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Initial Studies of Dynamic Responses of Female and Male Volunteers in Rear Impact Tests

by

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INITIAL STUDIES OF DYNAMIC RESPONSES OF FEMALE AND MALE VOLUNTEERS IN REAR IMPACT TESTS

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ABSTRACT

Whiplash Associated Disorder (WAD) - commonly denoted whiplash injury - resulting from vehicle impacts, is a worldwide epidemic. These injuries occur at relatively low changes of velocity (typically between 10-25 km/h) and in impacts from all directions. Rear impacts are however the most common in the accident statistics. Since the middle of the 1960s, statistical data has shown that females have up to three times higher risk of sustaining whiplash injuries than males, even under similar crash conditions. Studies have indicated that there may be characteristic differences in the rear impact dynamic response between males and females. The 50th percentile male dummy might thus limit the assessment and development of whiplash prevention systems that adequately protect both male and female occupants. Data from volunteer tests is needed to establish the dynamic response for females and males. Such data is fundamental for developing future occupant mathematical and/or mechanical models for crash safety development and assessment. These models can be used, not only as a tool in the design of protective systems, but also in the process of further evaluation and development of injury criteria.

The aim of this study was to quantify the differences in dynamic response between average sized females and males in low-speed rear impacts.

Two rear impact volunteer studies were conducted. In the first study, data for the 50th percentile female were extracted from a previously performed rear impact car-to-car crash test series with female and male volunteers at 4 km/h and 8 km/h. In the second study, a sled test series was performed with 50th percentile female volunteers at 5 km/h and 7 km/h. In both studies, response corridors for the female volunteers were generated and compared with previously published corridors for the 50th percentile male. Additionally, the Neck Injury Criterion (NIC) values, head-to-head restraint distances and contact times were compared for the female and male volunteers in both studies.

The overall result showed differences between the females and the males in the dynamic response and in the NIC values. For example, the head x-acceleration peaks were on average higher and earlier for the females; the head, T1, and head relative to T1 x-displacement peaks were on average lower and earlier; the initial head-to-head restraint distance was on average smaller for the females, resulting in earlier head-to-head restraint contact time for the females.

KEYWORDS: whiplash, neck injury, volunteers, response corridors, crash test, rear impact, females, dynamic response

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PAPER I

Linder A, Carlsson A, Svensson MY, Siegmund GP (2008) *Dynamic Responses of Female and Male Volunteers in Rear Impacts*, Traffic Inj. Prev., Vol. 9, No. 6, pp. 592–599

Division of work between authors: Siegmund performed the volunteer test series during the late 1990s and provided the data for further analysis. Linder and Svensson made the outline of this study, based on a subset of the volunteers from the first study. Carlsson made the analysis and the presentation of data. The paper was written by Linder and Carlsson, and reviewed by all authors.

PAPER II

Carlsson A, Linder A, Svensson MY, Davidsson J, Hell W, *Dynamic Responses of Female Volunteers in Rear Impacts and Comparison to Previous Male Volunteer Tests*, submitted for publication

Division of work between authors: Carlsson made the outline of this study with support by Linder and Svensson. Hell was responsible for the recruitment and medical examination of the female volunteers. Carlsson, Linder, Svensson, Davidsson, and Hell participated, partly or full time, during the tests. Davidsson provided the data from previous tests with male volunteers. Carlsson made the analysis and the presentation of data. The paper was written by Carlsson, and reviewed by all authors.

Conference Presentations of the present work

Carlsson, A., Linder, A., Svensson, M., Siegmund, G. (2008) *Dynamic Responses of Female and Male Volunteers in Rear Impacts*, World Congress on Neck Pain, January 20-22, 2008, Los Angeles, CA

Carlsson A, Linder A, Svensson M, Davidsson J, Schick S, Horion S, Hell W (2008) *Female Volunteer Motion in Rear Impact Sled Tests in Comparison to Results from Earlier Male Volunteer Tests*, Proc. IRCOBI Conf., Bern (Switzerland), pp. 365–366

DEFINITIONS AND ABBREVATIONS

Whiplash Associated Disorder (WAD) Cervical spine injury Whiplash-type neck distortion Flexion-torsion neck injury AIS1 neck injury Acute strain of the cervical spine	Examples of names used for whiplash injury
Anterior	In front of…
AROM	Active Range Of Motion
BioRID	Biofidelic Rear Impact Dummy
CSN	Central Nervous System
CFD	Computational Fluid Dynamics
Extension (of the neck)	Rearward bending (of the neck)
Kyphosis	Outward curvature of a portion of the spine
Flexion (of the neck)	Forward bending (of the neck)
Lordosis	Inward curvature of a portion of the spine
MRI	Magnetic Resonance Imaging
NHTSA	National Highway Traffic Safety Administration
NIC	Neck Injury Criterion
PMHS	Post Mortem Test Subject (human cadaver)
Posterior	Behind of…
Protraction	Head moved forward relative to the torso, with no angular change
Retraction	Head moved rearward relative to the torso, with no angular change
RID3D	Rear Impact Dummy version 3D
SAHR	Saab Active Head Restraint
T1	First thoracic vertebra
WhiPS	Volvo's Whiplash Protection System
WIL	Toyota's Whiplash Injury Lessening system

To

Märta & Greta

♥

This is for you …

1. INTRODUCTION

This thesis for the degree of licentiate of engineering considers the dynamic response of females and males induced by low-speed rear impacts. During this type of impact, the torso of the occupant is 1) pressed into the seatback and 2) pushed away from the seatback, while the head/neck is exposed to a whiplash type motion. Large loads can then arise in the fragile and complicated structures of the neck due to the head inertia, and result in so called whiplash injuries. There are many different names used for this injury, like Whiplash Associated Disorder (WAD), cervical spine injury, whiplash-type neck distortion, flexion-torsion neck injury, AIS1 neck injury, acute strain of the cervical spine, etc. In this thesis the most common name is used: whiplash injury. The injury mechanisms are not fully understood since the whiplash injuries are difficult to detect by diagnostic tools like X-rays or MRI (Magnetic Resonance Imaging). These injuries are classified as minor, although they can cause long-lasting pain and disability. The symptoms are well known, such as neck pain, stiffness, loss of sensation, memory impairment and concentration difficulties.

The risk of whiplash injury is up to three times higher for females compared to males under similar crash conditions. However, when assessing the vehicle safety, the only available occupant model for this impact scenario is a model of an average male. For males, significant progress in preventing whiplash injuries has been made due to dummy development and seat optimization. There is a need to establish the characteristics of the female response in rear impact and implement these data in future models for rear impact testing and evaluation.

1.1 WHIPLASH INJURIES

Whiplash injury resulting from vehicle impacts is a worldwide epidemic. From a societal perspective these injuries are costly since they are frequent and can lead to long-lasting pain and disability. In Europe, the yearly cost for whiplash injuries has been estimated to 10 billion euros (Richter et al. 2000). In Japan, 547,654 traffic related injuries were registered during 1996 and 44% suffered from neck injury (Watanabe et al. 2000). In USA, the number of whiplash injuries each year has been estimated to 800,000. Of these whiplash injuries, 270,000 were resulting from rear impacts with an annual cost of $2.7 billion (NHTSA 2004). In Sweden (with 9 million citizens), more than 30,000 whiplash injuries are reported after vehicle collisions every year and the associated socio-economic impact is on the order of 0.4 billion euros per year, (the Whiplash Commission 2005). Concerning modern cars on the Swedish market, whiplash injuries account for approximately 70% of all injuries leading to disability (Kullgren et al. 2007).

Data from different parts of the world has shown that the risk and the number of whiplash injuries have steadily increased from the late 1960s to the late 1990s (Galasko et al. 1993; von Koch et al. 1994; Ono & Kanno 1996; Hell et al. 1998; Temming & Zobel 1998; Richter et al. 2000; Morris & Thomas 1996). **Table 1** summarizes the findings from the studies.

Table 1. *The increase of whiplash injuries between late 1960s and late 1990s in different countries.*

Country	Reference	Data Source	Years	Whiplash Injury Increase*) From:	To:	
Sweden	von Koch et al. (1994)	Insurance Company	1977 – 1991	19%	47%	
Germany	Hell et al. (1998)	Insurance Company	1969 – 1990	20%	35%	
	Temming & Zobel (1998)	VW Accident Database	1987 – 1996	9%	17%	
	Richter et al. (2000)	Accident Research Team	1985 – 1997	10%	>30%	
Japan	Ono & Kanno (1996)	Insurance Company	1985 – 1991	44%	51%	
UK	Galasko et al. (1993)	Hospital data	1982 – 1991	8%	46%	
	Morris & Thomas (1996)	CCIS database	1984 – 1991	14%	31%	(females)
				10%	18%	(males)

*) The percentages given in the table are calculated in different ways, it is not possible to compare the data from different references with each other.

In Germany the incidence of 'cervical spine injuries' in motor vehicle accidents was almost doubled from 1969 to 1990 (**Figure 1a**) (Hell et al. 1998). In UK the 'soft tissue injuries to the cervical spine' increased from 8 per cent 1982 to 46 per cent in 1991 (**Figure 1b**), (Galasko et al. 1993). Morris and Thomas (1996) reported that the 'neck injury' rates increased almost linearly over the years from 1984 to 1991 and that the increase was larger for the females compared to the males (**Table 1**).

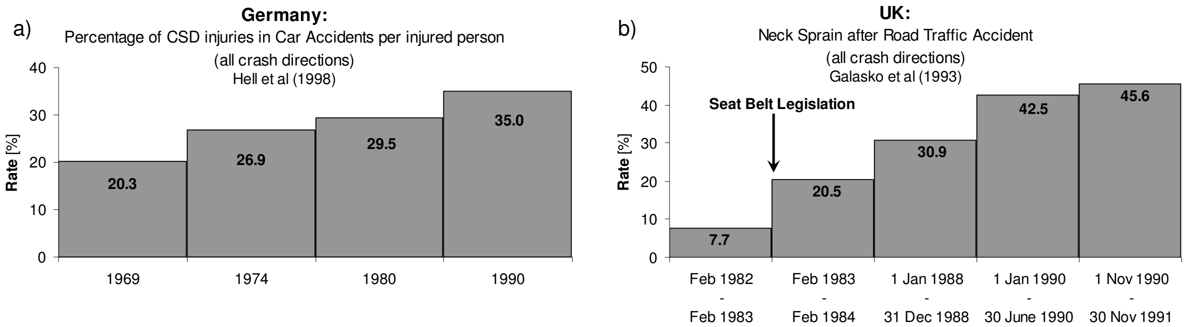


Figure 1. *Whiplash injury rate in a) Germany, and b) UK, with regards to time. Based on Hell et al. (1998) and Galasko et al. (1993).*

Several factors contribute to the increase of the whiplash injury risk and the numbers of whiplash injuries between the late 1960s and the late 1990s, as for example:

- the seatbacks have increased up to 5.5-fold in strength from the 1960s to the 1990s in order to provide greater occupant retention in high-speed crashes (Viano 2008). The increase in strength has resulted in higher seat stiffness, i.e. an increased change in force with rearward occupant displacement. The higher stiffness of the seat affects the interaction between the occupant and the seatback and may increase the forces on the neck.
- The improvements in vehicle intrusion (Delannoy & Diboine 2001), lead to a stronger and stiffer vehicle structure.
- The wearing of seatbelts increases the whiplash injury risk (Deans et al. 1987; Otremski et al. 1989; Maag et al. 1990; Galasko et al. 1993 among others). According to Galasko et al. (1993) the whiplash injury rate increased from 7.7 per cent one year before the the seatbelt legislation in February 1983, to 20.5 per cent one year after the seatbelt legislation in the UK (**Figure 1b**). Deans et al. (1987) reported that >1 year after the crash, 34 per cent who were wearing a seatbelt still experienced neck pain, while only 20 per cent of those who had not worn a seatbelt still felt neck pain.

A small decrease of the long-term whiplash injury risk in rear impacts (from 15.5% to 13.6%) was found for cars manufactured after 1997 and equipped with standard seats (i.e. no advanced whiplash protection systems) in comparison to cars manufactured before 1997 (Kullgren et al. 2007). Cars equipped with advanced whiplash protection systems had approximately 50 per cent lower risk of long-term whiplash injuries in rear impacts than cars manufactured after 1997 without whiplash protections systems (Kullgren et al. 2007). In frontal impacts it was found that airbags in combination with seatbelt pretensioners reduce the numbers of whiplash injuries by 41 ±15 per cent (Kullgren et al. 2000).

Since the middle of the 1960s, statistical data has shown that females have a higher risk of sustaining whiplash injuries than males, even under similar crash conditions (**Figure 2**) (Narragon 1965; Kihlberg 1969; O'Neill et al. 1972; Thomas et al. 1982; Otremski et al. 1989; Maag et al. 1990; Morris & Thomas 1996; Dolinis 1997; Temming & Zobel 1998); Richter et al. 2000; Chapline et al. 2000; Krafft et al. 2003; Jakobsson et al. 2004; Storvik et al. 2009 among others). According to these studies the whiplash injury risk is up to three times higher for the females compared to the males.

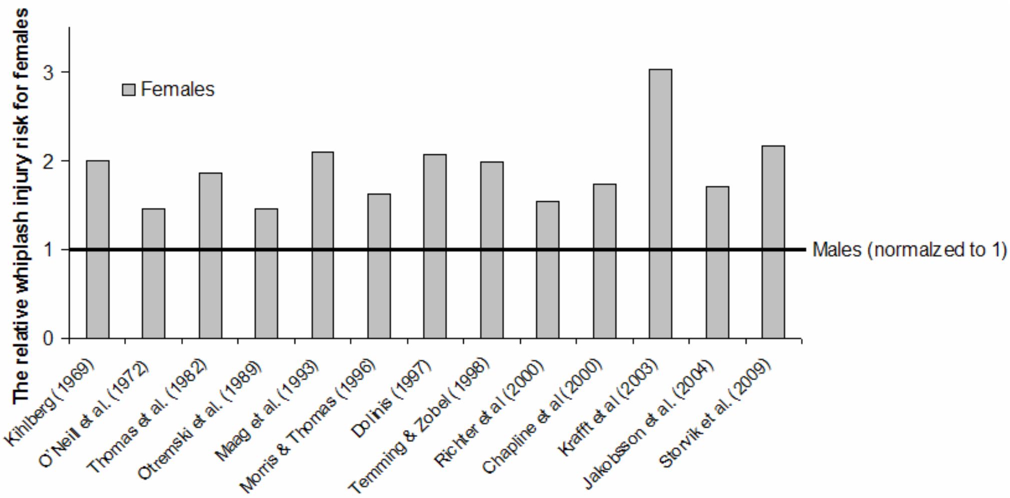


Figure 2. *The relative whiplash injury risk for females (shaded light grey) compared to males (normalized to 1).*

From an individual perspective the whiplash injury can have a major influence on the daily life with symptoms like neck pain, stiffness, loss of sensation, memory impairment and concentration difficulties (the Whiplash Commission 2005), which affect the quality of life and the ability to work. The majority of those who experience initial neck symptoms after the car crash will recover, most of them within a week after the crash, but 5–10 per cent of them will experience permanent disabilities of various degrees (Nygren et al 1984; the Whiplash Commission 2005). These injuries occur at relatively low changes of velocities, typically between 10–25 km/h (Eichberger et al. 1996; Kullgren et al. 2003) and in impacts from all directions (Galasko et al. 1993; Krafft 1998). Rear impacts are however the most common in the accident statistics (Watanabe et al. 2000).

