms in all vertebrae (C1–C7). From Figure 17(b) and (c), it is seen that when the resultant velocities at vertebrae (C1–C7) take a dip at 0.06 s, the accelerations at these vertebral locations show a maxima whereafter the accelerations also decrease.

At 150 ms, the velcocities againshow a dip, when the accerelations at thes vertebral locations show a second peak.

Intervertebral neck forces

The main advantage of the model constructed is the possibility of predicting the neck forces at different

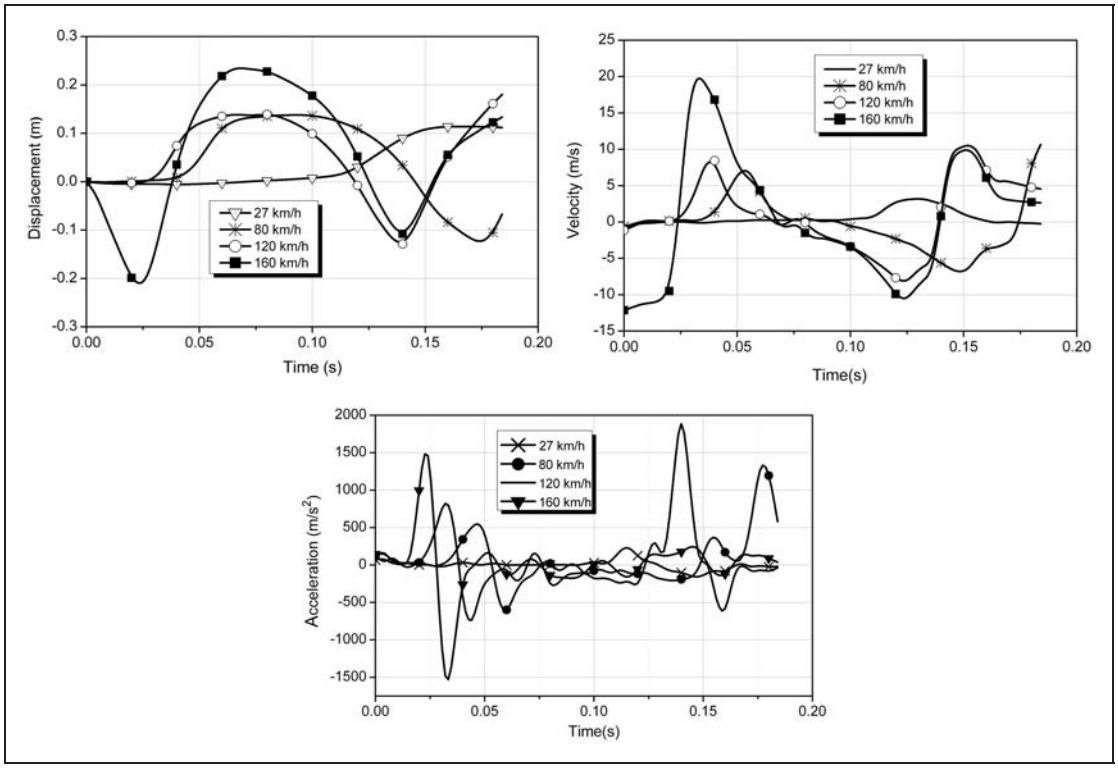

Figure 15. Comparison of displacement, velocity and acceleration of head in frontal impact for belted driver.

