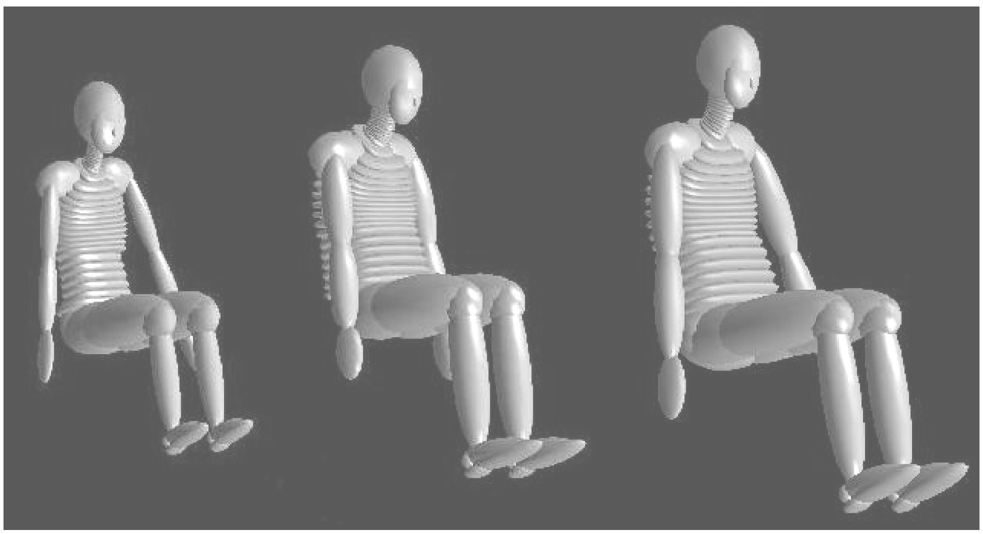
THESIS FOR THE DEGREE OF MASTERS OF SCIENCE IN AUTOMOTIVE ENGINEERING

Whiplash Injuries in Rear-end Car Impacts

"Influence of Seat Optimization Based on one Dummy Size on the Risk of Whiplash Injury for Different Size Occupants"



By

Ahtesham Asad Khan
Manuel Forero

Under the supervision of

Dr. Janusz Kajzer

Crash Safety Division
Department of Machine and Vehicle Systems
Chalmers University of Technology
Göteborg, Sweden, January 2005

ABSTRACT

In the current study, the influence of crash dummy size and weight on whiplash injury related parameters in rear-end car impacts and the influence of dummy size on the optimization of seat construction related to whiplash injury protection systems were investigated. For this purpose, computer crash simulations with MADYMO models were performed. The BioRID II 50th percentile male dummy MADYMO model was scaled to 95th percentile male and 5th percentile female dummies by using the MADYMO/MADYSCALE module, and a Toyota Yaris seat MADYMO model was used. Three crash pulses (high, medium and low severity) were applied. It should be noted that forward rebound and the use of seatbelts were not considered.

Automotive companies perform tests and computer simulations in order to optimize and design safety devices in their vehicles, including seats for rear impact protection. Generally, these tests are performed using only one dummy size which is considered to be more representative. This leads to uncertainties in the performance of rear impact protection devices when occupants are of different size. In order to analyze how the optimization of automobile seats based on one size dummy affects the risk of whiplash injury when occupants are of different size, the present study was divided in two procedures:

- investigation of the influence of crash dummy (BioRID-II) size and weight (95th percentile male, 50th percentile male and 5th percentile female) on whiplash injury related parameters in rear-end car impacts;
- investigation of the influence of crash dummy size and weight in optimization of seat construction i.e. analysis of the whiplash injury protection systems e.g., active head restraint system and seat back recliner system.

It is observed that the small female and a large male show the highest whiplash injury related parameter values. This is due to the very clear influence of the smaller mass of the small female. When the 95th percentile male dummy was used the values are higher than in the case of average male dummy, because of not so obvious but still significant reasons: less than optimum seat interaction and stiffer joints.

The lowest values of injury parameters are in the case when the 50th percentile male dummy was used. This means that the seat used in current study has been designed and optimized by the seat manufacturer for the average male dummy.

Active Headrest Systems decrease injury parameters for varying anthropometry even with the use of the 50th percentile optimized seat.

Seat recliner systems are more complex to optimize since each dummy size and crash pulse have to be considered independently.

NIC and Nkm are sensitive to active head rest systems (i.e., the whiplash injury protection systems that reduce the occupant relative velocity between lower neck and head). Both indicated the change in the whiplash injury risk parameter values by reduction of head to headrest contact time.

NIC is sensitive to seat recliner systems (i.e., the whiplash injury protection systems that try to control the occupant overall acceleration as well as the relative velocity between lower neck and head), however Nkm seems to be less sensitive to the variation of recliner stiffness. NIC indicated

higher risk for all occupant sizes when recliner stiffness was changed and Nkm indicated high risk for the small female only.