1.2 REAR IMPACTS

Rear impact induced whiplash injuries account for ~50 per cent of the total number of whiplash injuries according to data from hospital records and insurance companies (Galasko et al. 1993; Krafft 1998; Hell et al. 1998) (**Figure 3a,b**). Data extracted from traffic accident databases is often biased towards more severe crashes and is consequently dominated by whiplash injuries induced by frontal impacts (**Figure 3c**).

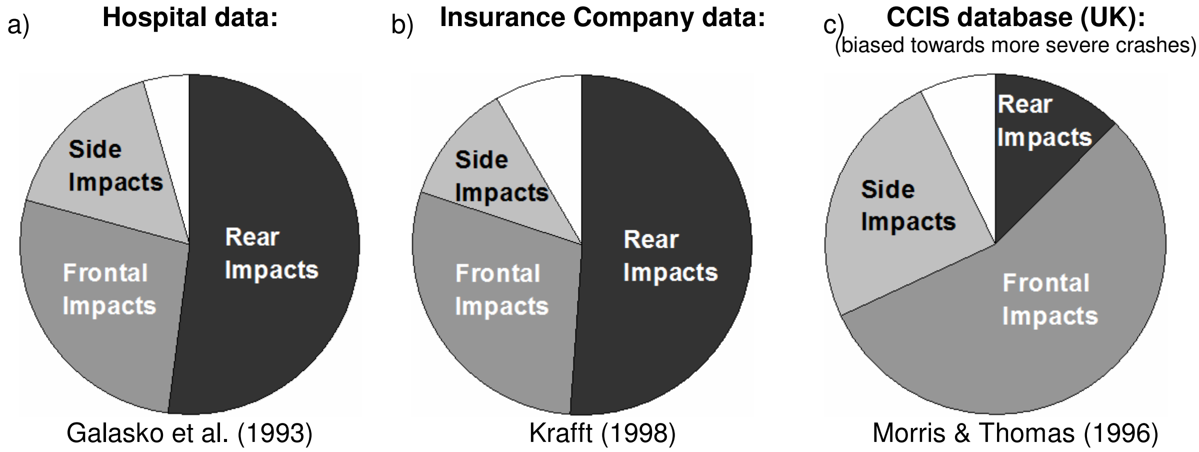


Figure 3. *The distribution of whiplash injuries with regard to different impact directions, based on a) hospital records (Galasko et al. 1993); b) insurance claims (Krafft 1998); and c) the CCIS database (Morris & Thomas 1996).*

The majority of the rear impacts (79%) occur when the struck vehicle has come to a stop (21% at a stop light or sign, 11% making left turn, 10% in an intersection, 8% stopped in a queue, and 6% standing at the side of the road). The rest of the vehicles were hit while driving (8%); during hard braking (7%); and while slowing down (5%) (Viano & Olsen 2001).

Statistical data has identified how whiplash injury risks in rear impacts are influenced by different factors such as impact severity, vehicle specific features, seating position, head restraints, design and mechanical properties of the seat, and cccupant related factors, further discussed below.

Impact Severity

Several studies have shown correlation between whiplash injury risk and impact severity (Ryan et al. 1994; Eichberger et al. 1996; Krafft et al. 2002; Kullgren et al. 2003). Based on data from crash recorders, Krafft et al. (2002) and Kullgren et al. (2003) found that the long-term whiplash injury risk approached 100 per cent for mean vehicle accelerations above 7g. At mean accelerations below 5g the long-term injury risk was low (Kullgren et al. 2003). It was also found that for mean accelerations below 3g the risk approached zero (Krafft et al. 2002).

Vehicle Specific Features

Vehicle specific features such as car model, car mass, and mechanical properties of the crash zones of the involved vehicles influence the whiplash injury risk in rear impacts. The long-term whiplash injury risk varies widely between different car models, even if their mass is the same (Krafft 1998). A 22 per cent higher risk was found for long-term disability in rear impacts into a car with tow-bar compared to without a tow-bar (same car model) (Krafft 1998).

Seating Position

The whiplash injury risk depends on in which car seat the occupant is positioned. Several studies have indicated that front seat occupants have a higher whiplash injury risk than rear seat occupants (States et al. 1972; Carlsson et al. 1985; Jakobsson et al. 2000). But when looking at the long-term consequences, Krafft et al. (2003) found a different relationship for the females. In this study a paired comparison was performed of all neck injuries reported to Folksam, a Swedish insurance company, in rear impacts during 1990–1999. The males had a lower injury risk in the rear seat compared to the front seats, while the females had a considerably higher injury risk in the rear seat; the lowest risk for the females was found for the front passenger seat (**Figure 4**). The risk of permanent disability was three times higher for female drivers compared to male drivers. Similarily, it was 1.5 times higher for female front seat passengers, and more than five times higher for female rear seat passengers.

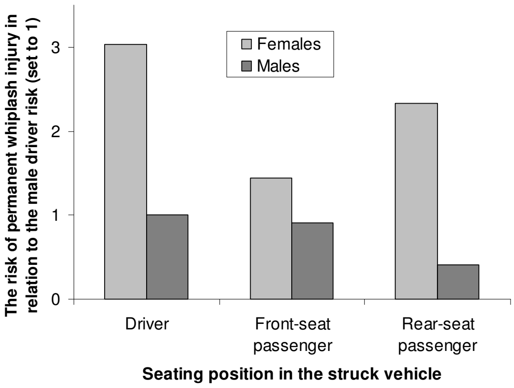


Figure 4. *The risk of permanent whiplash injury in relation to the male driver risk (normalized to 1) for different seating positions in rear impacts. Based on Krafft et al. (2003).*

Head Restraints

The effectiveness of head restraints in rear impacts have been evaluated in many studies (O'Neill et al. 1972; States & Balcerak 1973; Kahane 1982; Nygren 1984; Lövsund et al. 1988) among others. Integral head restraints reduce the overall injury risk in rear impacts by 17 per cent, while adjustable restraints had a reduction by 10 per cent, according to Kahane (1982). The difference found for integral and adjustable head restraints could be attributed to the failures by occupants to position their adjustable restaints correctly. No injury reducing effect due to head restraint was found in the rear seat, while the effectiveness in the front seat was 29.8 per cent (Lövsund et al. 1988).

Several studies have reported that improved head restraint geometry reduces the whiplash injury risk to a higher extent for females than for males (**Table 2**) (States et al. 1972; O'Neill et al. 1972; Thomas et al. 1982; Chapline et al. 2000; Farmer et al. 2003). A 37 per cent reduction of whiplash injury frequency was found among female drivers compared to "very little effect" among male drivers in the improved seat of model year 2000 Ford Taurus and Mercury Sable (Farmer et al. 2003). Foret-Bruno et al. (1991), on the other hand, reported that the injury reducing effect of head restraints was about the same for males, 34 per cent (from 23% to 15%), and females, 33 per cent (from 45% to 30%).

Table 2. *The injury reducing effect of head restraints.*

Reference	The Injury Reducing Effect of Head Restraints			
	Females		Males	
States et al. (1972)	51%=>38%	**-25%**	40%=>35%	**-12%**
O'Neill et al. (1972)	37%=>29%	**-22%**	24%=>22%	**-10%**
Thomas et al. (1982)	44%=>39%	**-11%**	22%=>27%	**"no reducing effect"**
Foret-Bruno et al. (1991)	45%=>30%	**-33%**	23%=>15%	**-34%**
Chapline et al. (2000)	52%=>29%	**-44%**	29%=>18%	**-38%**
Farmer et al. (2003)	not specified	**-37%**	not specified	**"very little effect"**

Increased head-to-head restraint distance has been reported to be associated with increased whiplash injury risk (Carlsson et al 1985; Nygren et al. 1985; Olsson et al. 1990; Deutcher 1996; Farmer et al 1999; Jakobsson et al. 2004). A head restraint located less than 10 cm from the back of the head was found to be more beneficial with regards to whiplash injury outcome compared to a head-to-head restraint distance higher than 10 cm. Based on mathematical simulations, Stemper et al. (2006) suggested that by limiting the head restraint head-to-head restraint distance to less than 6 cm, either passively or actively after impact, further whiplash injury reduction may be accomplished. In contrast, Chapline et al. (2000) reported that the horizontal distance was not significantly related to neck pain. Instead, it was the height of the head restraint that was the primary factor related to head restraint effectiveness, especially for females (**Figure 5**). Although it is not statistically significant for male drivers, the percentages of both female and male drivers reporting neck pain increased as the height of the head restraint further decreased below the head's center of gravity.

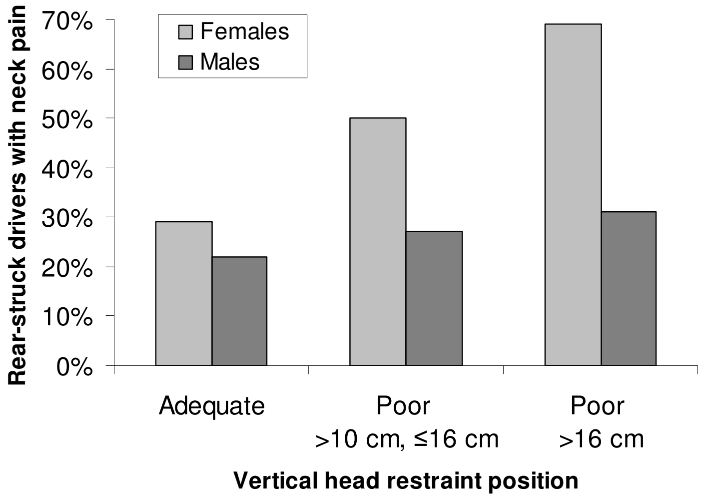


Figure 5. *Per cent of rear-struck drivers with neck pain by vertical head restraint position (distance between the centre of gravity of the head and the top of the head restraint). Based on Chapline at al. (2000).*

Females tend to be positioned closer to the head restraint than males (Szabo et al. 1994 (estimation from graph); Minton et al. 1997; Hell et al. 1999; Jonsson et al. 2007; Schick et al. 2008), and the distance depends on the seating position (Jonsson et al. 2007). The average head-to-head restraint distance for males and females in these studies are summarized in **Figure 6**. Jonsson et al. (2008) found that the head-to-head restraint distance increased on average ~4 cm for both female and male volunteers while driving the vehicle in comparison to the distance measured in a stationary vehicle. Cullen et al. (1996), on the other hand, did not find any significant difference in head-to-head restraint distance for males and females during driving. In this study vehicle occupants were filmed when the vehicles passed rigged cameras and the head-to-head restraint distance was estimated from film analysis.

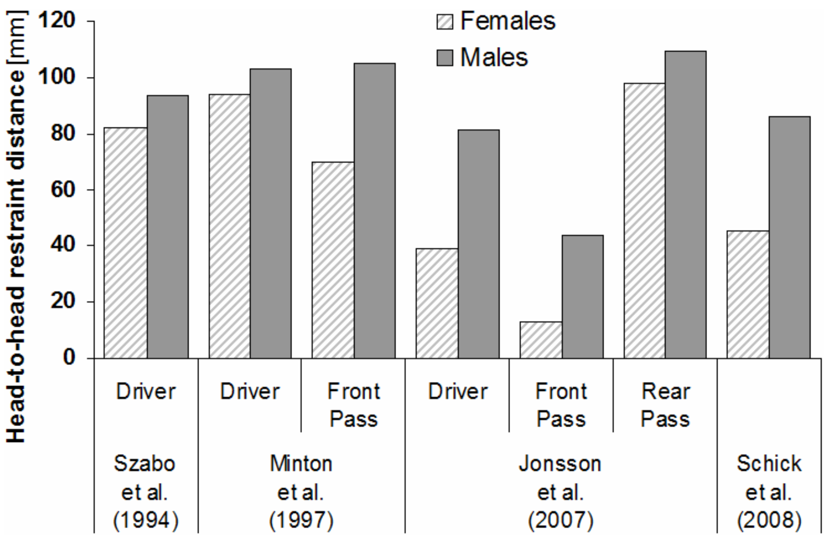


Figure 6. *The average head-to-head restraint distance for male and female volunteers in different studies. Females are represented by striped bars and males by grey bars.*

Design and Mechanical Properties of the Seat

The design and mechanical properties of the seat and seatback affect the whiplash injury risk. From the 1960s to the 1990s the seatbacks have increased up to 5.5-fold in strength in order to increase the vehicle crashworthiness in high-speed rear impacts (Viano 2008). It is hypothesized to be one of the reasons for the increase of whiplash injuries starting in the late 1960s, especially for females (Viano 2003). Yielding or collapsing of the seatback have been reported to decrease the whiplash injury risk in rear impacts (Kihlberg 1969; States et al. 1969; O'Neill et al. 1972; Thomas et al. 1982; Foret-Bruno et al. 1991; Parkin et al 1995; Morris & Thomas 1996; Krafft et al. 2004). Thomas et al. (1982) concluded that "damage seat-back or seat track failure have a greater effectiveness than head restraint, considering cervical pain reduction" and that "seat damage is effective for females only and reduces their whiplash injury risk by 45 per cent." Parkin et al. (1995) found that the AIS1 neck injuries were approximately twice as frequent in an undamaged seat than in a yielding seat. Krafft et al. (2004) found a 84 per cent injury reducing effect on the long-term whiplash injury risk in a study where 8,000 cars with poor head restraint geometry were redesigned with yielding seat attachment brackets, which was the only design change.

Occupant Related Factors

The whiplash injury risk has been shown to be influenced by stature, age, initial position, and the awareness of an impending impact, apart from gender. The whiplash injury risk generally shows an increasing trend for increasing statures for both males and females (Kihlberg 1969; Temming & Zobel 1998; Lundell et al. 1998; Jakobsson et al. 2000). When male and female drivers of the same statures were compared in these studies, the injury risk were on average 2.1 times higher for the females. The risk of whiplash injury seems to have a peak at the middle age, and a decrease at older age (Jakobsson et al. 2000; Farmer et al. 1999; Temming & Zobel 1998). Awareness of an impending impact decreases the long-term whiplash injury risk according to Sturzenegger et al. (1995) and Dolinis (1997), while Minton et al. (1997) could not find such a correlation. Rotated head during an impact resulted in a higher incidence of persistent symptoms after the collision (Sturznegger et al. 1995; Jakobsson 2004). The mass of the occupants appears to have little effect on the whiplash injury risk (States et al. 1972; Minton et al. 1997; Temming & Zobel 1998).

1.3 THE ANATOMY AND RANGE OF MOTION OF THE NECK

The spine - or the *vertebral column* - is formed by a series of bones - the *vertebrae* (**Figure 7**). The vertebrae are grouped under the names cervical, thoracic, lumbar, sacral, and coccygeal spine according to the regions of the spine they occupy. The neck - or the *cervical spine* - is formed by seven vertebrae, denoted C1-C7. The topmost vertebra, C1 (*atlas*), forms together with the vertebra closest beneath, C2 (*axis*), the joint connecting the spine to the the occipital bone of the skull. The cervical spine is comparatively mobile compared to other parts of the spine, but the movement of rotating the head to left and right happens almost entirely at the C1-C2 joint - the *atlanto-axial joint*. Similarily, the movements of bending of the head take place predominantly at the joint between C1 and the occipital bone - the *atlanto-occipital joint*.

The majority of the vertabrae consist of a front segment – the vertebral body – and a rear segment – the vertebral arch – which enclose the *vertebral foramen*. The two topmost cervical vertebrae, C1 and C2, have a somewhat different structure in order to increase the range of motion of the head. The *spinous* and *transverse processes* serve as attachment points for muscles and ligaments. These muscles and ligaments account for stability and movements, especially of the head and the neck (Schmitt et al. 2004). When the vertebrae are articulated with each other, the bodies form a strong pillar for the support of the head and the trunk, and the vertebral foramen constitute a canal for the protection of the vulnerable *spinal cord*. Cervical vertebrae possess *transverse foramina* to allow for the vertebral arteries to pass through on their way to the foramen magnum to end in the circle of Willis - a circle of arteries that supply blood to the brain.

