Biomechanical analysis of whiplash injuries; women are not scaled down men

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ABSTRACT

Whiplash is the most common soft tissue injury sustained in car accidents. The term is commonly associated with hyperextension of the neck as the head rotates backwards in rear end collisions but the exact injury mechanism is not fully understood because the neck is an anatomically and mechanically complex structure. Experimental studies of the mechanism of injury are limited by several ethical and practical factors, so biomechanical computational simulation, based upon experimental research and mathematical modelling, appears to be the most appropriate method of investigation.

During the last decade, significant progress has been made in improving car occupant safety through the use of safety devices, such as airbags and advanced seat belts, as well as the construction of the car body itself. Much still needs to be done, especially for female occupants, because statistically they incur twice the risk of whiplash injury as male car occupants. No simple explanation has so far been found for this difference. It is thought that the anatomic dissimilarity of the sexes is the principal reason, but there are undoubtedly a number of secondary, sociological reasons: women tend to drive smaller cars than men and are more likely to be passengers. The lack of a full explanation arises from the fact that, although there have been several FE-models of the male cervical spine reported, female models are rarely documented.

This paper addresses the problem by developing a biomechanical FEM model of the 50th and the 5th percentile female cervical spines, based on the earlier published male model created at the Nottingham Trent University, which relies on grafting a detailed biomechanical model of the neck and head onto a standard HYBRID III dummy model. All numerical analyses have been undertaken using LS-DYNA. Special attention was paid to the behaviour of the scaled down male model in comparison with the model, which included female characteristic features. FEM models of males and females in a representative seat were therefore subjected to 9.5 km/h rear-end simulated collisions and were compared against reported experimental tests. The detailed behaviour varied significantly with gender. The female models revealed greater and earlier peak horizontal acceleration of the head and smaller peak relative extension than the male models. It was concluded that the presented FE models were reasonably in accordance with available crash data on instrumented volunteers in terms of head motion.

The results confirmed that females couldn't be modelled as scaled-down males, thus underlining a need for separate male and female biomechanical models. Further investigation is required to quantify the gender differences, and then recommendations can be made for changes to the design of car seats and head restraints in order to reduce the risk of soft tissue injury to women. The findings of this study suggest that a revision of car test programmes and regulations, which are currently based on the average male, would be beneficial to women.

INTRODUCTION

1. WHIPLASH INJURY MECHANISM

Whiplash-associated soft tissue neck injuries are one of the most common injuries reported from automotive rear-end impacts. Although classified as minor (AIS=1), their high incidence rate and long-term consequence lead to significant social cost [1]. The annual cost in the UK is estimated at £2.5 billion [2]. Symptoms include neck pain, stiffness, headaches, dizziness, blurred vision and numbness and may be associated with damage to the cervical muscles, ligaments, facet joints, nerve roots, vertebral arteries, or brain stem. However, despite numerous studies on human volunteers, cadavers, and animals, there is no consensus about specific mechanisms responsible for the majority of neck injuries to car occupants in rear-end impacts, although several have been proposed. A list of injury types and mechanisms relevant to rear impact are presented in Table 1

Mechanism	Theory	Injury	Parameters	Female response compared to male during volunteers test
Cervical hyperextension [3]	Head inertia causes cervical extension as torso is pushed forward by seat back, resulting in excessive stress on neck	Central nervous system injury, intervertebral disc herniation, rupture of ligaments	Head rearward rotation, head angular acceleration relative to torso, neck rearward bending moment	Higher and earlier peak head acceleration [4][5]
Cervical flexion during rebound motions [6][7]	Rebound of the occupant out of the seat results in neck flexion.	Posterior tissue stresses, anterior compressed	Head linear acceleration and velocity relative to torso during rebound	More rebound motions, larger maximum thorax flexion [5]

Shear (retraction), Shear-compression [8][9][10]	Torso pushed forward by seatback but head stays motionless due to inertia resulting in rearward translation of head relative to torso, causing shear stress on cervical spine structures. Additional axial compression due to ramping up reduces shear stiffness of neck.	Ligaments, facet joint capsule	Head rearward displacement relative to torso, torso linear acceleration relative to head.	20% smaller head retraction relative to C7-T1 [11], higher horizontal acceleration relative to earth [4,5]
Localized Cervical Compression and Tension [11] [12] [13]	Formation of S-curvature causes compression in posterior elements of lower neck, tensile forces in facets and contraction of tissue of upper neck.	Facet joint injury, ALL ligament, intervertebral disc	Neck axial forces, neck bending moments, head acc in z-direction, torso linear acc., C5 to C6 rotation.	Higher T1 horizontal acceleration relative to earth [4,5]
Spinal canal pressure gradients [14]	Growing pressure gradient in spinal canal during reward translation of head relative to torso	Dorsal root, ganglia	Head linear acceleration and velocity relative to upper torso	Higher and earlier peak head acceleration relative to earth [4]
Muscle strain [15]	Para cervical muscle contraction after impact may result in cervical excessive strain	Muscle injury due to lengthening contraction	Head linear acceleration relative to torso, during acceleration and rebound motion, angular acceleration, neck shear forces	Higher and earlier peak head acceleration relative to earth [4]