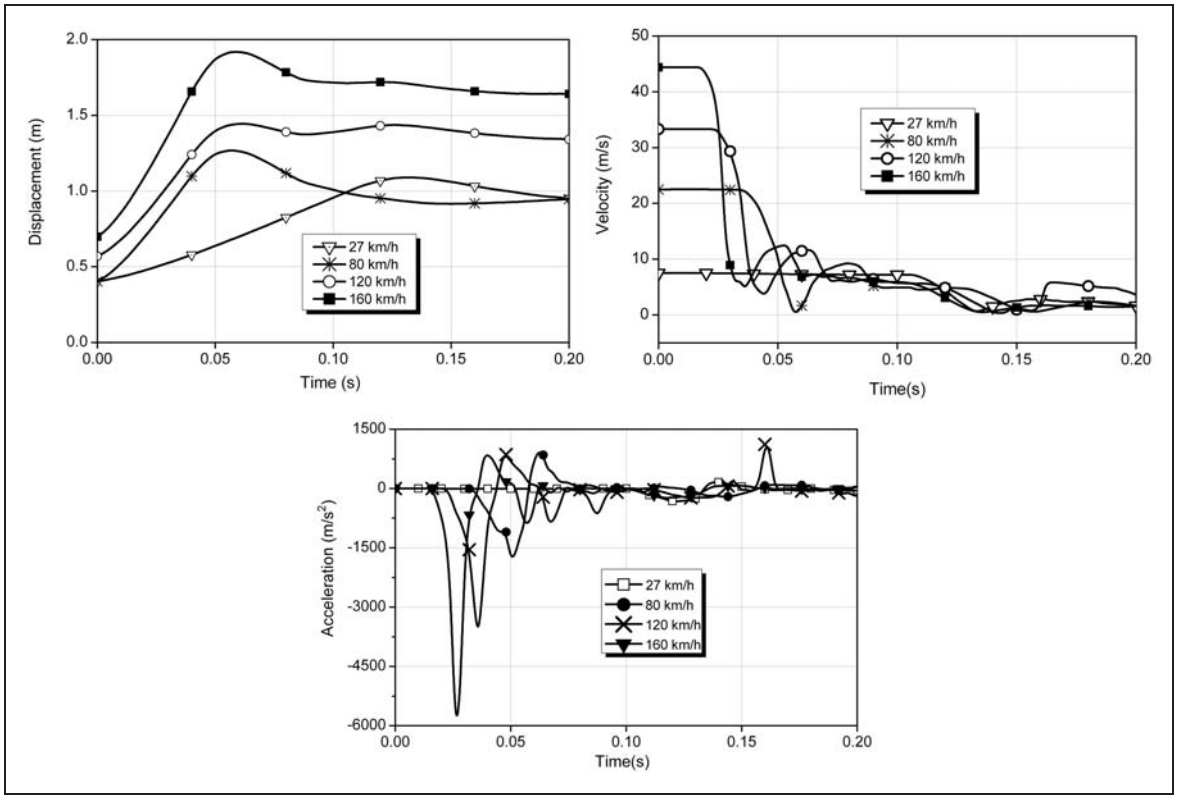

Figure 16. Comparison of displacement, velocity and acceleration of thorax in frontal impact for belted driver.

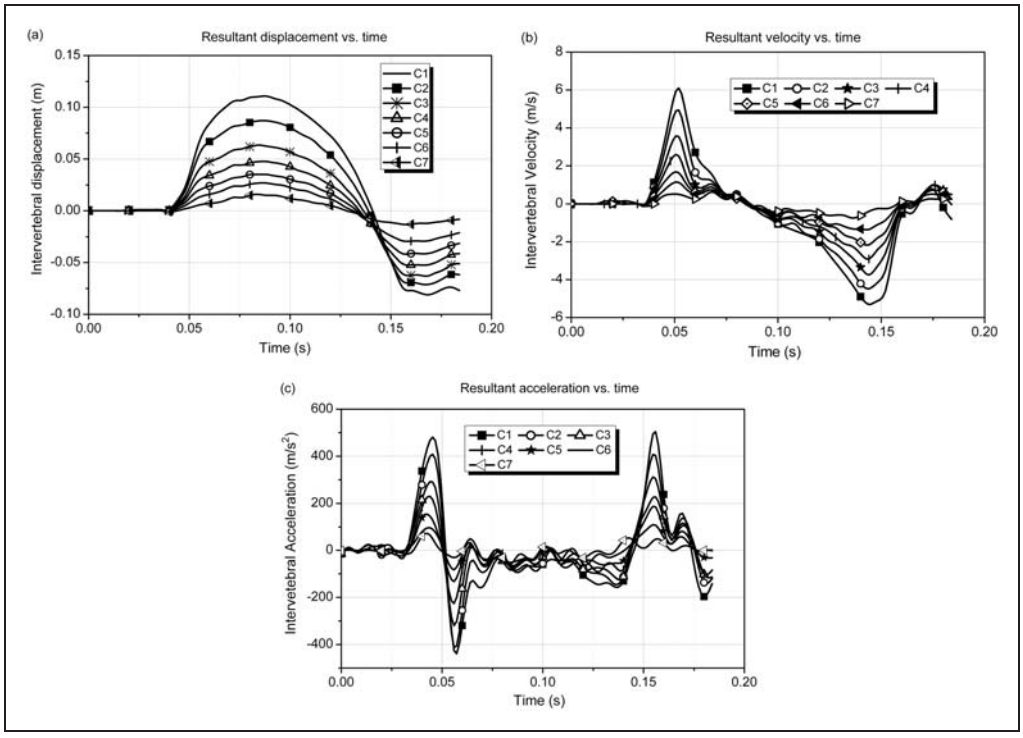

Figure 17. Comparison of the motion at C1–C7 vertebra in 80 km/h frontal impact for belted driver: (a) Resultant displacement, (b) Resultant velocity and (c) Resultant acceleration.