There are two *facet joints* between each pair of cervical vertebrae from C2 to C7. The facet joint is a synovial joint enclosed by a thin, loose ligament known as the facet capsule (Siegmund et al. 2009). Adjacent vertebrae are separated by *intervertebral discs*.

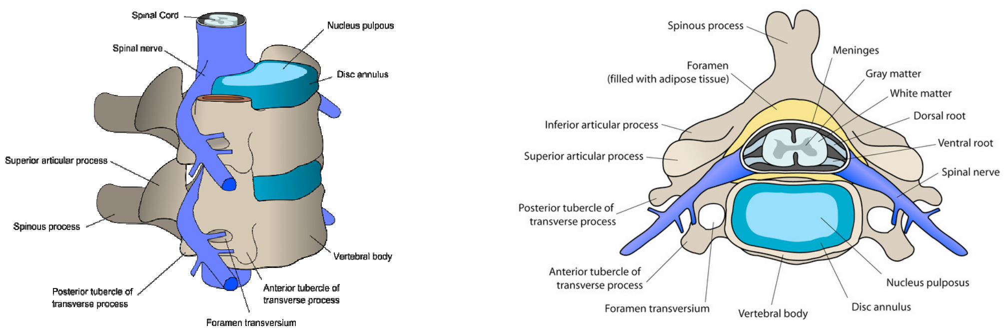


Figure 7. *a) Human cervical spine segment. b) Cervical spine vertebra. The pictures were downloaded from Wikimedia.*

Differences in the anatomy and physiology of the neck have been reported for males and females, which may contribute to the higher whiplash injury risk for females. For example, it has been reported that:

- females have a lower strength of the neck muscles (Vasavada et al. 2001; Vasavada et al. 2008; Foust et al. 1973). Vasavada et al. (2001) reported, based on measurements on 11 males and 5 females, that the males had 2-2.5 times greater moment-generating muscle capacity than the females. Vasavada et al. (2008) studied differences in head and neck geometry and neck strength in 14 pairs of male and female subjects matched for standing height and neck length. It was found that female necks had significantly lower strength than male necks; 20 per cent lower in extension (**Figure 16c**) and 32 per cent lower in flexion,

(**Figure 16d**). They also concluded that the females had 33% more head mass per unit neck muscle area than the size-matched males. Foust et al. (1973) found that the average neck flexor and extensor muscle strength of males is greater than that of females in every age and stature group. Females tend to decrease gradually in neck strength throughout their lives, while males are often stronger at middle age than they were when young, (**Figure 8**).

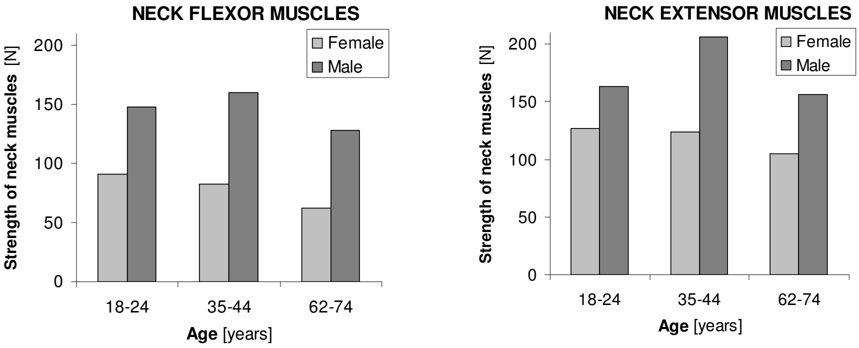


Figure 8. *The strength of a) flexor b) extensor neck muscles in degrees for females (shaded light gray) and males (shaded dark gray) of the same size, for different age categories, based on Foust et al. (1973).*

- females have faster neck muscle reflexes than males (Foust et al. 1973). "For example, on the average, females reflex about 11 per cent faster than males, but are only 60 per cent as strong."
- females have smaller neck circumference (Vasavada et al. 2001), and more slender necks (Vasavada et al. 2008).
- females have smaller necks relative to the head size (1:151) compared to males (1:135), States et al. (1972) (ratio = head circumference^3/neck circumference^2).
- females have smaller vertebral dimensions than males (DeSantis Klinich et al. 2004; Stemper et al. 2008; Vasavada et al. 2008). DeSantis Klinich et al. (2004) concluded that ''there are differences in vertebral size associated with gender that do not solely result from the stature differences between men and women.'' Stemper et al. (2008) performed computed tomography scans of the cervical spine on size-matched young healthy male and female volunteers. Geometrical dimensions were obtained at the C4 level. It was found that all geometrical measures were greater in males. Vertebral width and disc-facet depth were significantly greater in males. Additionally, segmental support area, combining interfacet width and disc-facet depth, was greater in males, indicating more stable intervertebral coupling. Vasavada et al. (2008) reported similar results in their study.
- females and males have different range of motion of the neck (**pages 11-13**).
- female have decreased collagen content and increased elastin content in lumbar ligaments (Osakabe et al. 2001). Differences in ligament structural components may lead to decreased stiffness in female spines, cited from Stemper et al. (2008).
- female tolerance limits for lower neck shear force (384 N) is considerable lower than male tolerance limits (636 N) (Stemper et al. 2007).
- females have a narrower spinal canal (Pettersson et al. 1995; Tatarek 2005). In the study by Pettersson et al. (1995) it was found that the spinal canal was significantly smaller in whiplash patients with persistent symptoms. Differences were found between males and females; the spinal canal was significantly narrower for the females. Tatarek (2005) found that females had a narrower spinal canal compared to males based on measurements of the sagittal and transverse diameters of the cervical canal in skeletons.

For more detailed reviews of the anatomical and physiological differences of the neck for males and females, see Mordaka (2004), Vasavada et al. (2008), and Stemper et al. (2008).

Extension

The shape of the neck that occurs when the head is angled rearwards relative to the torso is called an *extension* of the neck, **Figure 10**. In a study by Youdas et al. (1992) it was found that the active range of motion (AROM) of neck extension was largest for young people, and that the AROM of neck extension was decreasing as the age increased (**Figure 12a**). At 11-19 years of age the AROM of neck extension was ~85° from the neutral position, while at 80-89 years of age it was reduced to ~50° from the neutral position. The females had a significantly larger AROM of neck extension at 20-69 years of age compared to the males (**Figures 12a,b**). The most pronounced difference was found at 40-49 years of age, with a 24 per cent larger AROM of neck extension for the females, (**Figure 12a**).

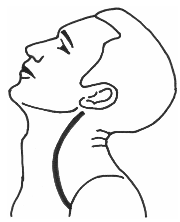
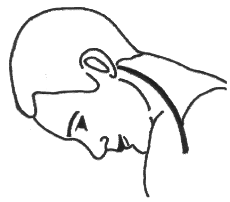


Figure 10. *Extension of the neck. Adapted from Linder (2001).*

Flexion

The shape of the neck that occurs when the head is angled forward relative to the torso is called a *flexion* of the neck (**Figure 11**). The neck flexion motion is normally limited by the chest. In a study by Youdas et al. (1992), the neck flexion was found to be significantly associated with age, but the rate of loss per year and the number of degrees of AROM at a given age were not found to differ for male and female subjects. For male and female test subjects of the same age, the AROM of neck flexion was estimated to be the same (**Figure 12b**). Seacrist et al. (2009), on the other hand, found a statistically significant increase in the cervical spine flexion angle in adult females compared to adult males. A decrease in flexion angle was found for increasing age in that study also.



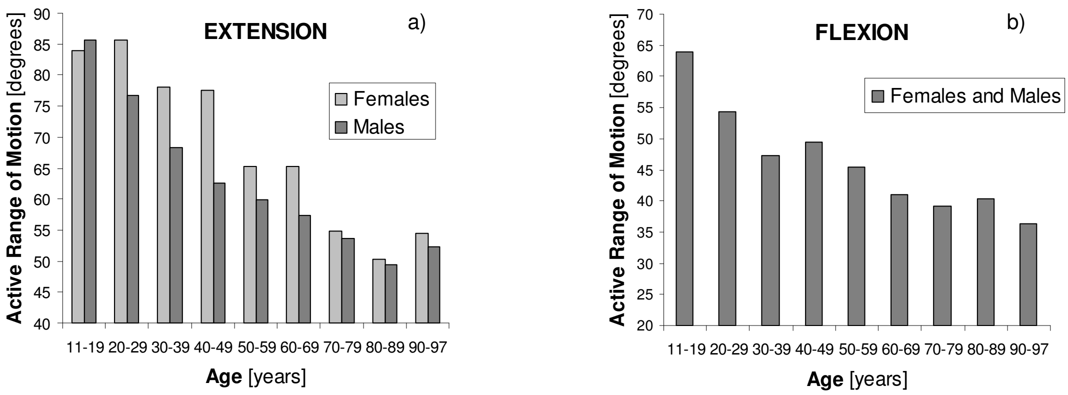


Figure 12. *a) The active range of motion (AROM) of neck extension (in degrees) for females and males for different age categories. b) The AROM of neck flexion in degrees for females and males for different age categories. Based on Youdas et al. (1992). Females are represented by light gray bars and males by dark gray bars.*

Retraction

Figure 13. Retraction of the neck. Adapted from Linder (2001).

The shape of the neck that occurs when the head is moved rearward relative to the torso, with no angular change, is called a *retraction* of the neck (**Figure 13**). The upper part of the neck is flexed and the lower part is extended during the retraction, which results in an S-curvature of the neck.

In static tests, Jonsson (2007) did not find any significant differences between males and females in cervical retraction capacity.

In dynamic tests, a more pronounced S-curved shape of the neck for females compared to males have been reported (Stemper et al. 2003; Ono et al. 2006). Stemper et al. (2003) performed rear impact tests with ten intact human cadaver head–neck complexes (5 males, 5 females) and the intervertebral kinematics were analyzed as a function of spinal level at the time of maximum cervical S-curvature. Segmental angles were significantly greater in female specimens at C2–C3, C4–C5, C5–C6, and C6–C7 levels. In the study by Ono et al. (2006) six human volunteers (4 males, 2 females) were subjected to rear impacts at 6 km/h. The dynamic response of the neck was monitored by a high speed x-ray camera. The cervical vertebral rotation angle of females was higher than that of the males, and their cervical spine exhibited a more significant S-curved deformation.

Total Range of Motion

The total range of extension–flexion motion (**Figures 10–11**) is larger for females compared to males (Buck et al. 1959; Foust et al. 1973). Buck et al. (1959) reported that the total range of extension–flexion motion was 150° for 18–23 years old females, while is was 139° for the males of the same age group. Foust et al. (1973) studied the total range of extension–flexion motion for males and females with regards to age, (**Figure 15a**). Age had a pronounced effect, but the females tended to lose mobility gradually throughout their lives, while the males deteriorated more rapidly between youth and middle age than they did thereafter.

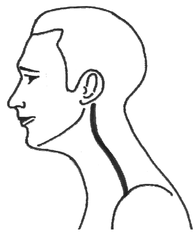


Figure 14. Protraction of the neck. Adapted from Linder (2001).

The total range of retraction–protraction motion (**Figures 13–14**) is smaller for females compared to males in seated posture (Hanten et al. 1991; Hanten et al. 2000). Hanten et al. (1991) studied the total range of retraction–protraction motion for males and females with regards to age (**Figure 15b**). For the males the range of motion increased until 50 years of age before it started to decrease, while for the females the range of motion had a small increase until 40 years of age before the decrease started. According to Hanten et al. (2000) the total range of motion was 10.0 cm for females, and 12.8 cm for males. It is unclear whether the larger motion range for the males was due to differences in stature/size of the males and females in these studies.

Hanten et al. (1991) reported that within their available retraction–protraction excursion range, females held their heads in a more forward position. Mean per cent distance from retracted to resting head postion was 47 per cent for females and 43 per cent for males. Similar results were found in Hanten et al. (2000) with 43.4 per cent for the females and 39.5 per cent for the males.

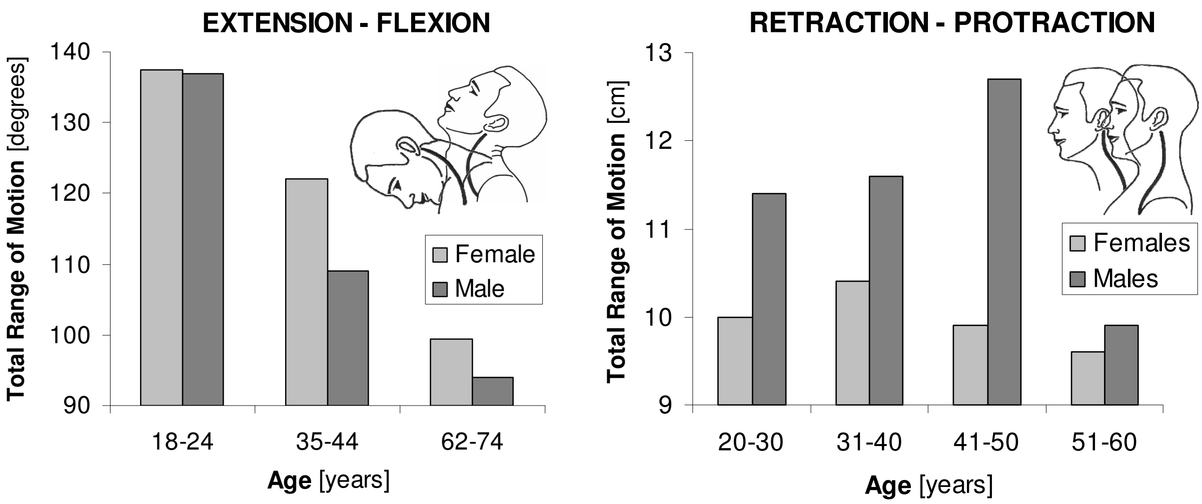


Figure 15. *a) The total range of extension–flexion motion of neck (measured in degrees) for different age categories. Based on Foust et al. (1973). b) The total range of retraction–protraction motion of neck (measured in cm) at different age categories. Based on Hanten et al. (1991). Females are represented by light gray bars and males by dark gray bars.*

1.4 REAR IMPACT DYNAMIC RESPONSE

During a rear impact the car is exposed to a forward acceleration causing a sudden velocity change. How this sudden increase of velocity of the vehicle will affect the motion of the head and neck of the occupant is illustrated in **Figure 16**.