Table 1 Whiplash mechanism

The typical load scenario in a rear-end collision can be is as follows:

1. The vehicle accelerates forward when struck
2. The torso is pushed forward by the seat
3. The spine starts straightening and the neck/torso joint rises
4. The head lags behind the torso due to its inertia
5. The upper cervical spine undergoes flexion while the lower part undergoes extension, promoting an S-shape
6. The rise of first thoracic vertebra, in (2) above, leads to a ”ramping phenomenon ” which causes cervical compression
7. The head rotates backward, producing a C-shape with extension of the entire cervical spine. Presence of a head restraint reduces the C-shape
8. The occupant rebounds out of the seat, leading to flexion of the cervical spine.

2. RISK OF INJURY DURING LOW SPEED REAR-END ACCIDENTS

Study of crash and insurance data shows a 1.5-2 times higher risk of neck soft tissue injury for female occupants than for men, not only in rear-end impacts but also front or lateral impacts [16]. Females more often suffer distortion and soft tissue bleeding (joint capsules, ligaments, muscles)[17] and sustain more often (+44%) long-term consequence [18][19]. The reasons for this are not clear to date. Some attempts have been made to attribute the gender difference in injury risk to anatomical, physiological, behavioural, and sociological parameters.

2.1 SOCIOLOGICAL FACTORS

Women tend to driver smaller, lighter cars than men and this situation is disadvantageous since the car mass is a key factor that determines injury outcome. However, Koch et al [20] reported that the relative risk of injury in smaller struck cars was still higher for females than males, even when the female was the driver. Otte et al [21] also suggested difference in sex-specific accident framework conditions and confirmed that women suffer neck injury in small cars more frequently then men. In the UK, where medium cars are driven most frequently by both sexes, 42% of female driver collisions are in small cars compared to 23% for males [22]. Furthermore men have lower disability levels than women despite having on average less optimal head restraint positions [23]. It was suggested that females tend to sit farther forward in their seats than males so their heads move farther before the headrest is reached [24]. Seating position also can affect spinal kinematics and increase the risk of injury. Matsumoto et al [25] showed that the percentage of kyphosis position is much higher for females than males. Spine misalignment as a reason for soft tissue injury for women was pointed out by Ono et al [11], who showed that rotational angles of cervical vertebrae were larger at kyphosis for females, producing higher probability of injury.

2.2 Psychological factor

A completely different hypothesis was suggested by Spitzer et al [26] on the basis of clinical experience; women are likelier report pain and disability cased by the injury. However, no data has been presented by insurance companies to confirm this.

2.3 PHYSIOLOGICAL AND ANATOMICAL FACTORS

Physiological and anatomical differences imply that the biomechanical tolerance of the female neck is lower than males and may explain differences in neck injury frequency.

Weight and height

Temming et al [27] indicated that the risk of whiplash injury for both females and males increases with body height but females have higher risk of injury. Also injury risk is higher for females in each weight group [28], disproving the hypothesis of Kraft et al [29] that women are more vulnerable to soft neck injuries because they are generally lighter.

Vertebrae

Significant gender differences were noted for depths of the superior and inferior endplates of the cervical vertebrae, with those for males being larger [30,31]. Detailed studies on the C5 (fifth cervical vertebra) showed significant differences in bone mineral density by gender with females demonstrating a 13% decrease in area–density product, giving a greater risk of compressive injury, particularly facet joint injury [32]. Injuries of this type are certainly more common in the female lumbar spine.

Muscle

Differences in neck musculature between men and women are suggested as an important factor in neck injuries [33]. Cervical muscles can be sources of pain and influence neck motion, both passively and actively. Statistically females have smaller neck circumferences, suggesting this may be the actual risk area. Furthermore, most muscles in women have lower cross section than those in the men [34]. States [3] attributed the differences in injury risk to the ratio of head volume to cross sectional area of necks. For 50th percentile males the ratio is 1:135 and for the comparable female it is 1:151, indicating females have narrower necks relative to head size. Male neck muscles are also stronger than female cervical muscles; the female strengths were 30 - 40% lower than their male counterparts [35] or according to others 20-25% lower [36]. According to Vasavada [37] males have 2-2.5 times greater moment-generating muscle capacities and only 1.1-1.3 greater mass and head inertia relative to women, suggesting female muscles work closer to maximum capacity. Muscle activation occurs 5% -15% earlier [38] [39] for females than males, which may be another source of higher risk for females. As females tend to have smaller and weaker supporting muscles in the cervical spine and also less body weight to collapse back support it can make them more vulnerable to neck injury [40].