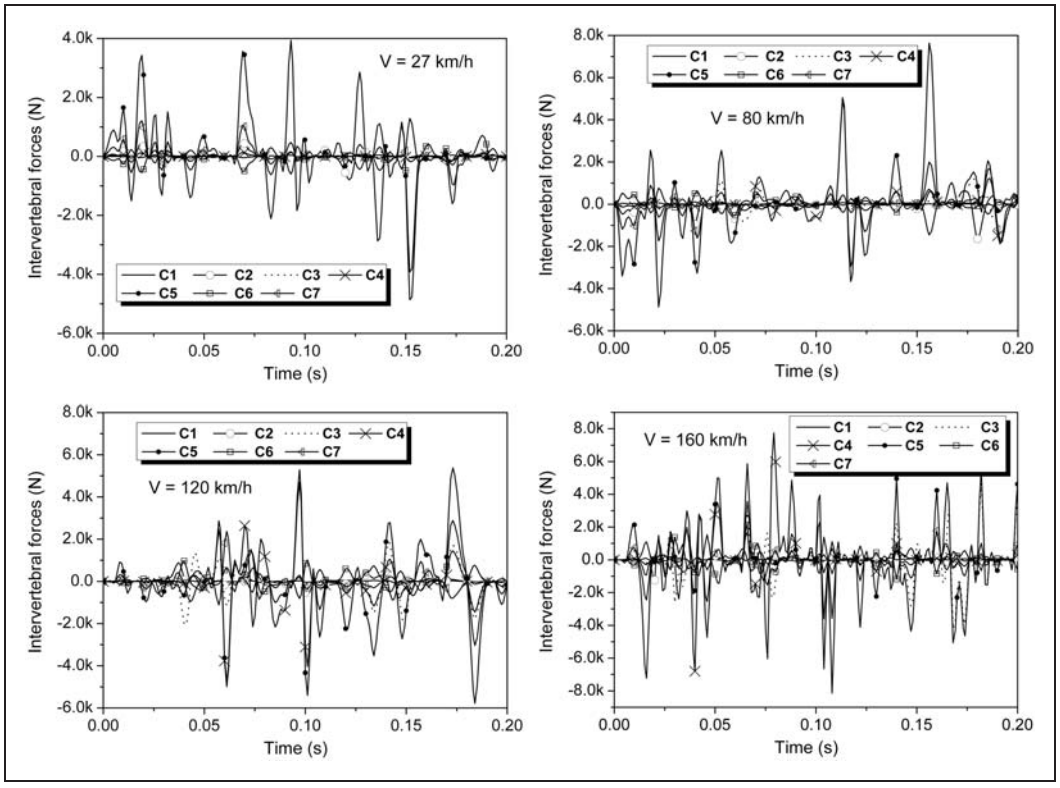

Figure 18. Comparison of dynamical forces of cervical vertebrae of belted driver at impact speeds of 27, 80, 120 and 160 km/h.