To analyze the activation characteristics of the whiplash injury protection systems for different body sizes, the influence of dummy size and recliner stiffness on seatback forces and moments has been examined. The 5th percentile female reaches its own maximum force value faster than the larger dummies. The 95th percentile dummy achieves the highest values, but these values are only slightly higher than in the case of 50th percentile dummy, due to the large male's poor interaction with the seat. The large male achieves a certain force and moment value faster than the smaller dummies. When increasing seat stiffness the force and moments increase, but the change in force is insignificant compared to change in the recliner stiffness.

These findings clearly demonstrate the need to consider anthropometry of subject in addition to seat and headrest characteristics in the assessment of rear impact injury, and in the optimization of seats and whiplash injury protection systems.

LIST OF SYMBOLS AND ABBREVIATIONS USED

NIC	-	Neck Injury Criteria
Nkm	-	Injury Criteria That Accounts Shear Forces And Bending Moments
Nij	-	Injury Criteria That Combines Upper Neck Forces And Moments
IV-NIC	-	Intervertebral Neck Injury Criteria
AIS	-	Abbreviated Injury Scale
BioRid II	-	Biofidelic Rear Impact Dummy
C1-C7	-	Cervical Vertebrae, Numbered From The Top Downwards
T1-T7	-	Thoracic Vertebrae, Numbered From The Top Downwards
Whiplash	-	Soft Tissue Neck Injury
Arel (max)	-	Maximum Relative Acceleration
Vrel (max)	-	Maximum Relative Velocity
Ax	-	Acceleration In The Linear (X) Direction
Tmax	-	Maximum Available Torque
SB dis	-	Seat Back Displacement With respect To H-Point In Degrees
Fmax SB	-	Maximum Force Available At The Seat Back
PPmax F	-	Maximum Force Available At The Pressure Plate
HR	-	headrest
delta-V	-	change in velocity

ACKNOWLEDGEMENTS

This study was proposed and supported by the Whiplash Commission, Sweden.

We would like to thank all those who helped us in making this work a success.

Dr. Janusz Kajzer, our supervisor and coordinator international masters program in Automotive Engineering at Chalmers University of Technology, for his valuable time and guidance.

Linda Eriksson, Industry Ph.D. Student, Crash Safety Division, Department Of Machine And Vehicle Design, Chalmers University Of Technology, for providing us with the mathematical model of the Toyota Yaris seat.

Mats Svensson, Associate Professor and Graduate Studies Director, Crash Safety Division, Department Of Machine and Vehicle Design, Chalmers University of Technology, for his guidance.

Jikuang Yang, Associate Professor, Crash Safety Division, Department Of Machine and Vehicle Design, Chalmers University Of Technology, for his guidance in relation to the dummy scaling techniques.

Marika Hedin, Head Secretary, Whiplash Commission Of Sweden, for her support.

Finally, all others from the Crash Safety Division for their help and guidance.

Ahtesham Asad Khan
Manuel Forero

Table of Contents

Abstract **ii**
List of symbols and abbreviations used **iv**
Acknowledgements **v**
Table of contents **vi**

1. INTRODUCTION …………………………………………………………….. 1

1.1. Whiplash Injury 1
1.1.1. Accident Data 1
1.1.2. Factors Influencing Whiplash Injury 3
1.1.3. After effects of Whiplash Injury 3

1.2. Anatomy of the Spine 4

1.3. Testing Methods in Rear Impact Neck Injury Investigation 5
1.3.1. Physical Testing Methods 6
1.3.1.1. Volunteer testing 6
1.3.1.2. Cadaver Testing 6
1.3.1.3. Animal testing 6
1.3.1.4. Dummies 7
1.3.1.4.1. Hybrid III 7
1.3.1.4.2. BioRID II 8
1.3.1.4.3. K-D Neck Model 9
1.3.2. Mathematical Models 10
1.3.2.1. BioRID II Mathematical Model 10

1.4. Injury Criteria 11
1.4.1. Biomechanical Neck Injury Predictor (Nij) 12
1.4.2. Nkm 12
1.4.3. Intervertebral Neck Injury Criterion 12
1.4.4. Neck Displacement Criterion (NDC) 12
1.4.5. Neck Injury Criterion (NIC) 12

1.5. Whiplash Injury Protection Systems 13
1.5.1. Toyota 13
1.5.2. Nissan 13
1.5.3. Volvo 14
1.5.4. Saab 16

1.6. Influence of Seatback Cushion on Neck Injuries 16

1.7. Influence of Crash Pulses on Neck Injuries 17

1.8. Dummy Scaling 17
1.8.1. Scaling Method 18
1.8.2. Influence of Dummy Size on the Estimated Risk of Whiplash Injury 19

1.9. Aim of the Study 19

2. METHODOLOGY…………………………………………………………….. 21

2.1. Modeling Requirements 21
2.1.1. Dummies 21
2.1.1.1. Scaling Procedure 21
2.1.2. Seat Systems 24
2.1.3. Crash Pulses 24