Ligaments

Surprisingly, there are no comprehensive data describing differences between female and male ligaments in terms of geometry (cross area, length) and material properties (Young's modulus, load/deformation); this is a major shortcoming for any biomechanical analysis.

MODEL

Only a few attempts have been made to examine the biomechanical response of female cervical spines during car accidents. There is no 50th percentile female ATD (Anthropomorphic Test Dummy) or FEM dummy model in common use. The population of female drivers and occupants is represented by the 50th percentile male dummy in conjunction with 5th percentile female dummy, even though it was shown by Welsh [41] that 90% of female drivers in the UK are lighter and shorter than the 50th percentile male dummy. Table 2 indicates how poorly women are represented when designing safety systems.

	Mass [kg] [42]	Stature [cm] [42]	Head Mass [kg] [43]
5th percentile Female	45	151.44	2.93
50th percentile Female	66.7	162.00	3.64
50th percentile Male	79.75	175.51	4.44
5th percentile female ATD	50	152	3,7
5th percentile male ATD	71,2	165	?
50th percentile male ATD	77	178	4,54

Table 2 Basic measurements

Because the cervical spine is a complex biomechanical system, the finite-element method seems well suited for parametric analytical study. FEM offers the advantage that it can handle complex geometric configurations and material, contact and geometric nonlinearities. LSTC offers Hybrid III 5th, 50th and 95th dummy models. The 5th and 95th percentile dummy models are scaled versions of the 50th percentile dummy model. There is also a 5th percentile female dummy model available from FTSS (First Technology Safety Systems). This study deals with the relative head and neck motion in whiplash, focusing on differences between female and male models and aiming to explain the higher incidence of injury among women. The biomechanical responses from a 50th percentile male dummy and a simple scaled-down 50th percentile male dummy were compared against 50th and 5th percentile female models. The principal comparison is between a 50th percentile female model and a 93%scaled-down male.

The basic 50th percentile male neck model was created by the Biomechanics Group at Nottingham Trent University [44] and consists of a biomechanical head-neck complex combined with the rigid Hybrid III dummy model in a simplified vehicle seat environment. Bony structures are modelled using shell elements with the geometry modified to achieve better interaction with soft tissue. All ligaments are represented, using a mixed structure of shell and non-linear springs elements, except for the Nuchal Ligament which is modelled with shell elements only. The force/deformation load curves for discrete element are based on experimental results [45]. Shell element stiffness properties were calculated from 1% of the breaking force and corresponding deflection. Ligament geometry is based on experimental available data. Muscles are modelled by spring elements, as only passive action is represented, with material properties based on sternocleidomastoid muscles [46]. Intervertebral discs are represented using solids elements of Blatz-Ko rubber.

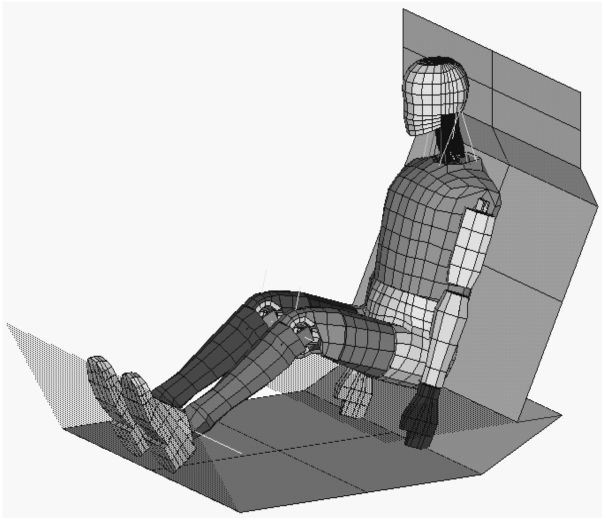

Figure 1 Isometric view of the FEA Model

The first approach in this study was structural scaling of the male model, defined as overall pure size reduction of the male spine without incorporating the characteristic female features. The model was scaled to 93%, assuming that 50th percentile females are 93% as tall as males.

The second model female was more sophisticated; it was again based on a 93% scaled male but allowed for a disproportionately larger female head mass (the 50th percentile female/male body mass ratio is 80% whereas the corresponding head mass ratio is 82%[43]). Strength properties of ligaments and muscle were modified assuming constant Young's modulus but reduced cross-sectional areas due to scaling. It was also taken into consideration that female vertebrae are more slender [31].

		Vertebra height (anterior height divided by depth)	
	Female	**Male**	**Female-Male ratio**
C3	0,935	0,911	**1,026344676**
C4	0,883	0,844	**1,046208531**
C5	0,845	0,812	**1,040640394**
C6	0,834	0,846	**0,985815603**
C7	0,909	0,923	**0,984832069**

Table 3 Anatomic parameters of vertebral height of the cervical spine

The impact loading data for the present study was based on sled experiments with a standard car seat mounted on a trolley, accelerated to simulate rear-end impact at ΔV = 9.5 km/h [5].