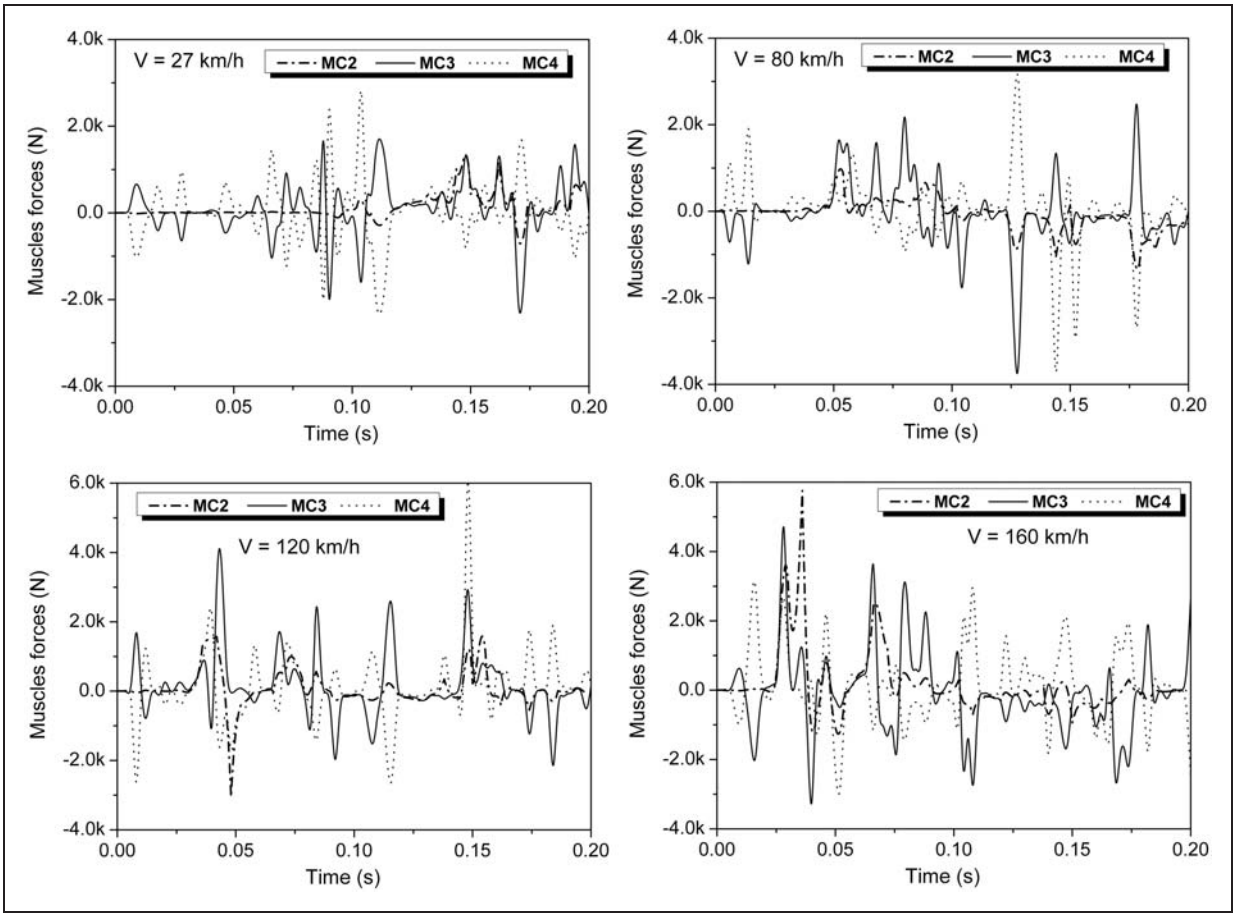

Figure 19. Comparison of tensile forces in back muscles of cervical spine of belted driver at frontal impact speeds of 27, 80, 120 and 160 km/h.

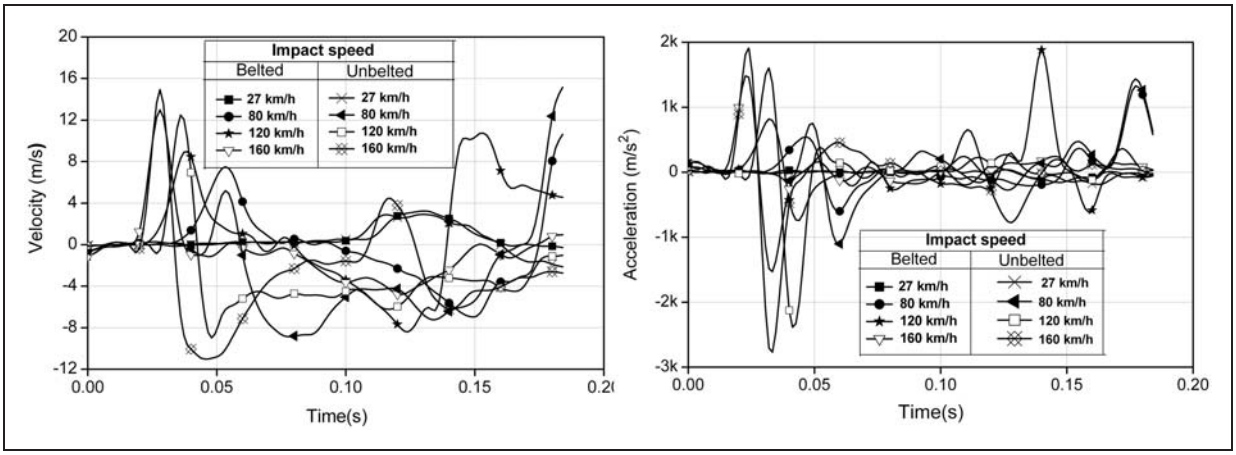

Figure 20. Velocity and acceleration of head CG for belted and unbelted driver during frontal impacts.
CG: centre of gravity.

intervertebral levels. The neck forces were determined at the most dangerous (defined according to injury severity) external positions of the human body during frontal impacts.

Figure 18 shows the dynamic forces in the cervical vertebrae. The forces were recorded for cervical vertebrae of the belted driver at impact speeds of 27, 80, 120 and 160 km/h.

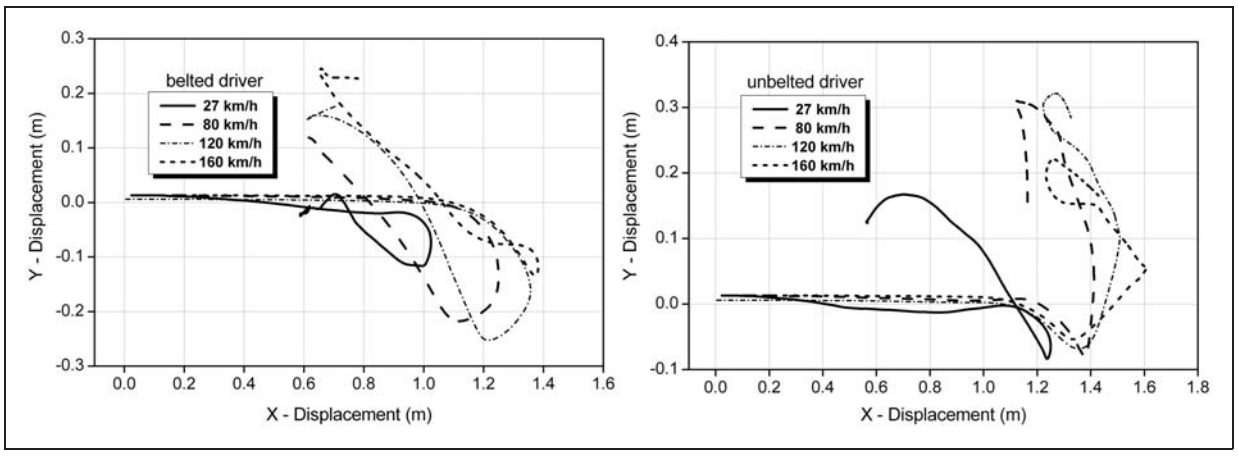

Figure 21. Trajectory of the head for belted and unbelted drivers at 27, 80, 120 and 160 km/h frontal impacts.