2.2 Simulation Procedure 25
2.2.1. Case 1 25
2.2.2. Case 2 26
2.2.3. Case 3 26
2.2.4. Case 4 27
2.2.5..Seatback forces and moment analysis 27

3. RESULTS AND DISCUSSIONS ………………………………………. 28

3.1. Case 1 28
3.2. Case 2 29
3.3. Case 3 32
3.4. Case 4 36

4. SEATBACK FORCES AND RECLINER MOMENT ANALYSIS ……… 38

4.1. Seatback Forces Analysis 38
4.2. Seatback Recliner Moment Analysis 40

5. CONCLUSIONS ……………………………………………………………42

6. GENERAL CONCLUSIONS ………………………………………………46

List of References 48

Appendix-I 53
Appendix-II 54
Appendix-III 57
Appendix-IV 58
Appendix-V 59

1. INTRODUCTION

To understand the methodology used in the current study, and the importance and magnitude of the problem treated here, it is necessary to know the background of rear-end impact injuries, the conducted research and research procedures that are being utilized to analyze whiplash injury for a single body size as well as for different body sizes. Once the problem is exposed, the aim of the present study can be defined.

1.1. Whiplash Injury

" ... is the acceleration-deceleration mechanism of energy transfer to the neck. The magnitude of the problem is great ... at least one percent of the entire population will experience chronic pain due to whiplash."
(Olson, L. American Physical Therapy Association)

The name Whiplash Injury derives from the etiopathogenic description of the sudden sharp whipping movement of the head and neck, produced at the moment of a traffic accident, particularly subsequent to collisions from the rear-end, head-on or side collisions. Whiplash Injuries are also called AIS 1 neck injuries.

In the case of a head-on collision, a forward displacement of the body is produced as a result of the inertia (weight of the body times speed), provoking tension upon the safety belt together with a neck hyperflexion followed by an hyperextension of it, thus producing "the whiplash".

In the case of a collision from the rear-end, the mechanism is inverted, first hyperextension followed by hyperflexion of the neck

Understanding whiplash injury has become a priority for researchers around the world. It is widely agreed that the cause is the differential whip-like movement of the head and neck relative to the torso, but still there is still debate about the causes of both long and short-term symptoms. While the hypotheses done so far have not been verified, it is agreed that there's more than one cause of the symptoms. It is also agreed that the term "whiplash", identifies a range of associated disorders.

1.1.1. Accident Data

Motivated by the need to address the problems that whiplash injury generate, researchers have conducted many studies to collect and analyze rear-end impact accident data. Some of the findings from these studies are presented in the coming sections. These findings illustrate why whiplash injury is such an important issue that affects all vehicle passengers, and needs the attention of all parties involved in automotive safety.

Volvo accident data indicates a rear-end impact neck injury risk which is approximately double the rate for frontal or side impacts. The frequency of different bodily injuries in rear-end impacts is shown in Figure 1.1. The graph is based on a subset of 605 belted drivers, in Volvo 700 and 900 models between 1985 and 1995 (Lundell et al., 1998). AIS 1 neck injuries are by far the most common injury type in rear-end impacts. Similar findings have been reported by Nygren, (1984).

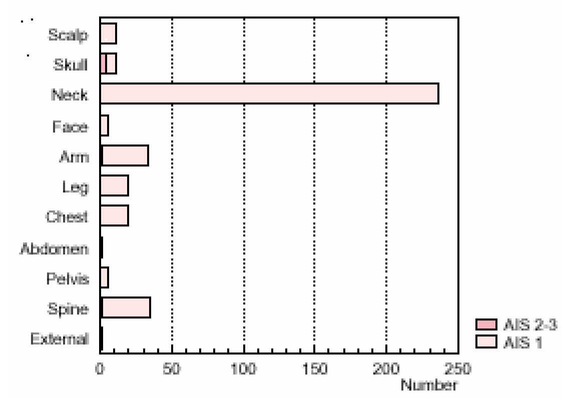


Figure 1.1: Injury distribution for rear-end collisions.

Neck injuries are reported at all impact speeds (Jakobsson, 1997 and Otte et al., 1997). From accident research as well as tests with volunteers, it is shown that people sustain neck injuries frequently even in impacts with very low severity (Olsson et al., 1990, Morris et al., 1996, Siegmund et al., 1997).

The U.S. National Highway Traffic Safety Administration has reported during 1995 on 5.5 million of American people involved in a traffic accident. Subsequent studies showed that 53% of them have suffered from whiplash injury.

In Germany, during 1992, 395,462 traffic accidents were registered from which 197,731 suffered from whiplash Injury.

Traffic accident statistics in Japan (Figure 1.2) show that rear-end impacts account for a large proportion of all injury-causing accidents, representing nearly 50% of the total. Over 90% of the injuries sustained by occupants whose vehicles are struck in rear-end collisions are to the neck region. More than 200,000 people suffer such injuries annually.