VALIDATION

Experiments made by Kronenberg at el [5], were used to evaluate the models. Linear acceleration of the head and the first thoracic vertebra (T1) were obtained. Head angle and trajectories were filmed. Data were taken from a subject whose measurements matched closely the mass and seating height of 50th percentile UK females and males.

The marked increase in head x-acceleration and differences in head-neck kinematics observed for females compared to males in the experiments was confirmed by the computational models. The peak head acceleration is higher and earlier for females than males. However, the acceleration is 10 times higher than experiment because in the FEM model an unrealistic rigid seat model is used.

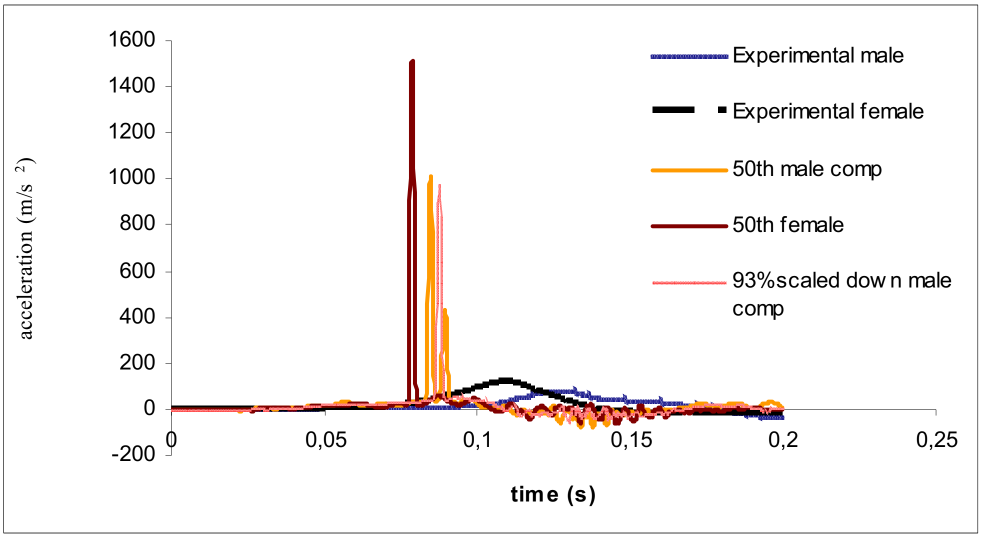

Figure 2 Head horizontal acceleration

Reasonably good agreement was found for head rotation. In sled test carried out by Siegmund et al [4] the females experienced smaller and earlier peak head extension than males. The FEM models confirmed this even though peak values were always higher than experiment. In this study only a passive muscle response is modelled. This seems to suggest that muscle contraction played significant role in cervical spine kinematics, although the muscle onset is developed 80-90 ms after impact [47] and full muscle forces are not developed until 60-70 ms later [39]. It also may explain later peak head extension for women, particularly that they activated their muscles earlier.

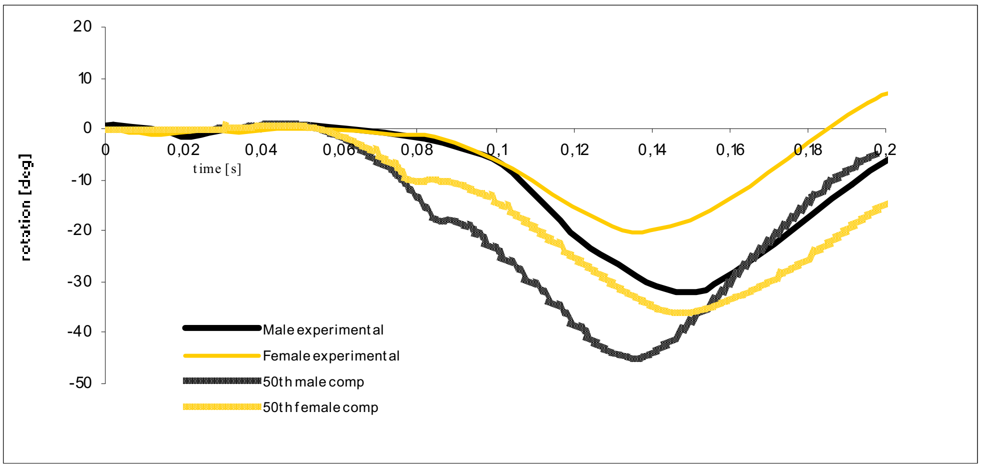

Figure 3 Head rotation relative to earth.

RESULTS

The biomechanical model of the female cervical spine is intended to solve the mystery of higher risk of injury for females. The simplified scaled down male model shows similarity in several parameters to male model rather than male (Figure 1). The relative rotation between the head and C3 produces hyperflexion that is considered a potential neck injury mechanism. The flexion is higher for the female models, both the 50th and 93% scaled down. The curve for the 93%scaled down model in the first second after impact shows the same shape as the 50th percentile male.