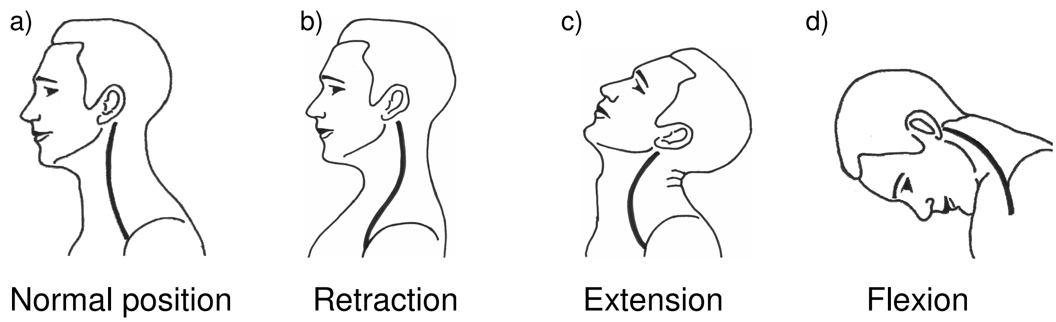


Figure 16. *The whiplash motion of the head and neck during a rear impact. a) Normal position b) Retraction of the neck d) Extension of the neck d) Flexion of the neck. The pictures are adapted from Linder (2001).*

In the normal position the neck has a slight curvature, a lordosis (**Figure 16a**). When the vehicle is pushed forward by the impacting car, the torso of the occupant will be pressed forward by the seatback while the head remains in the same position due to the inertia. This relative motion of the head and torso leads to a retraction of the neck. At the same time the natural curvature of the spine will be straightened, resulting in a contraction of the neck (**Figure 16b**). During the retraction phase the neck becomes exposed to significant mechanical loads before the head even reaches the head restraint. The retraction of the neck may be limited by the design and mechanical properties of the seatback and head restraint. As the torso of the occupant is pressed further forward, the head will be bent backwards and an extension of the neck will develop (**Figure 16c**). The presence of a well designed head restraint can prevent hyperextension of the neck, i.e. extension beyond its physiological limit. When the torso is pushed away from the seatback, the neck may be exposed to additional loads if the head lags behind. The forward motion of the torso is stopped by the seat belt, but the head will continue forward, resulting in a flexion of the neck (**Figure 16d**).

The acceleration of the body parts can exceed the acceleration of the vehicle (Severy et al. 1955; Eichberger et al. 1996; Szabo & Welcher 1996 among others). **Figure 17** shows an example from a volunteer test at 8 km/h (Siegmund et al. 1997). In this test the head acceleration was more than two times higher than the vehicle acceleration. Typically, there is a delay between the acceleration of the vehicle and the subsequent accelerations of the T1 and head. The vehicle acceleration starts first, and then the T1 acceleration, and finally the head acceleration, depending on the head/torso interaction to the head restraint and seatback.

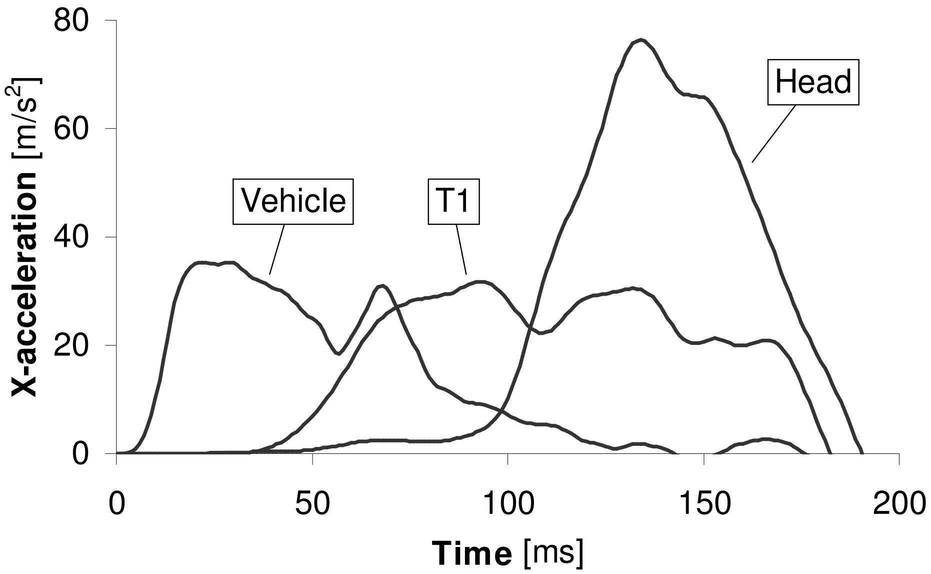


Figure 17. *The acceleration of the vehicle, T1, and head during a volunteer test in 8 km/h, data from Siegmund et al. (1997).*

1.5 INJURY MECHANISMS AND INJURY SITES

The term 'whiplash' is a description of the head/neck motion that causes neck injury, but is often used as a vague diagnosis for 'injury in the neck region'. Since whiplash injuries are located in the soft tissues of the neck, they are not possible to detect by diagnostic tools like X-rays or MRI (Magnetic Resonance Imaging). Due to the complicated structures of the neck, it is therefore difficult to decide the location of the injury and the cause of the symptoms.

There are several different theories for the injury sites and for the injury mechanisms, **Figure 18**. Possible injury sites may be facet joint, disc, muscle, ligament, artery, Central Nervous System (CSN), or dorsal nerve root ganglion, while the injury mechanisms may be abnormal vertebra motion, excessive neck loads, local hyperextension/flexion, or pressure pulses in the spinal canal.

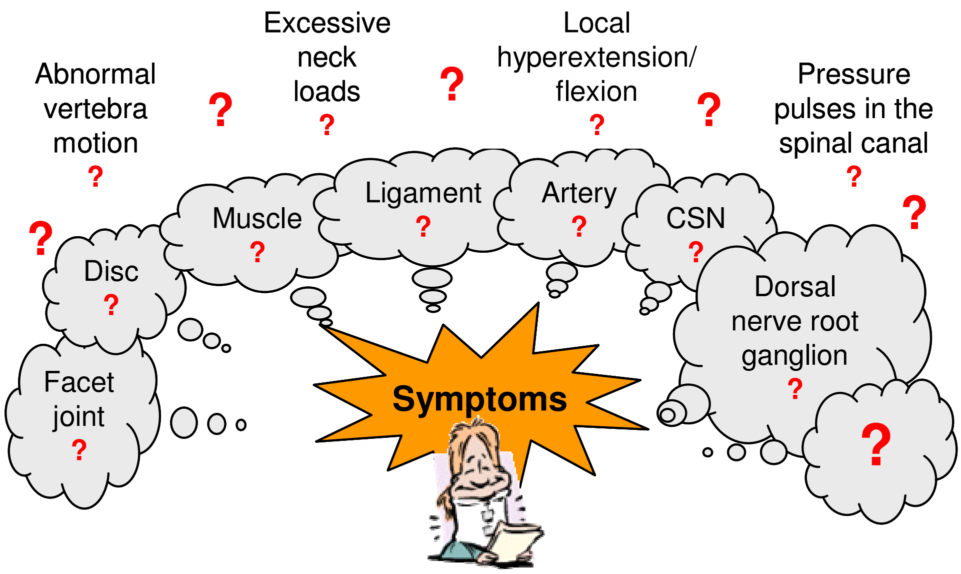


Figure 18. *Examples of possible injury sites and injury mechanisms. Based on a figure (lecture notes) by Johan Davidsson, Chalmers University of Technology.*

An early whiplash injury theory for rear impacts was hyperextension of the neck (**Figure 10**). Since the injury reducing effect of head restraints was limited (**page 5**), the research focus was shifted from neck hyperextension towards other possible injury mechanisms, mainly during the retraction (S-curvature) phase (**Figure 16b**). When the neck exceeds the physiological limit of retraction during rear impacts, the neck becomes exposed to significant mechanical loads. Since many head restraints were positioned too far behind the back of the head to prevent the neck from reaching maximum retraction, this was a possible explanation why head restraints did not offer better neck injury reducing effect. Studies have been reported that support this theory. For example, Mertz & Patrick (1971) reported a study with one volunteer exposed for substantial velocity changes and accelerations in several tests without initiating serious whiplash symptoms. This was possible since the volunteer was in contact to the seatback and head restraint from the beginning of the tests so that the retraction motion was minimized. Deng et al. (2000) performed rear impact PMHS tests using a high-speed X-ray to obtain cervical vertebral motions. Substantial facet joint strains were found before the head contacted the head restraint.

Aldman (1986) suggested that whiplash injuries may be caused by transient pressure gradients induced between the inside and outside of the spinal canal due the rapid motion changes of the head/neck during an impact. These pressure gradients may directly load the spinal nerve roots, potentially leading to whiplash-related symptoms like neck pain, headache, vertigo, blurred vision, and neurological symptoms in the upper extremities. Biological tests performed by Svensson et al. (1993a) and Örtengren et al (1996) supported this theory. Using Computational Fluid Dynamics (CFD) modelling, transient pressure patterns from earlier whiplash experiments on animals and PMHS were simulated with a suitable selection of the model flow properties (Svensson et al. 2009).

Two injury mechanisms of the facet joint have been proposed; pinching of the synovial fold (Ono et al. 1997; Kaneoka et al. 1999), and excessive strain of the capsule (Luan et al. 2000; Pearson et al. 2004). There are strong clinical evidence of facet-mediated neck pain (Barnsley et al. 1993; Bogduk & Marsland 1988; Aprill & Bogduk 1992) and Barnsley et al. (1994) claimed that cervical facet joints are the most common source of neck pain. Injuries to the neck ligaments and intervertebral discs in addition to the facet joints have been documented by magnetic resonance and autopsy studies of whiplash patients (Jonsson et al. 1991; Kaale et al. 2005). Ligament injuries may cause acute neck pain and lead to chronic spinal instability (Siegmund et al. 2009). Subfailure injuries of ligaments (spinal ligaments, disc annulus and facet capsules) may cause chronic back pain due to muscle control dysfunction (Panjabi 2006).

Altered blood flow due to spasm and/or narrowing of vertebral arteries in whiplash patients have been associated with chronic symptoms like headache, blurred vision, tinnitus, dizziness, and vertigo (Seric et al. 2000; Reddy et al. 2002; Linnman et al. 2009).

Symptoms from muscles are common among whiplash patients. It was suggested by Siegmund et al. (2009), that direct muscle injury may not be responsible for chronic whiplash pain, but may play an indirect role in modulating pain caused by injuries to other structures. Neck muscles potentially interact with other anatomical sites of whiplash injury in at least three ways: (1) neck muscles attach directly to the facet capsule, which has been implicated in chronic pain following whiplash; (2) neck muscle activation indirectly affects the loads and strains in other anatomicla structures; and (3) altered neuromuscular control may contribute to chronic pain via elevated and inapproprate muscle activation (Siegmund et al. 2009).

For a detailed review of the theories of whiplash injury sites and mechanisms, see Siegmund et al. (2009).

1.6 INJURY CRITERIA AND THRESHOLDS

An injury criterion is a function of physical parameters that can be measured in for instance a crash test dummy and that correlates with risk of injury for a certain body region. Generally, injury criteria are proposed and validated based on experimental studies and they are important tools for research, development, and evaluation of safety systems. Here follows a brief summary of the proposed neck injury criterion. For a more detailed description, see Schmitt et al. (2004).

The Neck Injury Criterion (NIC)

The Neck Injury Criterion (NIC) was proposed by Boström et al. (1996) and is based on the pressure gradient hypothesis formulated by Aldman (1986), and on the biological experiments by Svensson et al. (1993a) and Örtengren et al (1996). The NIC is calculated as

$$NIC = 0.2a_{rel} + v_{rel}^{2} \qquad \text{Eq (1)}$$

where a_{rel} is the relative horizontal acceleration between T1 and the occipital joint and v_{rel} is the horizontal velocity between T1 and the occipital joint. The NIC value is intended to be calculated at maximum retraction. In Boström et al. (2000) the NIC_{max} was proposed, which is the peak NIC value during the first 150 ms. The NIC formulated to be used for the Hybrid III dummy is denoted NIC50. The tolerance level for NIC, NIC_{max}, and NIC50 is 15 m^2/s^2.

Based on real-life accidents with crash recorders in combination with mathematical simulations, Kullgren et al. (2003) found that NIC_{max} is applicable to predict risk of whiplash injury when using a BioRID dummy. For NIC=15 a ~20 per cent risk of neck injury lasting more than 1 month was reported. Linder et al. (2004) reconstructed real-world rear impact crashes using sled tests with known injury outcomes in terms of neck injury symptoms of the front seat occupants. The results indicated that the risk for whiplash symptoms persisting more than one month was less than 10% for NIC_{max}<16.7.

The N_{ij} Criterion

The neck injury criterion, N_{ij}, was proposed by the US National Highway Traffic Safety Administration (NHTSA) to assess severe neck injuries in frontal impacts, including those with airbag deployment (DeSantis Klinch et al. 1996; Kleinberger et al. 1998). It combines the effects of force and moment measured at the occipital condyles and is based on both the tolerance levels for axial compression and bending moment. The Nij criterion is calculated by:

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \qquad \text{Eq (2)}$$

where F_z represents the axial force and M_y represents the flexion/extension bending moment. F_{int} and M_{int} are critical intercept values for the force and the moment, respectively. The intercept values for the 50th percentile Hybrid III male are proposed to be F_{int} (tension) = F_{int} (compression) = 4,500 N, M_{int} (tension) = 310 Nm, M_{int} (extension) = 125 Nm. Different intercept values are used for other dummy sizes. Four different load cases can be obtained; N_{te} for tension and extension, N_{tf} for tension and flexion, N_{ce} for compression and extension, and N_{cf} for compression and flexion. An injury threshold value of 1.0 applies for each load case (Schmitt et al. 2004).

The N_{km} Criterion

The N_{km} criterion (Schmitt et al. 2002) was derived to assess neck injuries in rear impacts. It is based on the N_{ij} criterion, and combines moments and shear forces. The N_{km} criterion is calculated by

$$N_{km} = \frac{F_x}{F_{int}} + \frac{M_y}{M_{int}} \qquad \text{Eq (3)}$$

where F_x represents the shear force and M_y the flexion/extension bending moment obtained from the upper neck load cell. F_{int} and M_{int} are critical intercept values for the force and the moment, respectively. The intercept values are F_{int} (anterior) = F_{int} (posterior) = 845 N, M_{int} (flexion) = 88.1 Nm, M_{int} (extension) = 47.5 Nm (Schmitt et al. 2002).

Four different load cases can be obtained; N_{fa} for flexion and anterior (positive) x-direction, N_{fp} for flexion and posterior (negative) x-direction, N_{ea} for extension and (positive) x-direction, and N_{ep} for extension and posterior (negative) x-direction. An injury threshold value of 1.0 applies for each load case.

Based on real-life accidents with crash recorders in combination with mathematical simulations, Kullgren et al. (2003) found that N_{km} is applicable to predict risk of whiplash injury when using a BioRID dummy. A ~20 per cent risk of neck injury lasting more than one month was reported for N_{km} =0.8. In sled tests, based on reconstructed real-world rear impact crashes with known injury outcomes, Linder et al. (2004) found that the risk for whiplash symptoms persisting more than one month was less than 10% for N_{km}<0.37.

The IV-NIC Criterion

The IV-NIC criterion, developed by Panjabi et al. (1999), is based on the hypothesis that a neck injury occurs when an intervertebral extension-flexion angle exceeds its physiological limits. It is defined as the ration of the intervertebral motion Θ_{trauma} under traumatic loading and the physiological range of motion $\Theta_{physiological}$. The IV-NIC is calculated by:

$$IV - NIC = \frac{\Theta_{\text{trauma,i}}}{\Theta_{\text{physiological,i}}} \qquad \text{Eq (4)}$$

There is no threshold value proposed for this criterion (Schmitt et al. 2004) and it can not be used in the existing crash test dummies.

The Neck Displacement Criterion (NDC)

The NDC, proposed by Viano & Davidsson (2001), is based on the angular and linear displacement response of the head relative to T1, obtained from volunteer tests. The criterion is given as corridors of the z- versus angular displacements, and x- versus angular displacements of the occipital condyle (OC) of the head relative to the T1. Working performance guidelines for the NDC in the Hybrid III and the BioRID P3 for low speed rear impacts are proposed in four different categories; Excellent, Good, Acceptable and Poor. Kullgren et al. (2003) found that NDC is less applicable to predict risk of whiplash injury when using a BioRID dummy. According to Schmitt et al. (2004) the discussions about NDC are still ongoing, and the corridors cannot be regarded as definitely set.