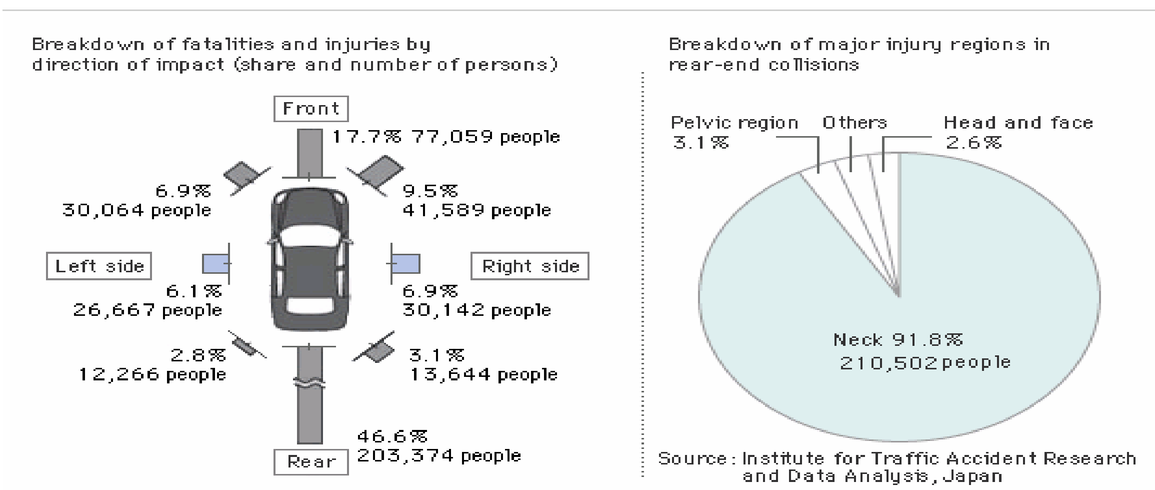


Figure 1.2: Traffic accident statistics in Japan

Little attention has been paid to variations on body size in the area of vehicle crash-safety design. For adults, regulations only establish testing with 50th percentile dummies. For frontal impact there are 5th percentile and 95th percentile Hybrid III dummies, and some studies have been done with these dummies for rear-end car impacts (DeRosia et al., 2002). Clinical data indicate that the female population is more vulnerable than the male population to sustain whiplash injury in rear-end car impacts (DeRosia et al., 2002).

1.1.2. Factors Influencing Whiplash Injury

To develop tools to prevent rear-end impact induced injury, it is important not only to know the occurrence of the injuries, but also to understand what factors influence the possibility of having whiplash injury. Extensive research has been made in this topic, and in this section the main factors found by researchers are described.

Studies carried out in different countries indicated the speed as one of the most important factors related to the occurrence of whiplash injuries, which are feasible from 20 km/h. Parkin et al., (1995) and Hell et al., (1999) found that AIS 1 injuries most frequently occur at delta-Vs below 30 km/h in the struck vehicle. Jakobsson et al., (2000) reported that AIS 1 injuries were found in minor or moderate crashes, i.e. crashes with the equivalent barrier speed (EBS) of 0-40 Km/h. Temming and Zobel (2000) reported that the risk of neck distortion injuries rose as the delta-V increased up to a limit of 25-30 km/h. They found at higher delta-V, the risk of neck distortion injuries decreased.

The risk of AIS 1 neck injuries has also been found to be influenced by head restraint position (Olsson et al., 1990, Chapline et al., 2000), seat back stiffness (Thomson et al., 1993, Parkin et al., 1995, Prasad et al., 1997) and the shape of the acceleration pulse on impact (Olsson et al., 1990, Kraft et al., 2002).

Moreover, the risk of AIS 1 neck injury in rear-end impacts has been shown to be influenced by height, gender, initial position and occupant's awareness of an impending impact. Lundell et al., (1998) and Temming and Zobel (2000) indicated that the AIS 1 neck injury risk increases with greater body height for both males and females. Rotated or inclined head position during an impact resulted in a higher incidence of persistent symptoms one-year after the collision (Sturzenegger et al., 1995).

Many epidemiological studies show that females have a greater risk of suffering AIS 1 neck injuries than males (Otremski et al., 1989, Morris et al., 1996, Kraft et al., 1997, Lundel et al., 1998, Lövsund et al., 1998, Jakobsson et al., 2000). Volvo accident data shows that medium height women are at the same level of risk as tall men (Lundel et al., 1998).

Another factor influencing the risk of neck injury in rear-end impacts is seating position of the car occupant. Volvo accident data reports a significantly higher risk of the driver sustaining a neck injury than the passengers (Lundel et al., 1998). They hypothesized that the differences between the driver and front seat passenger could be mainly due to different seating postures. Drivers are probable more prone to bend forward and away from the seat backrest and head restraint than passengers, who are more relaxed and probably more likely to rest their head against the head restraint. The relationship between increased distance to the head restraint and risk of neck injury has been shown, both in accident studies by Olsson et al., (1990) and Jakobsson et al., (1994) as well as in studies based on tests with volunteers by Deutscher, (1996).