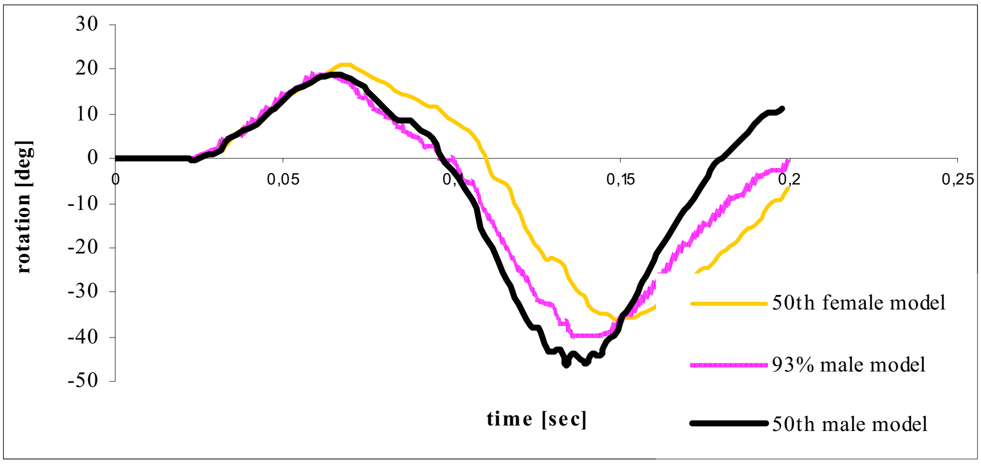

Figure 4 Head rotation relative to C3

Ono [11] found that during S-shape formation the lower vertebrae (C6,C5,C4) are extended and rotated earlier than upper vertebral segments, beyond the normal physiological range. The rotational angle between the fifth and sixth vertebrae is the largest. Such non-physiological motions were attributed to the mechanism of facet joint injury.

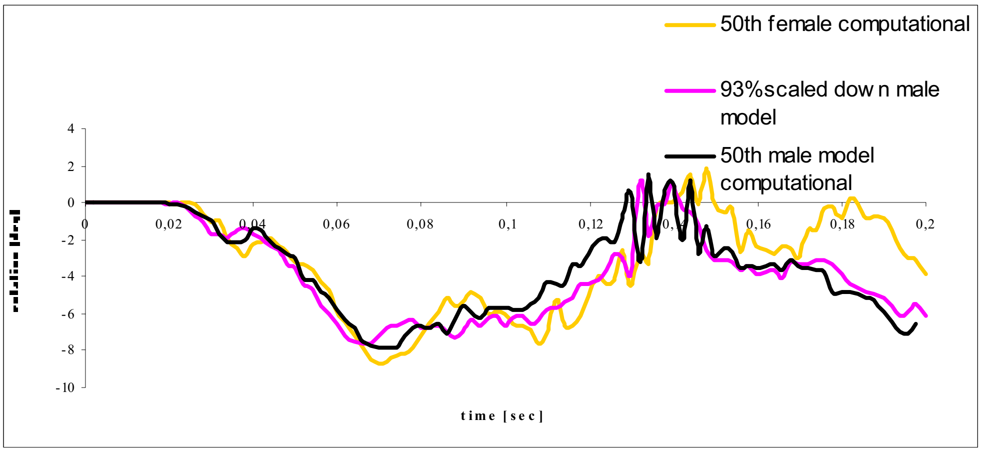

Figure 5 C5 rotation relative to C6

Hence, it could be hypothesized that females are at higher risk of this injury because of higher C5 rotation relative to C6. Yoganandan et al [13] suggested that during whiplash loading the lower facet joint undergoes dissimilar compression combined with anteriorposterior sliding of the facet joint, resulting in a pinching mechanism.