The Lower Neck Load Index (LNL)

The Lower Neck Load Index (LNL) (Heitplatz et al. 2003), takes into account three force components and two moment components measured at the lower neck. The LNL is calculated by:

$$LNL = \left| \frac{\sqrt{M_{y_{lower}}^2 + M_{x_{lower}}^2}}{C_{moment}} \right| + \left| \frac{\sqrt{F_{y_{lower}}^2 + F_{x_{lower}}^2}}{C_{shear}} \right| + \left| \frac{F_{z_{lower}}}{C_{tension}} \right| \qquad \text{Eq (5)}$$

M_i and F_i are the moment and force components, respectively. The intercept values are proposed to be C_{moment} = 15, C_{shear} = 250, and $C_{tension}$ = 900 for the RID dummy (Heitplatz et al. 2003).

1.7 WHIPLASH INJURY PROTECTION STRATEGIES

Improvement of the seat design is the most common way to increase the protection of the occupant from whiplash injury during a rear impact. The strategy is to minimize the relative motion to the head and torso, i.e. to reduce the relative motion between each spinal segment, and to reduce accelerations and rebound motion. This can be done by improved seat geometry and dynamic properties of the head restraint and seatback; by active devices that move in a crash as the body loads the seat; and by energy absorption in the seat. The protective performance of the seat can be seen in the injury statistics. Since 1997, more advanced whiplash protection systems have been introduced on the market. The most prominent, regarding whiplash injury reduction, are the Saab Active Head Restraint (SAHR), Volvo's Whiplash Protection System (WhiPS), and Toyota's Whiplash Injury Lessening (WIL) system. According to Kullgren et al. (2007) the relative risk of a sustaining whiplash injury leading to long-term symptoms is approximately 50 per cent lower in cars fitted with more advanced whiplash protection systems than in cars with standard seats launched after 1997. Compared with cars launched before 1997 with standard seats the difference is even higher.

Saab Active Head Restraint – SAHR

In 1997 (in US early 1998), Saab introduced the Saab Active Head Restraint (SAHR) system in the 9-5 model as a first application of crash activated systems to mitigate whiplash injuries. In addition to the active head-restraint, the SAHR system comprises design features in the seatback to control and distribute those loads on the occupant that are generated in rear impacts (Wiklund & Larsson 1998). The active head restraint is mounted to a pressure plate in the seatback by means of a spring-resisted link mechanism (**Figure 19**). When the seat pushes the occupant forward with more force than the spring can resist, the plate moves rearward into the seat. This forces the head restraint to move upward and forward, thus supporting the head before the relative motion between the head and the torso becomes significant (Wiklund & Larsson 1998). In 2002, the SAHR Generation II was introduced in the Saab 9-3 model. The main modification was that the pressure plate in the seatback was moved down to the lower back region, in order to receive an earlier motion of the head restraint.

Figure 19. *The Saab Active Head Restraint (SAHR) generation I (to the left) and generation II (to the right). Reprinted with permission by Stefan Olsen, Saab.*

The injury reducing effect of the SAHR system has been evaluated by Viano & Olsen (2001), Farmer et al. (2003), Kullgren et al. (2007), and Boström & Kullgren (2007) and ranges from 33 to 75 per cent in these studies (**Figure 20**).

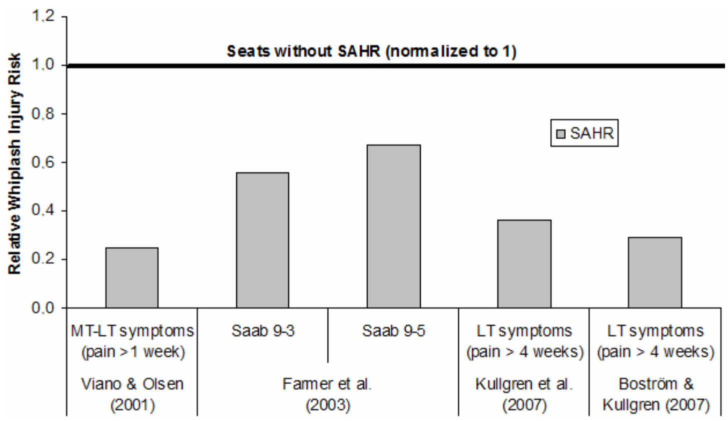


Figure 20. *The whiplash injury risk of the SAHR seat relative to seats without SAHR (normalized to 1). Based on data reported by Viano & Olsen (2001), Farmer et al. (2003), Kullgren et al. (2007, and Boström and Kullgren (2007).*

The effectiveness of the SAHR system with regards to males and females was evaluated by Viano & Olsen (2001). They wrote that "women experienced ST [*Short Term*] neck pain at a 52% higher rate than men, and the incidence was higher in vehicles equipped with SAHR. However, no women reported MT-LT [*Mid Term - Long Term*] whiplash injury in SAHR vehicles".

Whiplash Protection System – WhiPS

The Whiplash Protections System (WhiPS) was first introduced in Volvo cars in 1998, and was developed with the focus on three biomechanical guidelines (Lundell et al. 1998):

1. Reduce occupant acceleration.
2. Minimize relative movements between adjacent vertebrae and in the occipital joint, i.e. the curvature of the spine shall change as little as possible during the crash.
3. Minimize the forward rebound into the seat belt.

The main feature of the WhiPS is the recliner mechanism, enabling the seatback to move rearwards in relation to the seat cushion when loading an occupant during a rear impact (Jakobsson 2004). The normal occupant position is illustrated in **Figure 20a**. During a rear impact, the seatback first moves in a translational motion (**Figure 20b**) and then in a reclining motion (**Figure 20c**). During this motion, a deformation element in the recliner absorbs energy and thus reduces the occupant acceleration and forward rebound. In addition, the seatback was locally modified to give a uniform force distribution along the spine of the occupant; and the head restraint was modified to be positioned closer and higher relative to the head (Lundell et al. 1998).

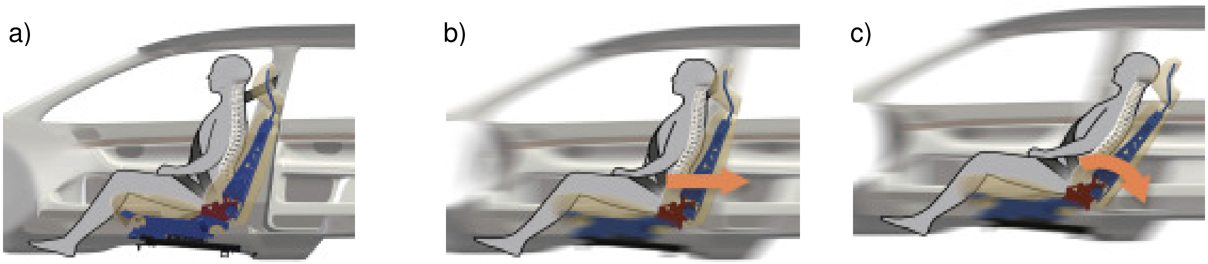


Figure 20. *Whiplash Protection System (WhiPS). a) Normal position b) Translational motion c) Reclining motion. Reprinted with permission by Ola Boström, Autoliv.*

The injury reducing effect of the WhiPS system has been evaluated by Farmer et al. (2003), Kullgren et al. (2007), Boström & Kullgren (2007), and Jakobsson et al. (2008) and ranges from 31 to 55 per cent in these studies (**Figure 21**).

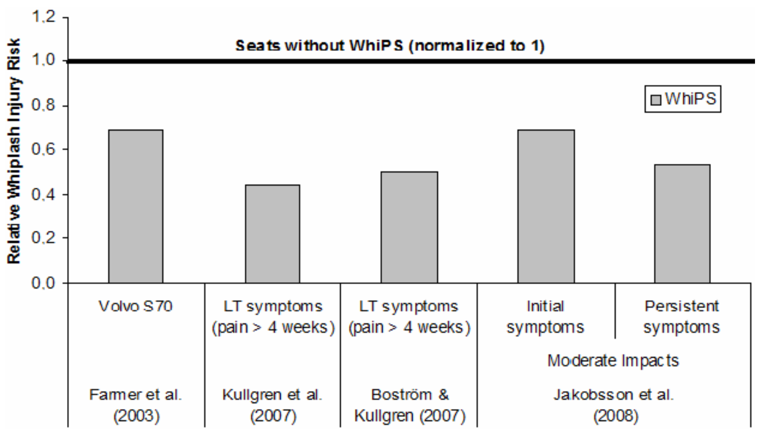


Figure 21. *The whiplash injury risk of the WhiPS seat relative to seats without WhiPS (normalized to 1). Based on data reported by Farmer et al. (2003), Kullgren et al. (2007), Boström & Kullgren (2007), and Jakobsson et al. (2008).*

The effectiveness of the WhiPS system with regards to males and females was evaluated by Jakobsson (2004). A 45 per cent reducing effect of initial symptoms for females as compared to 26 per cent for males was found in moderate impact severity (impacts in which the rear longitudinal members were deformed in any direction). The corresponding reduction of persistent symptoms (one year after the crash) was 67 per cent for females and 46 per cent for males.

Whiplash Injury Lessening – WIL

The Toyota Whiplash Injury Lessening (WIL) system has no active parts and is only working with improved geometry and softer seat back (Sekizuka 1998). The head restraint, especially the metal frame, has been moved forward and upward compared to previous seats. The upper part of the seatback frame has been moved rearwards, away from the upper part of the torso, but with remaining seat surface in order to support the upper part of the torso in the same way as previous seats. During a rear impact, the upper part of the torso sinks into the seatback while the stiffer head restraint meets the head of the occupant and reduces the whiplash motion of the head/neck. The pelvic support, at the lower part of the seatback frame, initiates the lower part of the torso to rebound first, and thereby helps to prevent neck extension.

The injury reducing effect of the WIL system has been evaluated by Farmer et al. (2003), Kullgren et al. (2007), and Boström & Kullgren (2007) and ranges from ~0 to 51 per cent (**Figure 22**).

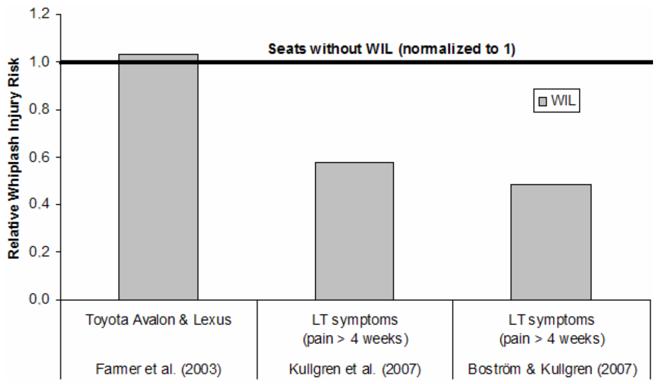


Figure 22. *The whiplash injury risk of the WIL seat relative to seats without WIL (normalized to 1). Based on data reported by Farmer et al. (2003), Kullgren et al. (2007), and Boström & Kullgren (2007).*

1.8 HUMAN MODELS USED FOR REAR IMPACT TESTS

Mechanical crash test dummies are used as human substitutes in crash testing at force levels most likely injurious for living humans. They are used in sled tests, or in full scale vehicle tests. The dummy should be sensitive to parameters that are related to an injury or an injury mechanism; it should be human-like in terms of size, mass and mass distribution; have a good repeatability; and it should be human-like in terms of the dynamic response during a crash. The validation of a mechanical dummy model is usually based on volunteer tests and/or post mortem test subject (PMHS) tests. Volunteer tests need to be carried out at very low, non-harmful velocities and accelerations, while PMHS tests can be performed at higher, injury inducing velocities and accelerations. However, the lack of muscle tone, internal pressure, and other changes in PMHS due to time after death makes the results less representative.

The Hybrid III dummy was developed for high-speed frontal crash testing, for evaluation of early automotive safety restraint systems. Reports covering the development process of the Hybrid III are collected in Backaitis & Mertz (1994). There are three sizes of Hybrid III adult dummies; the 5th percentile female (1.510 m, 46.82 kg), the 50th percentile male (1.751 m, 78.2 kg), and the 95th percentile male (1.873 m, 102.73 kg) (Schmitt et al. 2004). According to Schneider et al. (1983), a four-member dummy family was recommended to the NHTSA as optimal, including also the 50th percentile female. While the four-member family was approved by NHTSA, it was later determined that the level of funding would not allow completion of the study for all four dummy members, and the mid-sized female was dropped.

It is well established that the dynamic response of the the Hybrid III is not human-like in low-speed rear impact tests (Cappon et al. 2000; Scott et al. 1993 among others). Therefore, a new dummy, the Biofidelic Rear Impact Dummy (BioRID), was developed during the late 1990s. The size of the dummy represents a 50th percentile male (~1.77 m, 77.7 kg) and the mass distribution of the torso is human-like. The spine of the dummy consists of the same number of vertebrae as the human spine and have a lordosis of the neck and a kyphosis of the thoracic spine. The motion is restricted to the sagittal plane. The BioRID has been evaluated against data from volunteer tests (Davidsson et al. 1999), and PMHS test (Linder et al. 1999). Another low-speed rear impact test dummy, the Rear Impact Dummy version 3D (RID3D) was developed during the late 1990s. The RID3D is a modification kit (flexible spine and neck construction, a more realistic back shape, and the application of the ribcage design of the THOR dummy) for the 50th percentile Hybrid III male dummy (Cappon et al. 2000). Both the BioRID and the RID3D have been shown to be more biofidelic in low speed rear impact testing than the Hybrid III dummy (Davidsson et al. 1999; Siegmund et al. 2001; Philippens et al. 2002).

The 50th percentile male dummies roughly correspond to a 90th–95th percentile female in terms of stature and mass (Welsh & Lenard 2001), but not in terms of mass distribution and dynamic response. Hence, females are not well represented by the existing 50th percentile rear impact dummies, the BioRID and the RID3D. Only the extremes of the female population are accounted for by either the 50th percentile male dummy or the 5th percentile female dummy that may be used in rear impact crash tests (**Figure 21**). The 50th percentile male dummy is the most commonly used size during the development process of new seat concepts and design, for rear impacts. Consequently, the current seats are optimized to the 50th percentile male with no possibility to consider the female properties, in spite of the higher whiplash injury risk for females.

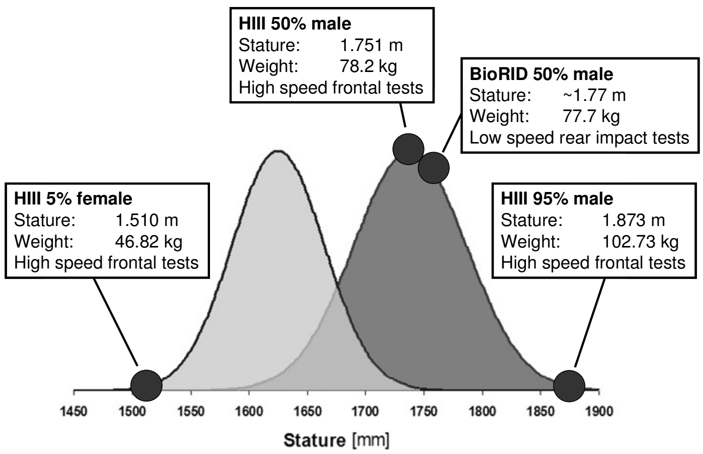


Figure 21. *The stature distribution of British male (shaded dark grey) and female (shaded light grey) car drivers in comparison to existing crash test dummies used for rear impact testing. The normal distributions of the statures are based on data from Pheasant (2006), page 57.*

Validated mathematical models are used in crash simulations as a complement to the mechanical models. There are two main types of mathematical simulation; the Multi Body System (MBS) and the Finite Element (FE) method. A multi body system is a system of rigid bodies connected by kinematic joints, as described in Schmitt et al. (2004). The motion of the system is analysed by subjecting the system to external loads. The main feature of MBS modelling is whole body kinematics. The FE method approximately solves a boundary value problem by dividing the geometry into smaller elements. With the FE method the deformations and stresses within the model as well as the kinematics are obtained. Contact interaction with interior and restraints are preferably simulated using the FE method.