1.1.3. After effects of Whiplash Injury

According to lesion topography, the whiplash trauma can be classified as follows (Otoophtalmical Neurophysiology, 2004):

Cervical Syndrome: Together with headaches, painful nape, restricted movements and muscular contractures, in an extreme case, it could produce torticollis.

Cervico-Brachial Syndrome: To the above-mentioned symptoms, it should be added sensitive disorders, loss of strength, paresthesias on shoulder and arm stretching towards the hand. This syndrome can be unilateral or bilateral.

Cervico-Medullar Syndrome**:** Lesions on the spinal cord are produced, which in line with its seriousness could turn into a commotion with temporary tetraparesis or, in the case of permanent section, they lead to tetraplegia

Cervico-Encephalic Syndrome**:** As a complement to the cervical syndrome symptoms, sharp headaches are added, also severe pain on the nape, tinnitus or ear buzzing, sensitivity to loud noises, vertigo, unsteadiness sensation, blurred vision, myodesopsias or photopsias (lights), equilibrium disorders, difficulties for concentrating and thinking, nauseas, vomits, etc.

In 80% of the cases there is a recovery of the cervicocephalic trauma, produced in different ways, from a few days of evolution up to two years. In the remaining 20%, the symptomatology continues as cervical, Cervico-Brachial, Cervico-Medullar or Cervico-Encephalic pains.

1.2. Anatomy of the Spine

It is important to consider the anatomy of the human spine, and its related terminology, which is used extensively by researchers of whiplash injury.

The normal anatomy of the spine is usually described by dividing up the spine into 3 major sections: the cervical, the thoracic, and the lumbar spine. (Below the lumbar spine there is a bone called the sacrum, which is part of the pelvis). Each section is made up of individual bones called vertebrae. There are 7 cervical vertebrae (C1-C7), 12 thoracic vertebrae (T1-T12), and 5 lumbar vertebrae (L1-L5).

An individual vertebra consists of several parts. The body of the vertebra is the primary area of weight bearing and provides a resting-place for the fibrous discs, which separate each of the vertebrae. The lamina covers the spinal canal, the large hole in the centre of the vertebra through which the spinal nerves pass. The spinous process is the bone one can feel when running hands down the back. The paired transverse processes are oriented 90 degrees to the spinous process and provide attachment for back muscles.

There are four facet joints (lumbar facet joints are shown in figure 1.3, since facet joints are easier to identify in the lumbar vertebrae) associated with each vertebra. A pair that face upward and pair that face downward. These interlock with the adjacent vertebrae and provide stability to the spine.

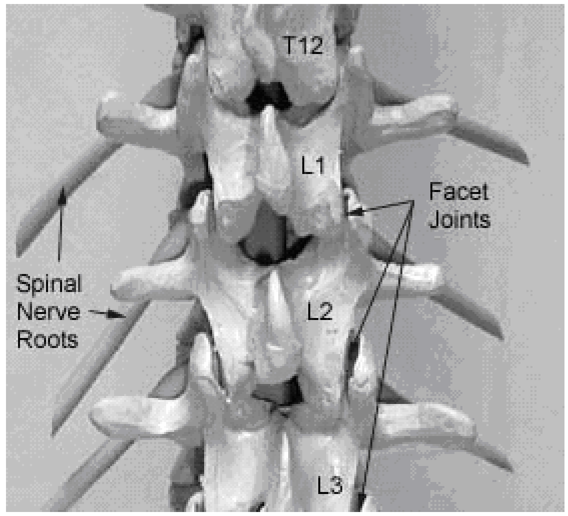


Figure 1.3: Spine lumbar posterior (Back view)

The vertebrae are separated by intervertebral discs, which act as cushions between the bones. Each disc is made up of two parts. The hard, tough outer layer called the annulus surrounds a mushy, moist centre termed the nucleus.

A very brief overview of the most common terms concerning the human has been presented here; if necessary, the reader is encouraged to refer to anatomy literature for a more detailed description.

1.3. Testing Methods used in Rear-end Impact Neck Injury Investigation

In addition to gathering accident data, there is an increasing need to understand and quantify the causes and consequences of rear-end impacts. In order to do this, several testing methods have been developed throughout the years. These methods have evolved with the advances of measurement techniques, instruments and information technology.

A brief overview of the various testing methods used for rear-end impact neck injury is given, in order to illustrate the importance and need of further research on the injury mechanisms of rear-end impact and the development of more biofidelic but still economically feasible models.

Testing methods for rear-end impact neck injury as well as any type of injury is a compromise between accuracy, costs, law and ethics. The ideal test for an engineer would be too expensive and/or would probably involve testing with live subjects or at least cadavers for maximum biofidelity, so other methods have been developed to avoid this, and still obtain accurate and consistent results.

Neck Injuries during rear-end impact have been of increasing importance since a few decades ago, when cars started to appear with standard headrests. Still, low speed rear-end impact injury is not considered as important as other types of injury since most of the time it is not fatal. Therefore, there is not much standard testing being done; for example, Euro NCAP does not have a standard evaluation method for whiplash injury. Current relevant regulation is based mainly in geometry and material properties of the head restraints (FMVSS, 2001).