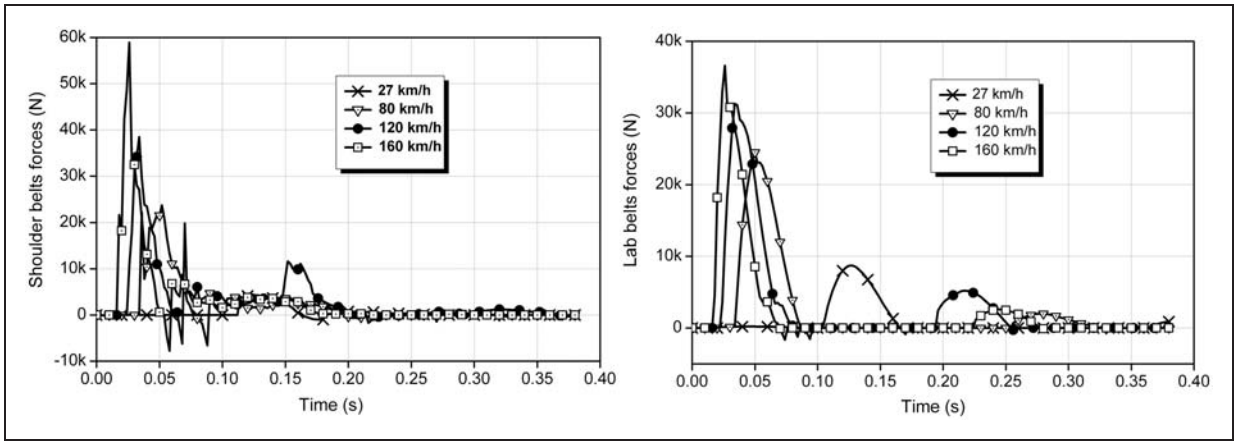

Figure 22. Seatbelt forces for the driver during the simulations of frontal impacts.

These forces play a significant role in the computation of PVP and prediction of neck injuries. Forces which are beyond tissue strength tolerance can cause serious injuries (3600 N is critical force for neck injury adequate to AIS6).[78]

As the maximum bending occurs at C4–C5–C6 intervertebral segments, higher peaks were observed for the same at the various impact speeds. Thus, during frontal crash, maximum force was observed at the lower cervical which is the most likely to sustain neck injuries.

The tensile forces on the back muscles of the cervical spine at impact speeds of 27, 80, 120 and 160 km/h is shown in Figure 19. It is observed that these forces increase during deep flexion of lower vertebra and at higher speeds.

Evaluation of seat belt effect

A quantitative comparison was made for the head CG resultant velocity and acceleration for belted and unbelted drivers at impact speeds of 27, 80, 120 and 160 km/h and presented in Figure 20.

It can be seen from Figure 20 that the highest velocity peak was recorded for unbelted driver at impact speed of 160 km/h. The highest head CG acceleration peak was also noted at the same impact speed for the unbelted driver. Thus, occupants without seat belts would have a higher probability of receiving serious and fatal neck injuries than those wearing seat belts. Hence, as expected, one can conclude that seat belts can provide significant protection in frontal impacts even at high impact speeds.[12,19,82]

Figure 21 shows the trajectory of the driver's head inside the vehicle compartment. It can be seen that the belted driver does not leave the seat and the motion of the head in the forward direction is constrained and the driver does not hit the steering wheel. When the driver is unbelted and has a frontal impact, it is seen that the displacements of the driver's body is larger in the forward horizontal direction and the displacement in the

Table 6. Neck injury criteria results in simulations.

Impact speed (km/h)	N_{ij}	*P* (MAIS $\geqslant$ 3) (%)
27 (belted)	0.626	4
80 (belted)	3.04	34
120 (belted)	4.75	64
160 (belted)	7.44	93
27 (unbelted)	0.67	8
80 (unbelted)	2.5	58
120 (unbelted)	4.44	82
160 (unbelted)	7.67	98

Table 7. Risk of neck injuries in the study.

Impact speed (km/h)	ΔV (km/h)	MAIS
27	18.09	1
80	47.90	2
120	65.18	5
160	88.35	6

vertical direction also becomes significant. It is obvious that seat belts permit the trajectory of occupants to be controlled so that energy-absorbing vehicle components can be effectively utilised.