Mordaka & Gentle (2003) developed a biomechanical FEM model of the 50^{th} and 5^{th} percentile female cervical spines based on a male model. The objective of that study was to see if females responded essentially as scaled-down males in rear impacts. It was found that detailed responses varied significantly with gender and it became clear that females cannot be modelled as scaled-down males, thus confirming the need for separate male and female biomechanical models. They also stated that there is a need for a revision of car test programmes and regulations which are currently based on the average male.

1.9 DYNAMIC RESPONSE OF FEMALES AND MALES

The dynamic response of both females and males needs to be established in order to understand the biomechanics that form the basis for whiplash injury. The primary source of such dynamic response data is from comparable volunteer tests with males and females. This data can be used as an input in the development process of improved occupant models such as computational models and crash test dummies. Today's dynamic response data from rear impacts, derived using volunteer tests, is dominated by male data. The comparisons between the female and male tests in these studies indicate differences in the dynamic response between males and females, but are inconclusive due to the lack of data and analysis of existing data. This is a summary of the results from rear impact volunteer tests including both males and females:

Szabo et al. (1994) performed rear impact car-to-car tests with two female and three male volunteers in a Ford Escort (model year 1982) at 8 km/h. The data shows that the females, compared to the males, on average had a:
- smaller rearward head x-displacement (females: 175 mm, males: 229 mm)
- smaller rearward shoulder x-displacement (females: 111 mm, males: 150 mm)
- smaller rearward head relative-to-torso angular displacement (estimated from graph)
- higher head x-acceleration (females: 12.5g, male: 10.1g)
- higher cervical x-acceleration (females: 6.5g, male: 4.5g)

In these comparisons, one test (Test 5 – subject C) was excluded due to intentionally increased initial head-to-head restraint distance.

Hell et al. (1999) performed rear impact sled tests with three female and 13 male volunteers in a "German standard car seat" at 6.5 km/h and 9.5 km/h. It was found that the females, compared to the males, on average had:
- smaller initial head-to-head restraint distance.
- "much larger forward flexion of the torso" in the rebound phase.
- higher and earlier head peak x-acceleration (in the order of 75% higher and 15 ms earlier at 9.5 km/h, and 100% higher at 6.5 km/h).
- higher and earlier T1 peak x-acceleration (in the order of 10–15% higher at 9.5 km/h).
- lower head angular peak acceleration "due to a relatively small head/head restraint distance".

Croft et al. (2002) performed rear impact car-to-car tests with one female and two male volunteers. It was noticed that the female, compared to the males had:
- higher maximum head x-accelerations.
- more pronounced "forward shear effect", as was her "forward bending moment".
- faster interaction with the seatback and head restraint and less resistance to the forward motion. This quicker interaction resulted in her earlier and higher amplitude acceleration.

For the males it was noticed that they had:
- greater resistance to the forward moving seat, effectively delaying their forward acceleration. This resulted in markedly reduced head linear accelerations.
- lower position of the head restraint relative to the head and a greater interaction with the seatback. This resulted in that the males "experienced markedly greater rearward phase extension and bending moments with corresponding less forward phase motion and bending moments".

Ono et al. (2006) performed rear impact sled tests with two female volunteers and four male volunteers at 6 km/h in a rigid seat with no head restraint. It was found that the females, compared to the males, on average had:
- greater cervical vertebral rotation angle, and their cervical spine exhibited a more significant S-shaped deformation.
- higher shear strain and compression strain of the front/rear edges of facet joints, indicating that females are at a higher risk of suffering neck injury.
- higher and (somewhat) earlier T1 x-acceleration peak.
- (somewhat) smaller, and earlier head angular acceleration, first peak.
- smaller and later head angular acceleration, second peak.
- earlier upper neck bending moment, first peak; the peaks were of the same magnitude for the females and males.
- smaller upper neck bending moment, second peak.

Schick et al. (2008) analyzed previously performed rear impact sled tests with eleven female volunteers and ten male volunteers at 9.5 km/h in a standard car seat. Data from Hell et al. (1999), see above, were included in this comparison. It was found that the females, compared to the males, on average had:
- smaller head-to-head restraint distance (females: 4.5 cm, males: 8.6 cm)
- higher head resultant acceleration peak (females: 12.7g*, males: 9.7g*)
- higher and earlier head x-acceleration peak (females: 8.2g/116 ms*, males: 7.3g/148 ms*)
- (somewhat) higher, and earlier T1 x-acceleration peak (females: 7.6g/87 ms, males: 7.2g/106 ms)
- higher and earlier thorax x-acceleration peak (females: 8.5g/93 ms*, males: 7.7g/109 ms*)
- smaller and earlier head angular acceleration, first peak (females: -513 rad/s/103 ms, males: -675 rad/s/126 ms)
- smaller and earlier head angular acceleration, second peak (females: 545 rad/s*/146 ms, males: 1063 rad/s*/160 ms)

** indicates statistical significance*

1.10 SUMMARY

It is well known that females have a higher whiplash injury risk in rear impacts compared to males. Studies have shown that there are differences between males and females in the anatomy and physiology of the neck, which might contribute to the higher whiplash injury risk for females. Studies have also indicated that there may be characteristic differences in the rear impact dynamic response between males and females. The 50^{th} percentile male dummy might thus limit the assessment and development of whiplash prevention systems that adequately protect both male and female occupants. Data from volunteer tests is needed to establish the dynamic response for females and males. Such data is fundamental for developing future occupant mathematical and/or mechanical models for crash safety development and assessment.

2. OBJECTIVES

The objective of this study was to quantify differences in dynamic response between average sized females and males in rear impacts. The work has resulted in two papers and the objectives of each paper are summarized beneath.

2.1 PAPER I

The objectives of Paper I were

1) To generate response corridors for the 50th percentile female based on the data set previously published by Siegmund et al. (1997) for rear impacts.
2) To compare response corridors for the 50th percentile female with the previously published corridors for the 50th percentile male in Siegmund et al. (2001) for rear impacts. Additionally, the NIC values, head-to-head restraint distances and contact times were to be compared for the 50th percentile male and female.

2.2 PAPER II

The objectives of Paper II were

1) To generate response corridors for the 50th percentile female based on data from rear impact sled tests with female volunteers.
2) To compare response corridors for the 50th percentile female with the previously published corridors for the 50th percentile male in Davidsson et al. (1998) for rear impacts. Additionally, the NIC values, head-to-head restraint distances and contact times were compared for the 50th percentile male and female.

3. METHOD AND MATERIAL (PAPER I & II)

3.1 PAPER I

The data used in Paper I was extracted from an earlier test series, originally presented in Siegmund et al. (1997). In that study, 42 volunteers - 21 males and 21 females - were exposed to rear impact car-to-car tests in 4 km/h and 8 km/h. From these data, response corridors were generated for a subset of 11 of the 21 male volunteers, representing the 50th percentile male, in Siegmund et al. (2001). In Paper I, an initial analysis has been performed with data from a subset of the female volunteers, representing a 50th percentile female.

Human Subjects

A subset of the females was extracted in a similar way as in the previous study for the males in Siegmund el al. (2001), and comparable response corridors of the females were generated. Of the original 21 female subjects, 12 (at 4 km/h) and 9 volunteers (at 8 km/h) were selected. These subjects were selected based on a stature range (156–167 cm) within ±5.5 cm of a 50th percentile female (161.5 cm; Diffrient et al. 1974). Their mass range (45.8–83.4 kg) varied between a 5th and 90th percentile mass for females of 50th percentile height (Najjar and Rowland 1987). The male corridors presented in Siegmund et al. (2001) were derived using data from eleven of the original 21 male subjects from Siegmund et al. (1997). These subjects were selected based on a stature range (173–178 cm) that was within ±3 cm of a 50th percentile male (174.7 cm; Diffrient et al. 1974). Their mass range (63–87 kg) varied between the 10th and 75th percentile mass for males of the 50th percentile stature (Najjar and Rowland 1987).

Test Procedures

The volunteers were seated in the front passenger seat of a 1990 Honda Accord LX four-door sedan. The rear of the Honda was struck by the front of a 1981 Volvo 240DL station wagon in the volunteer tests. The Volvo's impact speeds were selected to produce a change of velocity of about 4 km/h and 8 km/h on the Honda. The Honda's passenger seat was locked in the full rear position and the initial seatback angle was set to about 27 degrees from the vertical for all tests. The head restraint was locked in the full-up position. The volunteers were restrained by a lap and shoulder seatbelt and instructed to sit normally in the seat, face forward with their head level, place their hands on their lap, and relax prior to impact. The volunteers knew an impact was imminent but could not predict its exact timing.

Each volunteer underwent two tests; one each at a change of velocity of 4 km/h and 8 km/h. To minimize the effect of habituation, no practice or demonstration trials were given and the two tests were separated by at least one week. The volunteers were instrumented with accelerometers and video markers (**Figure 22)** as described in detail in Siegmund et al. (1997). The NIC value was calculated according **Eq (1).** Response corridors for the volunteers were defined as the average ± one standard deviation (SD) from the average response.

Figure 22. *The test setup of Paper I.*

3.2 PAPER II

A series of rear impact sled tests with female volunteers in 5 km/h and 7 km/h was performed, and the results were compared with previously performed sled tests with male volunteers under matching conditions (Davidsson et al. 1998).

Human Subjects

Eight female volunteers were recruited for the study. Their stature varied between 162–170 cm with an average of 166 cm, and their mass varied between 54–64 kg with an average of 60 kg. The male volunteers of Davidsson et al. (1998) had an average stature of 180 cm (at 5 km/h) and 182 cm (at 7 km/h), and an average mass of 79 kg at (5 km/h) and 77 kg (at 7 km/h). According to University of Michigan Transportation Research Institute (UMRTI), the stature and mass of the 50th percentile female is 162 cm and 62 kg respectively, while it is 175 cm and 77 kg for the 50th percentile male (Schneider et al. 1983). In comparison to the UMTRI data, the female volunteers in the present study were on average 2 per cent taller and 4 per cent lighter, while the male volunteers were on average 4 per cent taller and 1 per cent heavier.

Test Procedures

The volunteers were seated in a laboratory seat, which was placed on a still standing target sled that was impacted from the rear by a bullet sled **(Figure 23a**). The front structure of the bullet sled struck a deforming iron bar in a band brake device mounted on the target sled. This enabled tuning of the acceleration level of the target sled.

The volunteers were restrained by a lap seatbelt and instructed to sit normally in the seat, face forward with their head level, place their hands on their lap, and relax prior to impact. Each volunteer underwent two tests; first at a change of velocity of 5 km/h, and then of 7 km/h. The volunteers were equipped with a head harness which was fixed as tight as possible (**Figure 23b**). Accelerometers were placed at the same positions as in Davidsson et al. (1998). Tri-axial accelerometers were mounted on the left side of the head harness, and an angular accelerometer was mounted on the right side, approximately at the centre of gravity of the head on each side. Two linear accelerometers, in x- and z-direction, were placed on a holder that was attached to the skin close to T1. Linear accelerometers were placed on the bullet sled and on the target sled. A tape switch was attached to the steel bar on the target sled, and the signal from this switch defined the start of the impact, T=0. The motion of the volunteers was monitored with high-speed video at 1,000 frames per second.

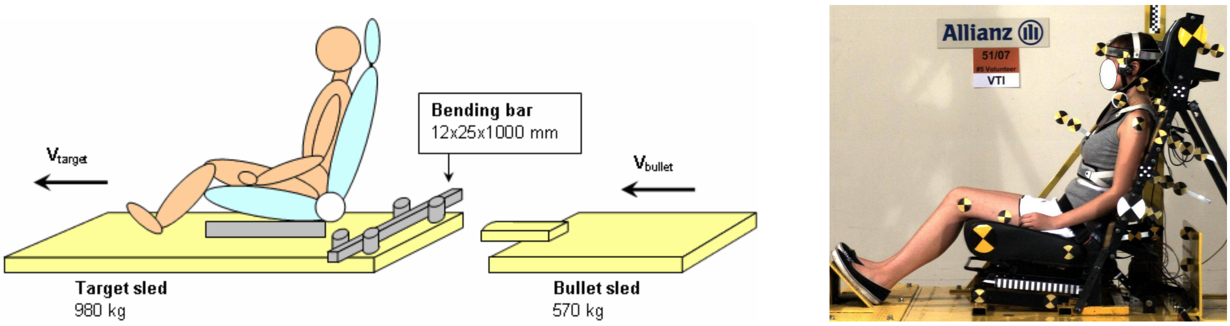


Figure 23. *Paper II: a) The test setup. b) The volunteer prepared for the test.*

Accelerations were determined from the transducer data. Linear and angular displacements were determined from the high speed video data, using TEMA 2.3 software. The horizontal displacement of the head and T1 were expressed relative to the sled coordinate system with the x-axis forward (in the sled travelling direction). The T1 displacement angle was defined as the angle between a film mark placed at the T1 and a film mark placed at the clavicle. The head-to-head restraint contact time was identified from film analysis and the head-to-head restraint distance was 1) measured before the test and 2) estimated from film analysis at T=0. The NIC value was calculated according to Eq (3), and the accelerations for the NIC calculations were filtered in CFC60. Response corridors for the volunteers were defined as the average ± one standard deviation (SD) from the average response.

4. RESULTS (PAPER I & II)

4.1 MOTION PATTERN - GENERAL DESCRIPTION (PAPER II)

The motion pattern of the female volunteers in Paper II, together with the average displacement peak values, is shown in Figure 24.

. During the first ~60 ms, the movements of the head and T1 relative to the sled were similar (between 0–20 mm at 5 km/h and 0–25 mm at 7 km/h). Initially, these motions were purely translational, but at ~40 ms the head and T1 started to rotate rearward.

At ~60 ms the x displacements of the head and T1 started to deviate due to the earlier interaction between the T1 and the seatback. This resulted in a more rapid rearward rotation of the T1 compared to the head. The T1 reached maximum rearward displacement ~15 ms before the head.

Until ~145 ms, the rearward rotation of the T1 continued, i.e after change of direction of the T1 x displacement relative to the sled. The rearward rotation of the head continued until 172 ms.

The head and torso moved forward relative to the sled from ~115 ms (for the T1) and ~130 ms (for the head). Maximum forward displacement was reached at ~360 ms (for the T1) and ~400 ms (for the head).