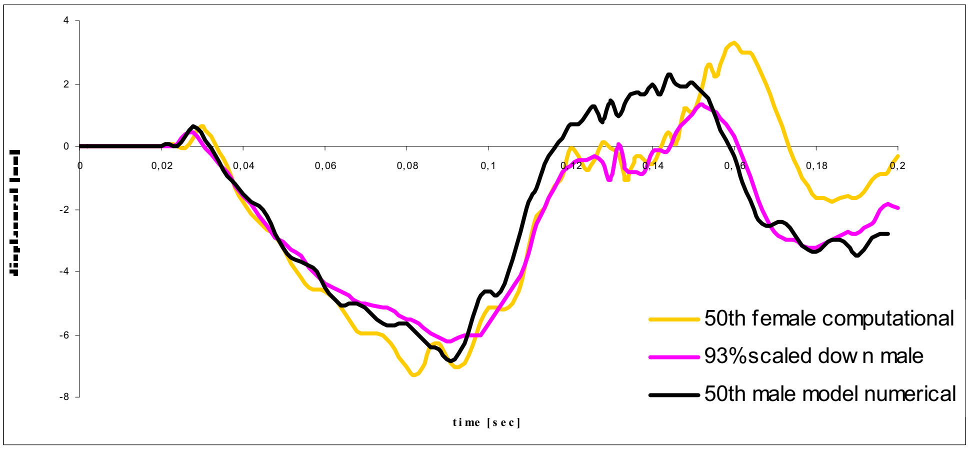

Figure 6 C5 horizontal displacement relative to C6

From Figure 8 and 9 it is observed that during the first phase the C5 vertebra slides backwards. At the same time vertical translation is observed. At peak extension C5 is shifted downward, which may produce compression in the posterior part of the vertebra and stretch the anterior tissue. Moreover, these relative vertical and horizontal motions are higher for female models, suggesting it might be related to a higher injury occurrence.

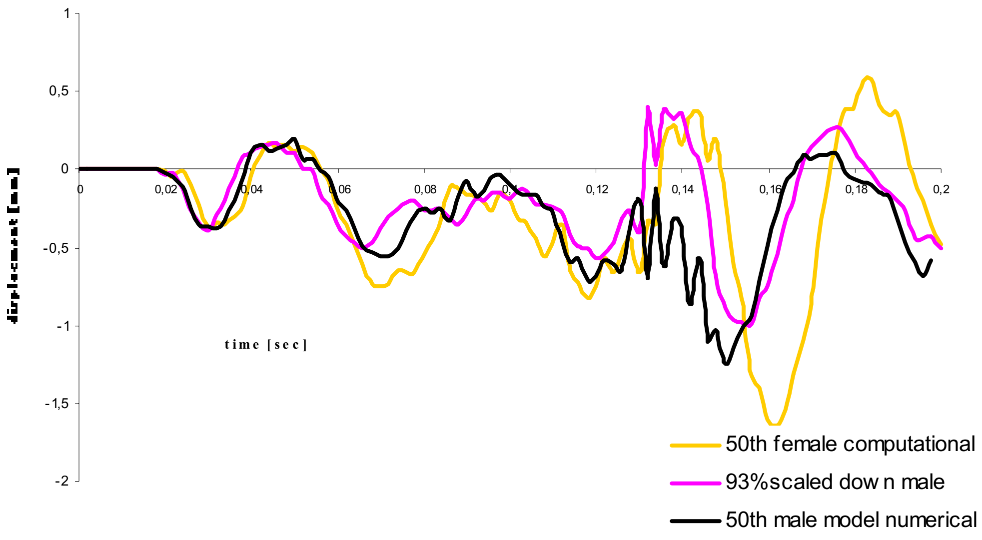

Figure 7 C5 vertical displacement relative to C6

It should be noted that for female models there is higher downward translation during rebound motion, followed by upward shift. The models show significant gender difference in vertebral motions that should not be neglected.

CONCLUSION

Female neck biomechanics is a complex issue, exacerbated by a distinct lack of biomechanical data. The exact mechanism of so-called whiplash injury is not established and there are several hypotheses about the source of pain. In spite of the observed higher risk of injury for female car occupants most research has involved male subjects or gender differences were not specified. The 50th and 5th percentile male dummies, both ADT and FEM models, do not represent the average female. The 93% scaled down male model is not adequate to simulate female responses even though the scaling constitutes a good height and mass match. The 50th percentile female model was in general agreement with test results considering the lack of data about female neck biomechanical properties. This preliminary female model exhibited satisfactory correlation with experimental results and the gender differences in kinematics prove the need for a 50th percentile female model. The observed difference in head rotation relative to C3 and C5-C6 relative motion could be potential causes of higher neck injury for female and need further consideration.
Further model developments are needed in the following areas:

1. Enhancement of muscle response by modelling active response using LS-Dyna code *MAT_SPRING_MUSCLE.
2. Remodelling vertebra geometry to incorporate gender differences in height and cross section area.
3. Better car seat representation.
4. Kyphosis evaluation.

More research should be performed to understand soft neck injuries, mechanisms and thresholds. Standardised calculation of risk for injury would enable comparison. Further study should be performed to evaluate gender differences in biomechanical head and neck response during whiplash.

REFERENCES

[1] Castro, W.H. et al: *Do 'Whiplash ' Occur in Low-Speed Rear Impacts*? Europe Spine Journal, 6(6), 1997, p.366-375

[2] Galasko, C.S.: *Whiplash Associated Disorders: Cost to Society*. International Symposium Whiplash'96. Brussels (Belgium),November 16-16, 1996, p.5

[3] States, J.D. et al: *Injury Frequency and Head Restrain Effectiveness in Rear – End Impact Accidents*. In: Proc. 16th Stapp Car Crash Conference, New York (US), 1972, pp. 228-245

[4] Siegmund, G.P at el: *Head/neck Kinematic Response of Human Subjects in Low-Speed Rear-End Collisions (973341).* Proc. 41st Stapp Car Crash Conference, Lake Buena Vista (USA), November, 1997, p. 357-385.