Re-evaluation of the importance of whiplash injury has been done, and researchers have been working recently on the development of standardized procedures for evaluations of whiplash injury risk in a low speed rear-end impact (Muser et al., 2001). There have been a lot of refinements made on the traditionally used models, especially for the dummies, being simplicity, biofidelity, reproducibility, sensitivity, robustness and cost the main objectives.

1.3.1. Physical Testing Methods

When physical testing methods are considered, the first thing that comes to mind are dummies, but the ones that should be mentioned first are tests made with organic tissue, such as those experiments made on volunteers, animals, and cadavers. These tests are the basis for development of biofidelic dummies, and are necessary for identifying neck injury mechanisms, associated injury criteria and tolerance data. They represent the basis for definition of biofidelity performance requirements for mechanical neck simulators.

1.3.1.1. Volunteer testing

One of the main purposes of volunteer tests is validation of dummy models (Davidsson et al., 1999). These tests have been performed with crash pulses of very low magnitude, which are thought to be harmless for volunteers. Generally, healthy and young people are chosen for the procedure, to reduce injury risk and to be as constant as possible with the choice of subjects in order to minimize variation in the results.

Volunteers can only be used at very low impact speeds (for their protection); therefore impact accelerations subjected to volunteers are below the range where significant injury normally occurs. Volunteer tests are generally performed using a sled, which is accelerated by a pendulum impact. Accelerators and film targets are attached to record movement. These tests vary considerably from subject to subject, since the volunteers may not be relaxed prior to impact, and health conditions, level of relaxation and muscle tone vary considerably even among a homogeneous age group.

1.3.1.2. Cadaver Testing

Cadavers have been used to identify injury mechanisms that would be impossible to investigate by the use of volunteers. These tests could be performed at much higher accelerations, and measurement equipment could be added where it would be impossible to do with a live human, such as pressure sensors and accelerators. Tests have been done using whole bodies (Mertz and Patrick, 1971) but also have been executed using only the part of the body that the test is concerned with (Tencer et al., 2001).

Biofidelity is very high with the use of human cadavers, but there are many ethical, legal and social concerns with the use of cadavers, especially those of children. Another problem is the fact that many crash test laboratories are not well equipped to handle decomposing human tissue, which in turn is a sanitary hazard. Also, cadavers lack muscle tone, so when comparing to volunteer tests even if the volunteers appear completely relaxed, cadavers respond differently (Tencer et al., 2001).

1.3.1.3. Animal Testing

Since it is not possible to measure certain quantities directly (for example inserting pressure sensors inside the body) in volunteers, and cadavers don't share the same properties or biomechanical behavior of a live subject, animals have sometimes been used as an alternative. Use of animals for experiments is also a very sensitive subject, but it has been proven useful to find certain aspects of injury otherwise impossible to determinate. In a test done by Svensson, (1993), pigs were anesthetized and were subjected to a whiplash motion while measuring pressure in the Central Nervous System and also some pigs were histopathologically examined in the nerve-root region for signs of injury.

Use of animals has the advantage of being able to determine injury mechanisms in live subjects using potentially harmful acceleration pulses, but it has the disadvantage of lacking morphological biofidelity.

The combination of volunteer testing, cadaver testing and animal testing provide the guidelines for establishing injury criteria and the current models used to assess injury in crash situations. These tests are very difficult to execute because of their great variability, logistics and ethical implications, but are still being used as more biofidelity and accuracy of the models is required.

1.3.1.4. Dummies

Until very recently, rear-end impact tests were performed with dummies as the Hybrid III, developed for frontal impact testing. There have been significant advances in the development of dummies intended for rear-end impact being the current impact dummies much more biofidelic.

1.3.1.4.1. Hybrid III

The Hybrid III family of dummies consists of a 3-year-old, 6-year-old, 10-year-old, small adult female, mid-sized adult male and large adult male. These dummies are designed for use in frontal impact tests of automotive restraint systems. The 3-year-old, 6-year-old, small female and mid-sized male are currently specified by the National Highway Traffic Safety Administration for frontal impact compliance testing (Figure 1.4).

Figure 1.4: Hybrid III dummy and active headrest (GM, 2004).

The Hybrid III dummy has been used in some experiments to assess rear-end impact injury. The general conclusion is that the Hybrid III dummy behavior is not very biofidelic for rear-end impacts, because of the high stiffness of its neck and torso. These are the main problems concerning the Hybrid III dummy for rear-end impacts (Davidsson et al., 1999):

- T1 angular displacements are far too small.
- The occipital condyle's motion started too early and the durations are too short.
- The Hybrid III dummy's neck did not produce any S-shape motion.
- In the Hybrid III difference in distance between the iliac crest and T1 (caused by the straightening of the thoracic spine kyphosis) is not within the volunteer response corridors.