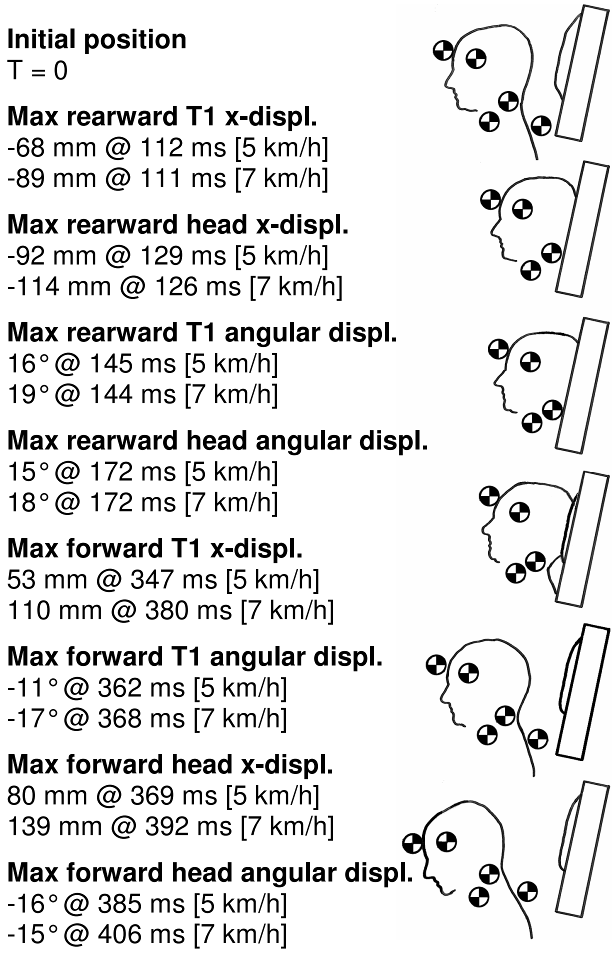


Figure 24. *Illustration of the motion pattern of the female volunteers during the rear impact tests. The average displacement peak values are listed to the left, in order of occurrence, at 5 km/h and 7 km/h in Paper I.*

4.2 X-ACCELERATIONS (PAPER I & II)

The head x-acceleration peaks were on average higher and had an earlier timing for the female volunteers compared to the male volunteers (**Figures 25a-d, Table 4**). In the tests with females, these peaks were on average 33 per cent (at 4 km/h) and 10 per cent (at 8 km/h) higher in Paper I; and 76 per cent (at 5 km/h) and 53 per cent (at 7 km/h) higher in Paper II. Correspondingly, for the females, these peaks occurred 7 per cent (at 4 km/h) and 9 per cent (at 8 km/h) earlier in Paper I, and 26 per cent (at 5 km/h) and 16 per cent (at 7 km/h) earlier in Paper II.

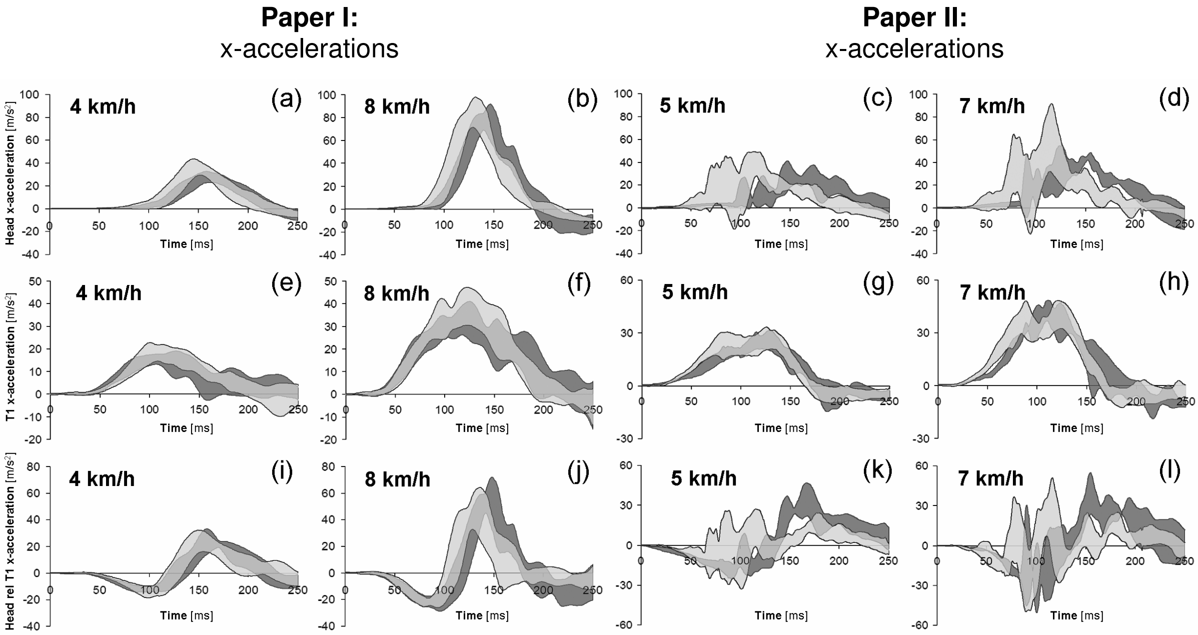


Figure 25. *The x-accelerations of the head, T1, and head relative to T1 at 4 km/h and 8 km/h (Paper I), and at 5 km/h and 7 km/h (Paper II). The head x-acceleration at (a) 4 km/h (b) 8 km/h (c) 5 km/h (d) 7 km/h; the T1 x-acceleration at (e) 4 km/h (f) 8 km/h (g) 5 km/h (h) 7 km/h; and the head relative to T1 x-acceleration at (i) 4 km/h (j) 8 km/h (k) 5 km/h (l) 7 km/h. The response corridors are calculated from the average ±one standard deviation (SD) from the average response. The response corridors for the males are shaded dark grey and the response corridors for the females are shaded light grey.*

Table 4 *The average x-acceleration peak amplitude and its occurrence of the head and T1 for the male and female volunteers at 4 km/h and 8 km/h (Paper I); and at 5 km/h and 7 km/h (Paper II).*

x-acceleration		**Paper I**				**Paper II**							
		4 km/h		**8 km/h**		**5 km/h**				**7 km/h**			
						1st peak		**2nd peak**		**1st peak**		**2nd peak**	
		[m/s²]	@ [ms]	[m/s²]	@ [ms]	[m/s²]	@ [ms]	[m/s²]	@ [ms]	[m/s²]	@ [ms]	[m/s²]	@ [ms]
Head	**Male**	30	159	84	140	29	112	39	157	44	102	51	145
	Female	40	149	92	128	51	83	49	115	67	86	75	122
T1	**Male**	18	122	39	124	24	95	30	126	40	92	48	116
	Female	22	115	45	123	27	84	31	128	39	86	45	121

The peak T1 x-acceleration was equal (at 7 km/h) or higher (at 4 km/h, 5 km/h, and 8 km/h), while the occurrence was similar (at 8 km/h) or earlier (at 4 km/h, 5 km/h, and 7 km/h) for the females compared to the males (**Figures 25e-h, Table 4**). In Paper I, the T1 x-acceleration peaks were on average 17 per cent higher and had a 5 per cent earlier timing at 4 km/h; and 16 per cent higher but with similar peak occurrence at 8 km/h, for the females compared to the males. In Paper II, the T1 x-acceleration had two positive peaks in the majority of the tests, at both 5 km/h and 7 km/h. At 5 km/h the first peak was 9 per cent higher with an 11 per cent earlier timing; while at 7 km/h the peaks were of the same magnitude with a 7 per cent earlier timing, for the females compared to the males.

4.3 LINEAR AND ANGULAR DISPLACEMENTS (PAPER II)

The head, T1, and head relative to T1 x-displacement peaks were on average smaller and earlier for the females compared to the males (**Figure 26**, **Table 5**). For example, at 7 km/h the x-displacement peak of the head was 25 per cent smaller and 13 per cent earlier, while the T1 peak was 14 per cent smaller and 11 per cent earlier, for the females. The resulting head relative to T1 x-displacement peak was 51 per cent smaller and 14 per cent earlier for the females compared to the males.

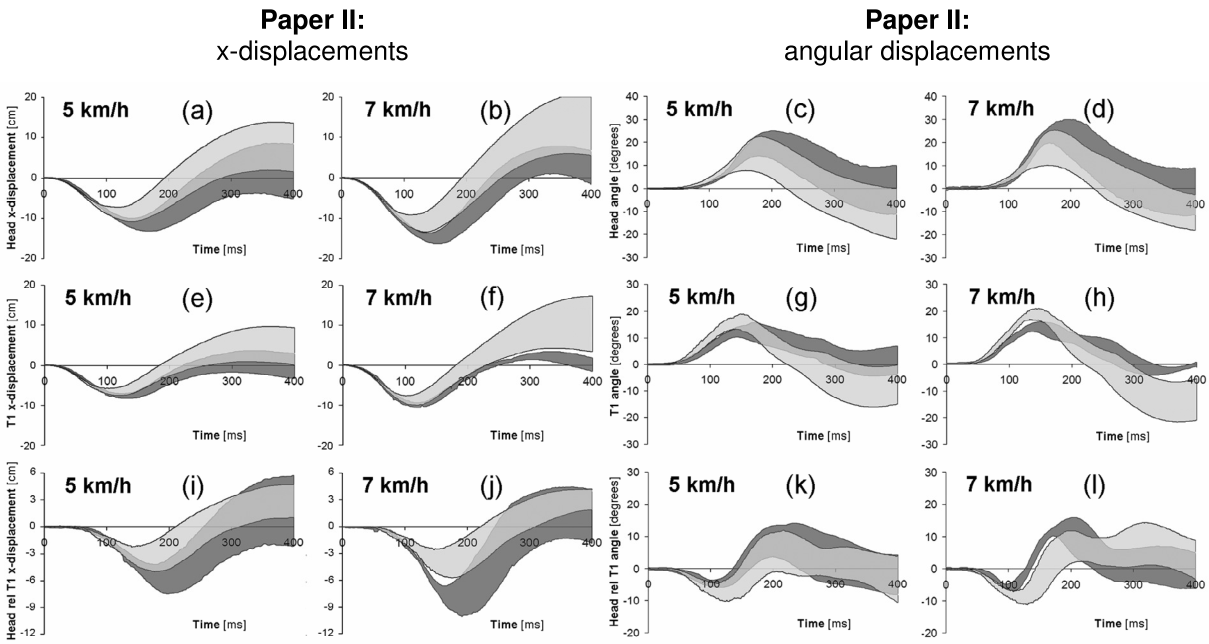


__Figure 26.__ The x-displacements (relative to the seat) and angular displacements of the head, T1, and head relative to T1 at 5 km/h and 7 km/h (Paper II). The head x-displacement at (a) 5 km/h (b) 7 km/h; the head angular displacement at (c) 5 km/h (d) 7 km/h; the T1 x-displacement at (e) 5 km/h (f) 7 km/h; the T1 angular displacement at (g) 5 km/h (h) 7 km/h; the head relative to T1 x-displacement at (i) 5 km/h (j) 7 km/h; the head relative to T1 angular displacement at (k) 5 km/h (l) 7 km/h. The response corridors are calculated from the average ±one standard deviation (SD) from the average response. The male response corridors are shaded dark grey and the female response corridors are shaded light grey.

The head angular displacement peaks were on average smaller and earlier for the females compared to the males, while the T1 and the head relative to T1 peaks were higher for the females (**Figure 26**, **Table 5**). At 7 km/h, the head angular displacement peak was 29 per cent smaller, while it was 33 per cent higher at T1, for the females compared to the males. The first (negative) peak of the head relative to T1 angular displacement was 46 per cent higher and 16 per cent later for the females at 7 km/h.

Table 5 The average linear and angular displacement peaks and their timing for the head and T1 for the male and female volunteers at 5 km/h and 7 km/h (Paper II).

Displacements		Paper II x-displacement 5 km/h [mm]	@ [ms]	7 km/h [mm]	@ [ms]	Paper II angular displacement 5 km/h 1st peak [°]	@ [ms]	2nd peak [°]	@ [ms]	7 km/h 1st peak [°]	@ [ms]	2nd peak [°]	@ [ms]
Head	**Male**	-118	155	-151	146	20.4°	193	-	-	25.1°	182	-	-
	Female	-92	129	-114	126	15.1°	172	-	-	17.9°	172	-	-
T1	**Male**	-79	131	-104	124	13.2°	154	-	-	14.3°	144	-	-
	Female	-68	112	-89	111	16.1°	145	-	-	19.0°	144	-	-
Head rel. T1	**Male**	-60	191	-84	188	-5.3°	106	10.2°	211	-6.5°	103	15.5°	185
	Female	-37	159	-41	162	-7.5°	115	7.7°	243	-9.4°	119	11.1°	262

4.4 THE NECK INJURY CRITERIA – NIC (PAPER I & II)

In Paper I, the average NIC value was similar for males and females at 4 km/h, whereas it was 20 per cent lower for the females (3.2 m^2/s^2) compared to the males (4.0 m^2/s^2) at 8 km/h (**Table 6**). In Paper II, the NIC value was on average 37 per cent lower and 33 per cent earlier at 5 km/h; and 45 per cent lower and 30 per cent earlier at 7 km/h, for the females compared to the males (**Table 6**).

Table 6 The average NIC value and its occurrence for the male and female volunteers at 4 km/h and 8 km/h (Paper I); and 5 km/h and 7 km/h (Paper II).

	Paper I 4 km/h NIC [m^2/s^2]	@ [ms]	8 km/h NIC [m^2/s^2]	@ [ms]	Paper II 5 km/h NIC [m^2/s^2]	@ [ms]	7 km/h NIC [m^2/s^2]	@ [ms]
Male	1.9	105	4.0	96	4.4	96	6.3	88
Female	2.0	107	3.2	90	2.8	65	3.4	62

4.5 THE HEAD-TO-HEAD RESTRAINT DISTANCE/CONTACT (PAPER I & II)

The initial head-to-head restraint distance was on average smaller for the females compared to the males, and there were large individual differences (**Table 7**). The initial head-to-head restraint distance was on average 9 per cent smaller in Paper I, and 30 per cent smaller and in Paper II, for the females compared to the males.

Table 7. The head-to-head restraint (HR) distance 1) pre-test (measured) and 2) at impact (estimated from film analysis) from Paper I and II.

		Paper I 4 km/h Mean [mm]	Range [mm]	SD [mm]	8 km/h Mean [mm]	Range [mm]	SD [mm]	Paper II 5 km/h Mean [mm]	Range [mm]	SD [mm]	7 km/h Mean [mm]	Range [mm]	SD [mm]
Pre-test	**Male**	-	-	-	-	-	-	93	70-140	26	86	70-120	19
	Female	-	-	-	-	-	-	62	20-120	28	62	20-110	25
At impact	**Male**	45	10-68	15	43	22-81	17	80	50-106	21	90	72-112	17
	Female	41	9-76	18	39	22-63	17	56	10-110	28	58	10-110	29

The female volunteers had and earlier head-to-head restraint contact time compared to the males (**Figure 27**). The contact time was on average 14 ms earlier at 4 km/h (Paper I); 11 ms earlier at 8 km/h (Paper I); 17 ms earlier at 5 km/h (Paper II) and 23 ms earlier at 7 km/h (Paper II) for the females compared to the males.

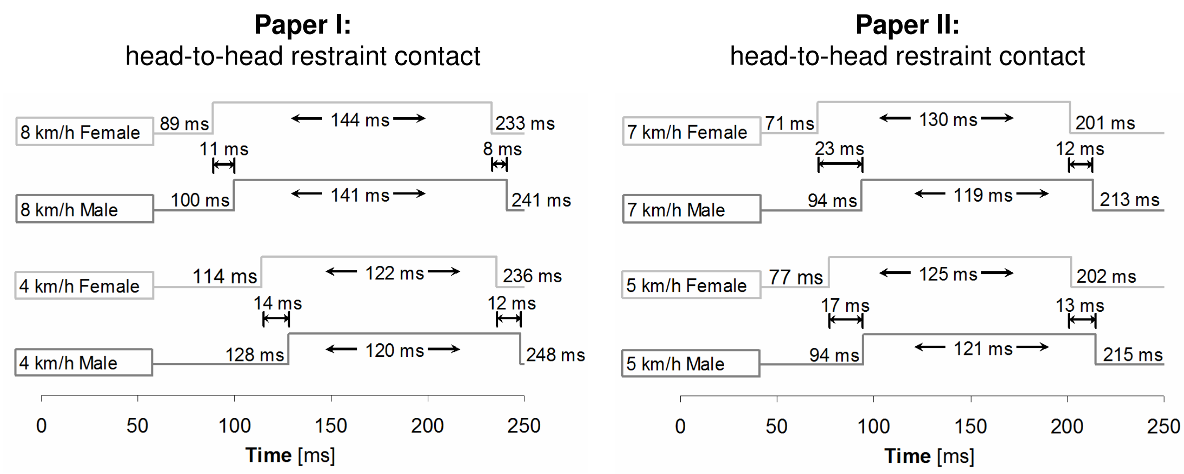


Figure 27. *The average head restraint contact times for the male and female volunteers at 4 km/h and 8 km/h (Paper I); and at 5 km/h and 7 km/h (Paper II).*

In Paper I it was found that for the same initial head-to-head restraint distance, head restraint contact occurred 11 ms and 7 ms earlier for the females than the males at 4 km/h and 8 km/h, respectively (**Figure 28**).