The seat belts and driver contact forces obtained from the simulations are as shown in Figure 22. The maximum shoulder belt forces are seen to be more than 5 kN, even at impact speeds of 27 km/h. This is higher than the normal limit of the pretensioners (4–5 kN). This result indicates that belt pull-through may have occurred at these speeds.

Risk of neck injury

The risk of neck injury depends upon the change of speed (ΔV) during the impact. As the speed of the vehicle increases, the chance of getting severe neck injuries also increases.[81]

Table 6 shows the N_{ij} values that have been computed for the different cases, and the corresponding probabilities of MAIS $\geqslant$ 3 injuries in the neck. While these values are slightly higher than those based on ΔV, the analyses based either on ΔV or on N_{ij} does not predict likelihood of injuries in each vertebral region, which can be obtained from the PVP at each intervertebral level.

The current simulations also reinforce the fact that the risk of neck injuries in frontal crashes is greater at higher speeds. Also, seat belts are seen to be effective in reducing the risk of neck injury. Table 7 shows the change in velocity (ΔV) computed from the simulations, and corresponding expected MAIS values of neck injuries. Table 7 corroborates that the severity likelihood of neck injuries increases with speed.

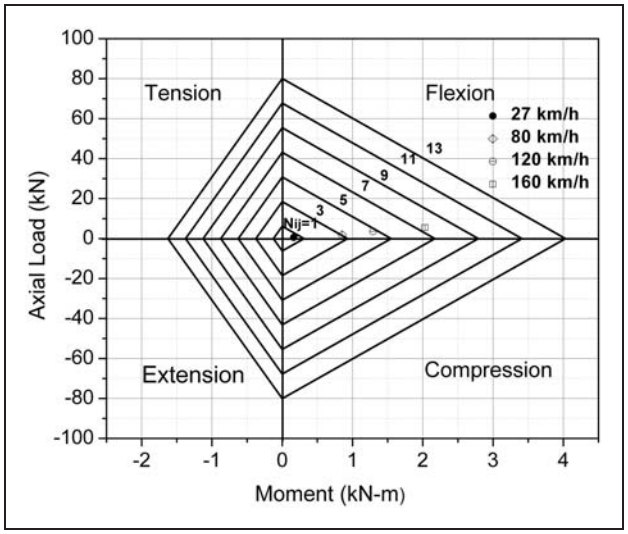

Figure 23. Graphical representation of N_{ij} for simulated frontal impacts.

Also, the results of the N_{ij} from the simulations of the frontal impacts are presented graphically in Figure 23.

Figure 24 shows the PVPn calculations for frontal impacts at various impact speeds using the typical passenger car model in this study. The results are only presented for the cervical spine, since that is the most important region considered in the whiplash injuries and neck injury criteria. In this figure, the first four graphs show force, linear velocity, bending moment and angular velocity of each vertebra. The fifth graph shows the PVPn calculated. The last graph shows the incidence of neck injuries at each intervertebral level obtained from field survey of real world crashes in the same orientation.[73]

A comparison of the last two graphs in Figure 24 gives a quantitative comparison between the PVPn and the injury likelihood. From the comparison it can be seen that whenever the PVPn is high, the neck injuries increase.

The statistical analysis shows that there is a highly significant correlation ($p > 0.2$) between the PVPn and the likelihood of neck injury at a particular intervertebral level for the different impacts in the study.

Figure 25 shows the PVPn and incidences of spinal injury for all the vertebra during frontal impacts. It is observed that the cervical region is the most frequently injured region as indicated in many studies.[1,4,6,14,16,73] It can also be seen that the distribution of the PVPn and that of the frequency of injury is similar. This further confirms the correlation between the PVPn and site as well as the incidence of neck injury.