- The Hybrid III torso is too stiff and does not interact with the same compliant way as the human spine, therefore resulting in non-biofidelic head restraint forces and neck loads.
- The human head can move relative to the torso with very small stresses to the neck, but not in the Hybrid III dummy.

1.3.1.4.2. BioRID II

The first attempt to systematically develop a biofidelic dummy for rear-end impacts, the Biofidelic Rear-end Impact Dummy (BioRID), was done in the late 1990s by a consortium of Chalmers University of Technology in Sweden, restraint systems manufacturer Autoliv, and automakers Saab and Volvo. The dummy was designed to represent a 50^{th} percentile or average-size man, 1.77m tall and with a mass of 77 kg. The BioRID was developed to address the need for an entirely new dummy with an articulated spine able to reproduce the kinematics of an occupant during low speed rear-end impact. The latest model is the BioRID II dummy.

Before the current BioRID II dummy, there have been several models for a more biofidelic rear-end impact dummy. The first one was the so called RID-neck for use on the Hybrid III dummy. The Hybrid III dummy equipped with the RID-neck predicted more biofidelic behavior than the Hybrid III with the conventional neck. Even though this was a significant improvement, the Hybrid III torso-seat interaction left much to be desired, because of the Hybrid III torso's excessive rigidity. Because of this, it was suggested that a low-speed rear-end impact dummy should be fitted with an articulated thoracic spine (Davidsson et al., 1999, Viano et al., 2002). It also was suggested that the back shape should have an improved anthropomorphic shape in order to improve interaction with the seat back (Sekizuka, 1998).

The BioRID II dummy (Figure 1.5) is based on the Hybrid III dummy but fitted with an articulated spine and a soft torso. The spine consists of 24 vertebrae that have a curvature that resembles that of a human seated in a car seat. This is very important for obtaining a human-like motion, since it the model allows the spine to straighten during rear-end impact loading. In the thoracic and lumbar spine, steel pin joints are used as linear torsion springs. The cervical spine consists of 7 vertebrae (C2 to C7 are identical), and muscle substitutes, which are cables guided through vertebrae C1 to T3. At T3 the cable loads are transferred to springs which are mounted in parallel with a rotational damper.

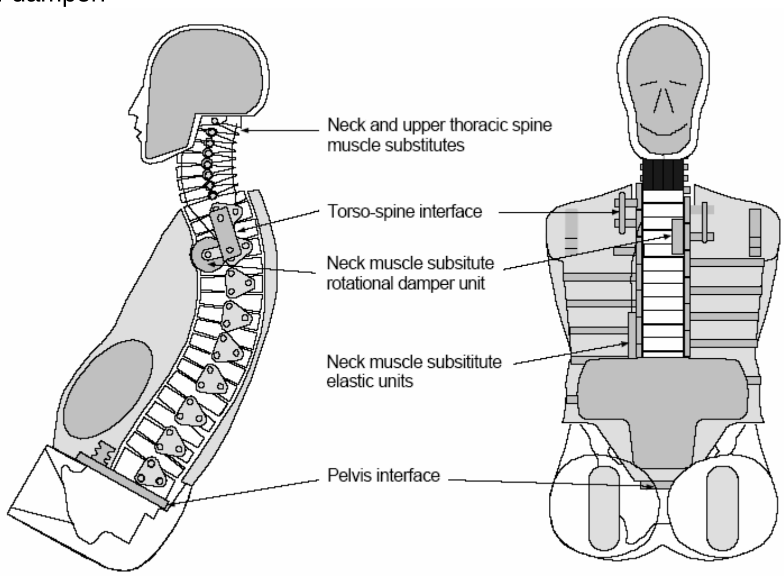


Figure 1.5: Schematic of the BIORID II torso, arm attachments, spine, modified Hybrid III pelvis and head (Svensson, 1999).

Performance and Validation Tests for the BioRID II.

The performance of the BioRID dummy was compared to those of volunteers and the Hybrid III dummy in rear-end impacts (Davidsson, 1999). He concluded that:

- In the validation tests, the BioRID II pelvis and T1 motion into the seat were close to those of the average volunteer in most occasions.
- Neck extension was also close to those of the average volunteer, while the Hybrid III neck extension was either too small, or started and peaked much too early.
- The pelvis z-displacements indicated that the BioRID dummy ramped up along the seatback in the same way as the volunteers when the tests were conducted with a flexible seat. The z-displacements were smaller than in the volunteers, but significantly larger than those of the Hybrid III dummy.
- The BioRID neck mimicked the complex volunteer responses (including the s-shape motions), but more tuning of the neck has been suggested in order to obtain more precise timing and amplitude. The Hybrid III neck responses were far from those of the average volunteer and no S-shape motion was observed.