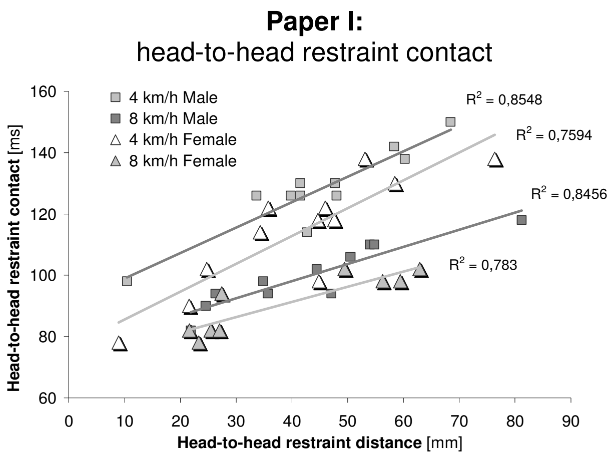


Figure 28. *The head-to-head restraint contact with regards to the head-to-head restraint distance for the male and female volunteers at 4 km/h and 8 km/h (Paper I).*

4.6 SUMMARY

The results from Paper I and Paper II are summarized in **Figure 29**. In the upper part of the figure the bars represent the *peak maximum* for the females relative to the males (normalized to 1). In the lower part of the figure the bars represent the *peak occurrence* for the females relative to the males (normalized to 1).

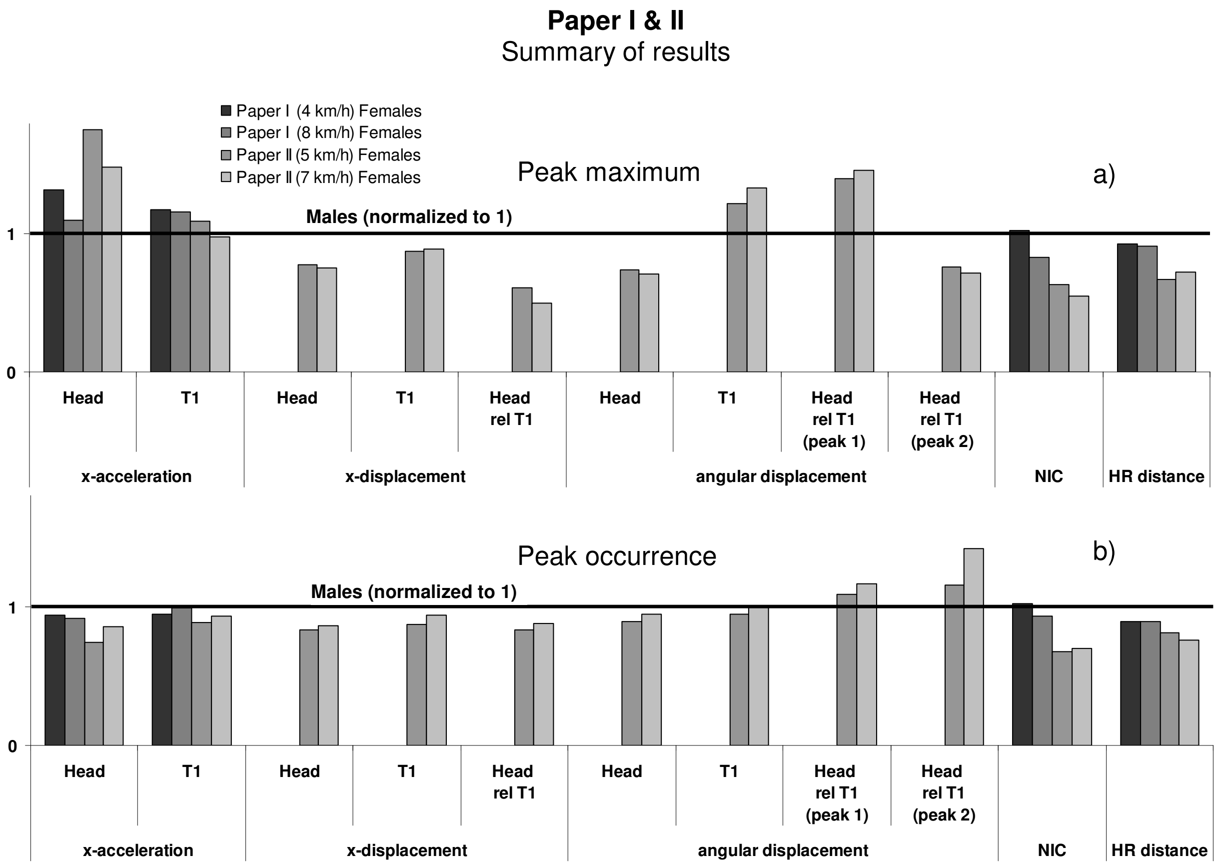


Figure 29. *Summary of results from Paper I and Paper II. The bars represent the females and the males are represented by a solid line (normalized to 1). a) The peak maximum for the females relative to the males. b) The peak occurrence for the females relative to the males.*

5. DISCUSSION

In this study, a new test series was conducted, and an earlier test series was further analysed with female volunteers in rear impact. The results were compared with previously published response corridors for male volunteers under matching test conditions (Davidsson et al. 1998; Siegmund et al. 2001). The overall result showed differences in the dynamic response of the females and the males in the head and T1 x-accelerations; linear and angular displacements; NIC value; head-to-head restraint distance; and head-to-head restraint contact time.

The initial head-to-head restraint distance was on average smaller for the females than for the males when seated in the same seat, especially in Paper II (28% and 33%), but also in Paper I (8% and 9%). Other studies have also reported a smaller head-to-head restraint distance for females (**Figure 6**) (Szabo et al. 1994 (estimation from graph); Minton et al. 1997; Jonsson et al. 2007; Schick et al. 2008).

The head-to-head restraint contact was on average earlier for the females than for the males. In Paper II, the main contributing factor for the earlier head-to-head restraint contact was the smaller initial head-to-head restraint distance for the females compared to the males. In Paper I, the females had a slightly smaller initial head-to-head restraint distance too, but it was shown that the females had an earlier head-to-head restraint contact even for the same initial head-to-head restraint distance. It was observed, from high-speed film analysis, that one contributing factor might be that the males have a generally higher seating height and hit the head restraint higher and at a different angle compared to the females. However, further analysis of the video data is needed in order to quantify these differences. Svensson et al. (1993b) observed that the head restraint moved rearward relative to the seat base as a result of the rearward seat back rotation as the occupant torso loaded the seat back during a rear impact. This resulted in a delayed head restraint contact. This effect may be smaller for the lighter and shorter females and would be another contributing factor to the earlier head restraint contact in the tests with female volunteers in Paper I. Earlier head-to-head restraint contact has also been shown in mathematical simulations with a 5^{th} percentile female model and a 50^{th} percentile male model (Viano 2003).

The peak head x-acceleration was on average higher and occurred earlier for the females compared to the males. Higher and earlier peak head x-accelerations have been reported also from other volunteer tests (Szabo et al. 1994; Hell et al. 1999; Croft et al. 2002; Schick et al. 2008) and from mathematical simulations (Mordaka & Gentle 2003). The earlier occurrence of the peak head x-acceleration in this study might depend on the earlier head-to-head restraint contact for the females. The higher peak head x-acceleration for the females may be due to differences in the head-to-head restraint interaction for the female and male volunteers. For example, smaller mass, different torso geometry, and/or another mass distribution of the females compared to the males may contribute to these differences. A mechanical or mathematical model of an average female would be a valuable tool in order to evaluate the seat interaction with regards to the female properties.

The peak T1 x-acceleration was equal (at 7 km/h) or higher (at 4 km/h, 5 km/h, and 8 km/h), while the occurrence was similar (at 8 km/h) or earlier (at 4 km/h, 5 km/h, and 7 km/h) for the females compared to the males. Higher and earlier peak T1 x-accelerations for females has previously been reported from volunteer tests (Hell et al. 1999; Ono et al. 2006) and shown in mathematical simulations with a 5^{th} percentile female model and a 50^{th} percentile male model (Viano, 2003).

The head, T1, and head relative to T1 x-displacements were smaller for the females compared to the males Paper II. The smaller head x-displacements might be due to the smaller head-to-head restraint distance for the females. The smaller T1 x-displacement for the females might be due to the smaller mass of the females compared to the males, i.e. the heavier males elongated the coil springs in the seatback to a higher degree.

<u>The head angular displacement was smaller</u> for the females compared to the males (Paper II). This might be due to the smaller initial head-to-head restraint distance and different geometry at the head restraint contact point for the females.

<u>The T1, and the head relative to T1 angular displacements were higher</u> for the females compared to the males (Paper II). The higher peak of the T1 and the head relative to T1 angular displacements for the females may be due to differences in seat interaction, or different shape and range of motion of the spine for the males and females.

<u>The NIC value was on average lower and occurred earlier</u> for the females compared to the males at 5 km/h and 7 km/h (Paper II), and at 8 km/h (Paper I), while it was similar for the females and males at 4 km/h (Paper I). The lower and earlier NIC value for the females might be due to their earlier head-to-head restraint contact and that the contact force between the upper part of the torso and seat structure peaked after the head-to-head restraint contact, resulting in a smaller relative acceleration between the head and T1 for the females. Keeping in mind the higher whiplash injury risk for females, one might ask whether the NIC threshold is different for females than for males. Further research is needed in order to address this question.

Assuming that head restraint geometry is less favourable with a large head-to-head restraint distance, it would be likely that the NIC value for the females would be higher than the NIC value for the males due to generally higher peak T1 x-accelerations for the females. Thus, if the head restraint provides good dynamic support during the crash, a lower NIC value may be expected for females compared to males, whereas if the head restraint would have provided poor dynamic support a higher NIC value may be expected for females compared to males. If so, females ought to be more affected than males by the position of the head restraint. Several studies have reported that improved head restraint geometry reduces the whiplash injury risk more for females than males (States et al. 1972; O'Neill et al. 1972; Thomas et al. 1982; Chapline et al. 2000; Farmer et al. 2003).

Geometrical differences between males and females - like stature, mass, and mass distribution of the torso - may contribute to the reported differences in dynamic response during a rear impact. The stature affects the geometry and thus the interaction between the head and the head restraint. A lower mass and/or a lower centre of mass may not only decrease the deflection of the seatback padding and springs, but also the seat frame due to a smaller lever with regards to the seatback pivot point. A smaller seatback deflection affects the plastic deformation and energy absorption and thus the dynamic head-to-head restraint distance and the rebound of the torso (Svensson et al. 1993b; Viano 2003).

There are also other physiological differences of the head/neck for males and females that may contribute to the differences in dynamic response and thus the higher injury risk for females. For example, females have lower strength in their neck muscles (Foust et al. 1973; Vasavada et al. 2001; Vasavada et al. 2008) and faster neck muscle reflexes (Foust et al. 1973). Females have smaller necks relative to their head size compared to males (States et al. 1972). Males have larger vertebral dimensions than females (DeSantis Klinich et al. 2004; Stemper et al. 2008); and larger segmental support area indicating a more stable intervertebral coupling (Stemper et al. 2008). Furthermore, differences in ligament structural components may lead to decreased stiffness in female spines (Stemper et al. 2008). It has been reported that the total range of extension–flexion motion is larger for females compared to males (Buck et al. 1959; Foust et al. 1973) and that the total range of retraction–protraction motion is smaller for females compared to males (in seated posture) (Hanten et al. 1991; Hanten et al. 2000). In dynamic tests, a more pronounced S-curved shape of the neck for females compared to males has been reported (Stemper et al. 2003; Ono et al. 2006). All of these pieces of information may add up and contribute to the reported differences in dynamic responses and injury risks in rear impacts for males and females.

Several studies have shown that the dynamic response of the BioRID 50th percentile male dummy is humanlike in low speed rear impacts (Philippens et al. 2002). The dynamic response of the BioRID dummy was validated with regard to male volunteer tests in Davidsson et al. (1999) - the same tests that the female volunteers in Paper II were compared with. The results from Paper II show that the female volunteers had a somewhat different dynamic response than the male volunteers and it is therefore important that new whiplash protection systems are developed and evaluated with the possibility to consider the female properties too. It is well established that females have a higher whiplash injury risk in rear impacts compared to males (**Figure 2**). Several studies have also indicated that there are differences between males and females in the anatomy and in the physiology of the neck, which may contribute to the higher whiplash injury risk for females. Therefore, testing with only a 50th percentile male dummy may limit the assessment and development of whiplash prevention systems that adequately protect both male and female occupants. Mordaka & Gentle (2003) concluded, based on mathematical simulations, that a "scaled down male model is not adequate to simulate female responses even though the scaling constitutes a good height and mass match". Additionally, Vasavada et al. (2008) found that "male and female necks are not geometrically similar and indicate that a female-specific model will be necessary to study gender differences in neck-related disorders". I.e a female model needs to be based on data from tests with females and therefore further volunteer and PMHS rear impacts studies need to be conducted in order to specify the characteristics of an average female and implement these data in models used for testing and evaluation. A model of an average female could, in addition to the existing average male model, complement the studies of Kullgren et al. (2003) or Linder et al. (2004) and might be used to define neck injury threshold values for females. By separating data for males and females it is possible that the neck injury threshold value for the males would be affected also.

6. CONCLUSION

Data from rear impact tests with average sized female and male volunteers have been analysed and compared. The results indicate that there might be characteristic differences in the dynamic response between males and females in rear impacts:

Compared to the males, the females had:

- a smaller initial head-to-head restraint distance
- an earlier head-to-head restraint contact
- a higher and earlier peak head x-acceleration
- an equal or higher peak T1 x-acceleration, with similar or earlier occurrence
- smaller head, T1, and head relative to T1 x-displacements
- smaller head angular displacement
- higher T1, and head relative to T1 angular displacement
- similar or lower NIC value, with earlier occurrence.

It is well established that females have a higher whiplash injury risk in rear impacts compared to males. The 50th percentile male dummy is validated with regards to tests with male volunteers and may thus limit the assessment and development of whiplash prevention systems that adequately protect both male and female occupants. It is therefore important that new whiplash protection systems are developed and evaluated with the possibility to consider the female properties too.

7. FUTURE WORK

In the future, this study can be used in the development and evaluation process of a mechanical and/or mathematical 50th percentile female dummy model for low-speed rear impact safety assessment. A validated 50th percentile female mathematical model may be the first step in this process. Such model can be used, not only as a tool in the design and development process of protective systems, but also in the process of further evaluation and development of injury criteria. By separating data with regards to females and males, the 50th percentile female and male models could be used to define different neck injury threshold values for females and males. A mathematical model would also provide knowledge and inputs to the process of establishing the need for a future 50th female mechanical model.

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