[5] Kroonenberg, A.J.- Wismans, J.S.: *Whiplash: Reduction of Neck Injuries and Their Societal Costs in Rear End Collisions*. Proc. JSAE Spring Convention, (41), 1999, p.13-16

[6] Larder, D.R.- Twiss, M.K.-Mackay G.M.: *Neck Injury to Car Occupants Using Seat Belts*. Proc. 29th Conference of the American Association for Automotive Medicine, Washington (USA), 1985, p. 153-165.

[7] Olsson, I et al: *An In-Depth Study of Neck Injuries in Rear End Collisions*. Proc. of the International IRCOBI Conference on the Biomechanical of Impacts, Bron (France), September 12-14, 1990, p.1-

[8] Yang, K.H. et al: *On the Role of Cervical Facet Joints in Rear End Impact Neck Injury Mechanisms*. SAE Technical Paper 970497, 1997, p. 127-129

[9] Brault, J.B. et al: *Clinical Response of Human Subjects to Rear-End Automobile Impacts*. Archives of Physical Medicine and Rehabilitation, 79, 1998, p. 72-80.

[10] Wheeler, J.B. et al: *Validation of the Neck Injury Criterion (NIC) Using Kinematic and Clinical Results from Human Subjects in Rear-End Collisions*. Proc. of the International Conference on the Biomechanics of Impact, Göteborg (Sweden), September 16-18, 1998, p. 335-348.

[11] Ono, K. et al: *Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motions and Head-Neck-Torso Kinematics During Low Speed Rear Impacts* (973340), Proc. 41st Stapp Car Crash Conference, Lake Buena Vista (USA), 1997,p. 339-356.

[12] Yoganandan, N. - Pintar, F.A.- Cusick, J.F.: *Biomechanical Analyses of Whiplash Injuries Using Experimental Model*. Accident Anal & Prev, 34(5), 2001, p. 101-109.

[13] Yoganandan, N. et al: *Cervical Spine Vertebral and Facet Joint Kinematics Under Whiplash*. Journal of Biomechanical Engineering, 120, 1998, p. 305-307.

[14] Bostrom, O. et al: *A New Neck Injury Criterion Candidate-Based on Injury Findings in the Cervical Spinal Ganglia after Neck Extension Trauma*. Proc. of the International IRCOBI Conference on the Biomechanical of Impacts, Dublin (Ireland), September 11-13, 1996,p. 123-136

[15] Brault, J.B. et al: *Whiplash Injury Biomechanics: Cervical Electromyographic Response an Influence on Whiplash Dynamics.* Proc of the International Symposium Whiplash '98. Compendium of Abstracts, Tempe (USA), November 5-6, 1998, p. 25.
[16] Morris, A.P.-Thomas P.: *Neck injuries in the UK Co-Operative Crash Injury Study*. SAE Technical Paper 962433, 1996, p. 317-329
[17] Hell, W. - Shick, S. - Langwieder, K.: *Epidemiology of Cervical Spine Injuries in Rear-End Collisions and Influence of Different Anthropometric Parameters in Human Volunteer Tests*. In: "Frontiers in Whiplash Trauma"(Editors: N. Yoganandan and F. A. Pintar). Amsterdam: IOS Press, 2000, p.146-163
[18] Krafft, M.: *A Comparison of Short and Long Term Consequences of AIS 1 Neck Injuries in Rear Impacts*. Proc. of the International IRCOBI Conference on the Biomechanics of Impacts, Göteborg (Sweden), September 16-18, 1998, p. 235-248.
[19] Hohl, M.: *Soft-Tissue Injuries of Neck in Automobile Accidents*. J. Bone and Joint Surgery, 56-A (8), 1974,p.1675-82.
[20] Koch, M. et al: *Soft Tissue Injury of the Cervical Spine in Rear End and Frontal Collision*. Proc. of the International IRCOBI Conference on the Biomechanics of Impact, Brunnen (Switzerland), September 13-15, 1995, p. 273-283.
[21] Otte, D. – Pohlemann, T. - Blauth, M.: *Significance of Soft Tissue Neck Injuries AIS1 in the Accident Scene and Deformation Characteristics of Cars with Delta-v up to 10 km/h*. Proc. of the International IRCOBI Conference on the Biomechanical of Impacts, Hanover (Germany), September 24-26, 1997, p. 265-283.
[22] Carter, M. -Welsh, R.: *In-car Safety and Personal Security Needs of Female Drivers and Passengers.* Proc of ImechE Conference 'VS 2002', London, May 2002, p. 267-281.
[23] Minton, R. et al: *Causative Factors in Whiplash Injury: Implications for Current Seat and Head Restraint design*. Proc. of the International IRCOBI Conference on the Biomechanical of Impacts, Hanover (Germany), September 24-26, 1997, p. 207-222.
[24] IIHS: *Whiplash Injuries*. Status Report 30(8), Arlington, USA: Insurance Institute for Highway Safety, 1995.
[25] Matsumoto, T. et al: *Relationship Between Cervical Curvature and Discs Degeneration in Asymptomatic Subject*. Journal of Eastern Japan Association of Orthopaedics and Traumatology in Japanese Journal, 1977, 9, p.1-4
[26] Spitzer, W.O. et al: *Scientific Monograph of the Quebec Task Force on Whiplash-Associated Disorders: Redefining Whiplash and its Management*. Spine, 20 (8S), p.1S-73S, 1995
[27] Temming, J. - Zobel, R.*: Neck Distortion Injuries in Road Traffic Crashes (Analyses of the Volkswagen Database*). In: "Frontiers in Whiplash Trauma"(Editors: N. Yoganandan and F. A. Pintar). Amsterdam: IOS Press, 2000, p.119-133.
[28] Temming, J. - Zobel, R.: *Frequency and Risk of Cervical Spine Distortion Injuries in Passenger Car Accidents: Significance of Human Factors Data*. Proc. of the International IRCOBI Conference on the Biomechanics of Impact, Cologne (Germany), September 13-16, 1998,p. 319-233.
[29] Krafft, M. et al: Whiplash Associated Disorders- Factors Influencing the Incidence in Rear-ed Collision. Proc. 15th ESV-Conference, 1996, p.1426-1432
[30] Ebraheim, N.A. et al: *The Vertebral Body Depth of the Cervical Spine and its Relation to Anterior Plate- Screw Fixation*. Spine, 23 (21), 1998, p. 2299-2302
[31] Frobin, W. et al: *Vertebral Height, Disc Height, Posteroanterior Displacement and Dens-Atlas Gap in the Cervical Spine: Precision Measurement Protocol and Normal Data*. Clinical Biomechanics,17(6), 2002, p. 423-31