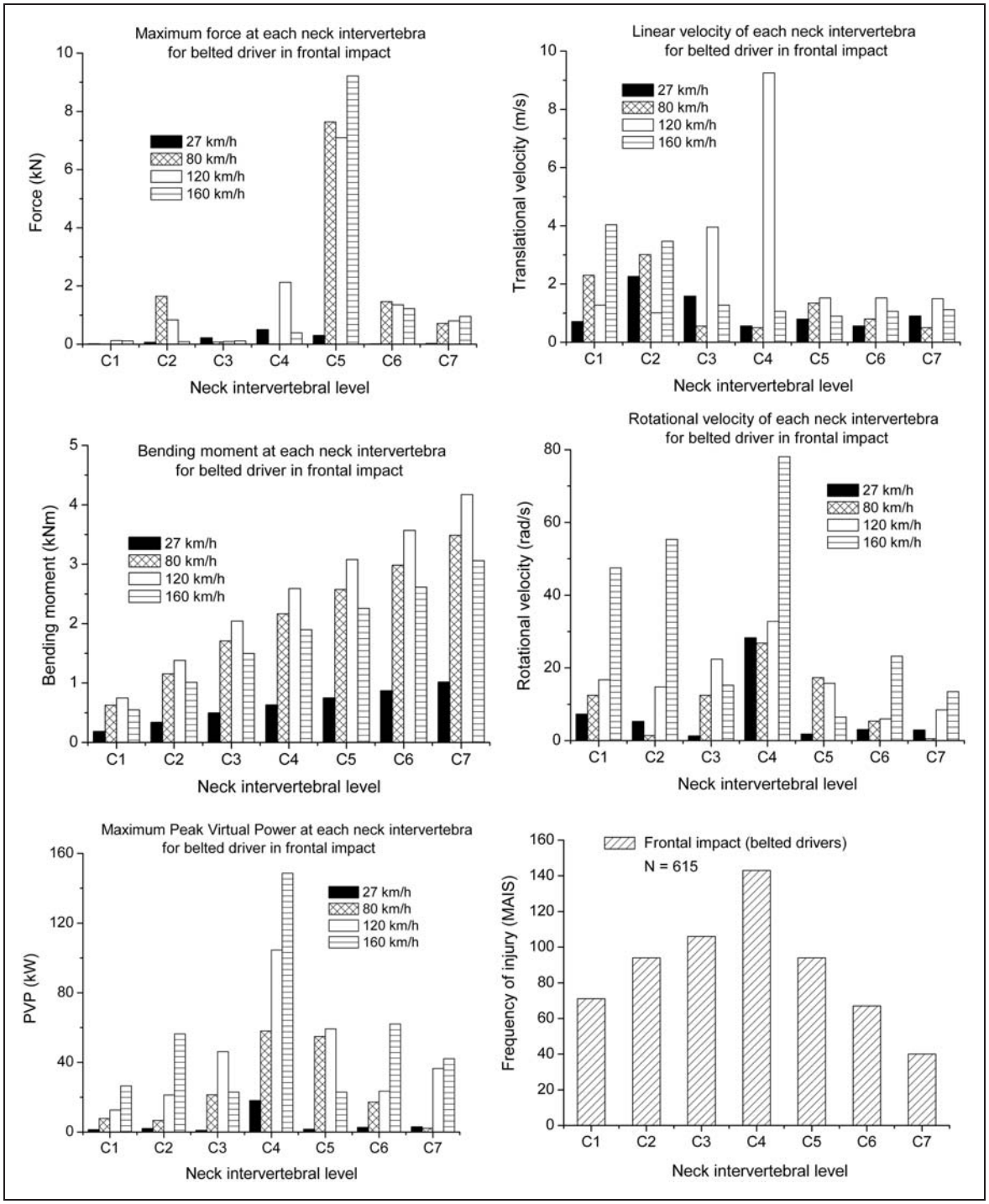

Figure 24. PVPn for belted driver in frontal impacts.
PVPn: Principle virtual power of neck.

The correlation between N_{ij} and PVPn is shown in Figure 26 for belted driver. The results are very close for both neck criteria. However, N_{ij} cannot predict the distribution of injury along the spinal column, as is done with PVPn. This is considered to be a major advantage of using PVPn.

Discussion and conclusions

An analysis of occupant kinematics, and the risk of neck injuries during frontal impacts has been presented. For the vehicle occupant, comparison of injury indices for belted and unbelted occupants indicates that the use

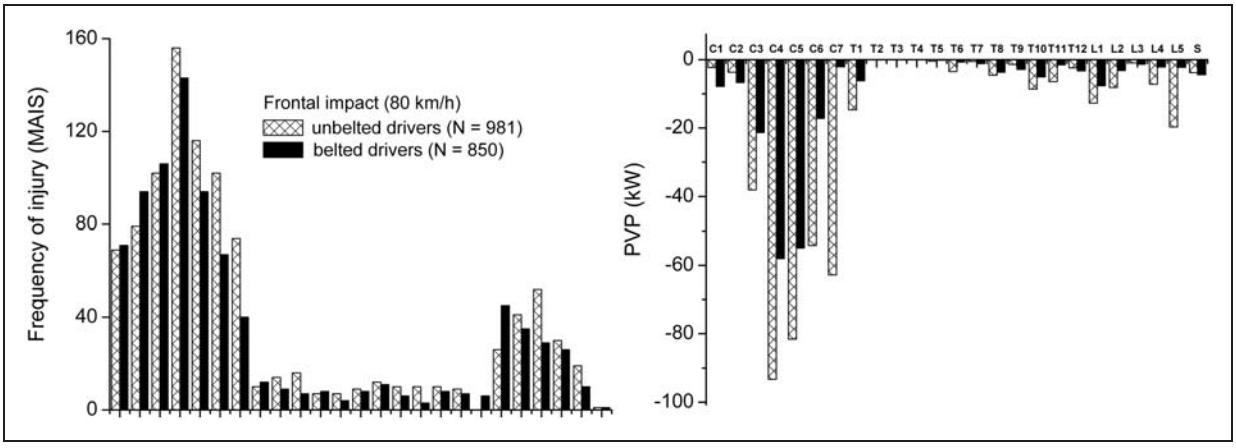

Figure 25. Distribution vertebral injuries of crash data vs PVPn from simulations for belted and unbelted drives at 80 km/h frontal impact.
PVPn: Principle virtual power of neck.