[32] Whitley, P.E.- Paskoff, G. R.: *Gender-Related Injury Assessment Considerations in Military Aviation Occupant Protection Modelling*. International Journal of Crashworthiness, 6(3), 2001, p. 295-305.
[33] Cerelli, E.: *Female Drivers in Fatal Crashes, Recent Trends*. Technical Report NHTSA DOT HS 808 106, National Technical Information Service, 1992.
[34] Kamibayash, L. et al: *Morphometry of Human Neck Muscles*. Spine 23 (12), 1998, p. 1314-23
[35] Kumar, S. - Narayan, Y. - Amell, T.: *Cervical Strength of Young Adults in Sagittal, Coronal and Intermediate Planes.* Clinical Biomechanics, 16,2001,pp. 380-388
[36] Jordon, A. - Mehlsen, J. - Bulow, P.M.: *Maximal Isometric Strength of Cervical Musculature in 100 Healthy Volunteers*. Spine, 24, 1999, p. 1343-8
[37] Vasavada, A.N.- Li, S. - Delp, S.L.: *Three-Dimensional Isometric Strength of Neck Muscles in Humans*. Spine, 26 (17), 2001, p.1904-1909
[38] Brault, J.R- Siegmund, G. P. -Wheeler J. B: *Cervical Muscle Response During Whiplash: Evidence of Lengthening Muscle Contraction.* Clinical Biomechanics, 15,2000, pp. 426-435
[39] Szabo, T.J- Welcher, J.B.: *Human Subject Kinematics and Electromyographic Activity During Low Speed Rear Impacts*. Proc. of the 40th Stapp Car Crash Conference (962432), Albuquerque (USA), 1996, p. 339-356.
[40] Otremski, I., et al.: S*oft Tissue Cervical spine Injuries in Motor Vehicle Accidents*. Injury, 20, pp. 349-351,1989
[41] Welsh, R. - Lenard, J.: Male *and female car drivers - differences in collision and injury risks*. Proc. of 45th Conference of the Association for the Advancement of Automotive Medicine, Texas (USA), September 2001, 45, 2001,p. 73-91.
[42] People Size Software, 1998, Open Ergonomics Ltd., Loughborough Technology Center, UK, 1998
[43] Kleiven, S. - Holst, H.: *Consequences of Head Size Following Trauma to the Human Head.* Journal of Biomechanics, 35, 2002, p. 153-160.
[44] Golinski, W.: *Three-Dimensional Dynamic Modelling of Human Cervical Spine in Whiplash Situations*. PhD thesis, Nottingham Trent University, Nottingham 2000
[45] Myklebust, J.B. et al: Tensile Strength of Spinal Ligaments. Spine, 13,1988, p526-531
[46] Yamadda, H.: *Strength of Biological Materials*. The Williams & Wilkins Comp, Baltimore, 1970, p.93-95
[47] Siegmund, GP, Brault JR (2001). *The Role of Cervical Muscles in Whiplash*. In: ”Frontiers in Whiplash Trauma”(Editors: N. Yoganandan and F. A. Pintar). Amsterdam: IOS Press, 2000, p.295-320.