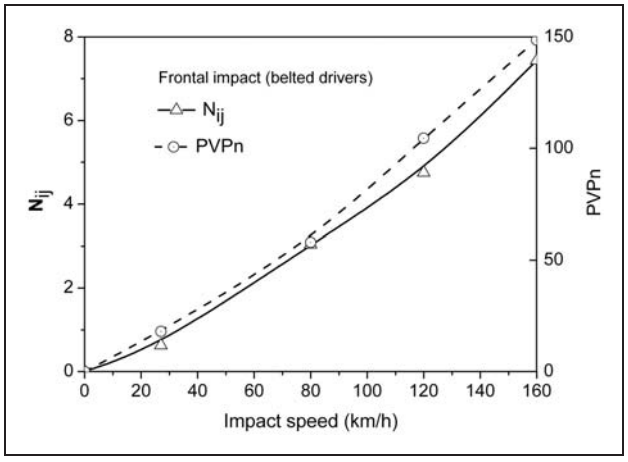

Figure 26. The Correlation between N_{ij} and PVPn for belted drivers in frontal impacts.
PVPn: Principle virtual power of neck.

of the seat belt is effective in reducing the injury likelihood in most cases. The injury indices are also observed to increase with speed. Several studies have shown that the neck injury risk is associated with seatbelt use.[7–12]

The results presented in this paper also show that the greatest forces occur on the lower cervical levels, giving a greater risk of disk and ligamentous injuries at the C4, C5, C6 levels in frontal crashes. The greatest forces appear at the moment when the head is in a deep flexion position. For frontal collision, the forces on the lower levels of the cervical spine are larger than those in other cervical vertebrae. The same has also been observed in previous studies for occupants injured during low velocity accidents.[83,84] The previous biomechanical and clinical studies have suggested that during frontal impact, the predominant mode of intervertebral hyperflexion may be the injury mechanism producing disk and ligamentous injuries.[13,14]

PVPn has also been computed at each intervetbral level. It is observed that PVPn at each intervertebral level correlates well with the likelihood of injuries in frontal crashes. This indicates that PVPn could be a good measure to predict injuries at specific intervertebral levels. Results also show that in a frontal impact at a velocity above 100 km/h the passenger's chances to survive decrease rapidly, due to high accelerations, forces and PVPn. These results are also in consonance with crash trends reported in literature.[85]

This study has presented how simulations can be used to understand the occurrence of neck injuries in frontal crashes. It can be seen that these computational simulations of crashes provide a very powerful tool for understanding the dynamics of crashes and the risk of neck injuries.

Simulations in this work have been done in a 2D environment. Although, the 2D models provide a good indication for the use of PVPn as an injury predictor, this work can further be extended to 3D models. Detailed 3D finite element models would in principle be able to better predict the outcome in the crash. Further, while the computational models developed have been validated using experimental data available in literature, it must be pointed out that in the available experimental data the details of the tests are often not available. Experimental data is also available only at low speeds. This has made the validation quite restrictive.

The current work has demonstrated the capability of the software tools in understanding the crashes as well as the kinematics and injury mechanisms in frontal crashes.

Some of the main limitations highlighted in the development of these simulation models are the availability of vehicle data, availability of human dummy models, and availability of data for validation and estimation of injury indices in an attempt to estimate the likelihood of injuries. These simulations can however be strengthened to study the injury mechanisms of other body parts as well as other safety issues.

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Appendix

Notation

C_i	cervical vertebra ($i = 1, \ldots, 7$)
Delta-V, ΔV	change in velocity
F	resultant force
F_Y	axial force in y-direction
F_{YC}	critical axial load of neck tension and compression
g	acceleration of gravity ($9.81\,m/s^2$)
I_x, I_y, I_z	mass moment of inertia about x, y, z axis
L_i	lumbar vertebra ($i = 1, \ldots, 5$)
M	bending moment
M_{OCZ}	sagittal bending moment at the occipital condyle
M_{ZC}	critical neck flexion moment at the occipital condyle
N_{ij}	normalised neck injury criterion
P	probability at 95% significance level

R, R^2	correlation coefficient of regression
T_i	thoracic vertebra ($i = 1, \ldots, 12$)
V	resultant linear velocity
ω	angular velocity

Abbreviations

AIS	Abbreviated Injury Scale
CCIS	Co-operative Crash Injury Study
CG	centre of gravity
EBS	equivalent barrier speed
EES	equivalent energy speed
FEM	finite element method
IIHS	Insurance Institute for Highway Safety
MAIS	Maximum Abbreviated Injury Scale
MC	muscles connecting neck and thorax
MDB	multibody dynamics
NBDL	Naval Biodynamics Laboratory
NIC	neck injury criteria
PDoF	principal direction of force
PMHS	post-mortem human subject
PVP	peak virtual power