The BioRID II dummy was also compared to in vivo and in vitro experiments concerning spine stiffness and RoM (Range of Motion), and the following was concluded (Davidsson, 2000):

- The BioRID II cervical and lumbar spine angular RoM was similar and thoracic spine RoM was larger than those presented in the literature.
- The BioRID II spine stiffness has been compared to those of in vitro data. The BioRID II spine is less stiff than in vitro data if factors are used to compensate for ribcage stiffness. The BioRID II upper torso stiffness is similar to those of humans.
- The stiffness of the BioRID II neck flexor muscle substitute is between those estimated in volunteers in static and dynamic neck extension.
- Little is known about RoM and stiffness of the human spine, specifically values for relative angular displacements and the effect of muscle forces on segment stiffness.

1.3.1.4.3. K-D Neck Model

As an attempt to obtain more biofidelity in the movement of the neck and head during low speed rear-end impacts, Tanaka et al., (2003) has developed an anatomically true model of the neck (Figure 1.6). This model is composed of cervical vertebrae, ligaments, intervertebral disks, and other soft tissues. They used an integrated polymerized material with properties that closely resemble the matching parts of the human tissue.

Figure 1.6: K-D Neck Model

Three-dimensional textiles are used for the ligaments; silicon rubber is used for substituting the intervertebral disks and finally the cervical ligaments are covered with colorless rubber.

The model has been validated against previous studies, using pendulum tests and measuring neck rotation angle and neck bending moment. It was found that the K-D model had more flexibility compared to the Hybrid III neck model.

The main drawback of their study is the fact that it has been made using the torso of the Hybrid III dummy, which as it has been said previously, is too stiff and does not mimic the back to seatback interaction in a biofidelic way. The Hybrid III is not recommended to be used for low speed rear-end impacts (Davidsson, 1999). Nevertheless, their model is a considerable improvement over the Hybrid III neck model.

1.3.2. Mathematical Models

Validated mathematical models, with very accurate results when compared to mechanical testing, are a useful tool for simulating physical tests that would require a considerably larger amount of time and financial resources. Mathematical models have the great advantage of being very easy to modify, which is very useful especially when only one dummy size is available. Scaling techniques can be applied in order to generate different size dummies.

Two modeling techniques usually are used for development of mathematical models of humans and dummies: Multi Body Systems (MBS) and Finite Element Modeling (FEM). MBS models often have fewer details and require less computer time than FEM models. MBS models are suitable for parameter studies, but do not model material characteristics and contacts as accurately as FEM models. There have been some MBS models for rear-end impact, but none in FEM so far.

For optimizing the effects of injury protection systems, mathematical modeling complements mechanical tests. It can also be used to verify proposed injury criteria and to evaluate the influences of risk factors on occupant kinematics. Therefore, biofidelity in mathematical models is needed. Some tests have been done to evaluate the influence of seat stiffness and seat geometry on neck loads (Shin et al., 2003), but these tests have been performed using a MBS model of the Hybrid III dummy, which lacks biofidelity for rear-end impact situations.

1.3.2.1. BioRID II Mathematical Model

In response to the need of biofidelity in mathematical models for rear-end impact testing, the BioRID II dummy was developed by Eriksson, (2000) as an MBS model in MADYMO (Figure 1.7). It is important to mention that this MADYMO model is a model of the mechanical dummy, not a model of a human. The model has been simplified compared with the mechanical BioRID II, but still has very similar properties.

As in its mechanical counterpart, the MADYMO BioRID II model consists of seven cervical, twelve thoracic, and five lumbar vertebrae. Spine joint to joint distance, range of motion of the joints, and spine curvature has been made the same as in the mechanical model.

The surface contour of the BioRID II was modeled by ellipsoids to imitate the one of its physical counterpart. Two ellipsoids are attached to each spine joint: one representing the vertebra and one representing the torso contour. Each shoulder was modeled with one ellipsoid and the head was modeled to imitate the contour of the head of the mechanical dummy.

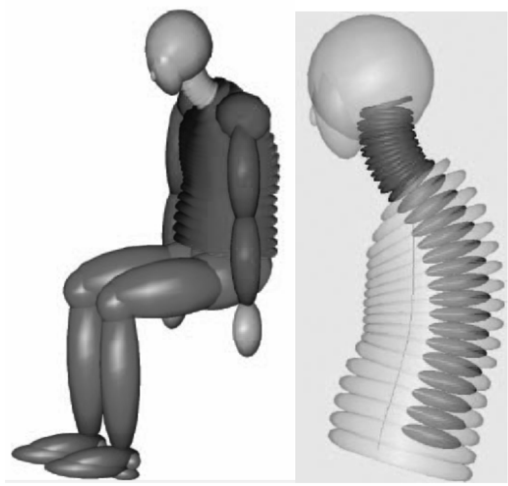


Figure 1.7: MADYMO BioRID II MBS model (left), and MADYMO BioRID II torso and head in sagittal plane (right) (TNO, 2004)

The static torque characteristics in the mathematical model are almost equal to those in the mechanical dummy. The cables used as muscle substitutes in the mechanical BioRID II neck were not modeled. Instead, the cable stiffness was added to the cervical vertebrae joints. This simplification is valid since neck hyperextension is not evaluated, due to the fact that modern head restraints limit head rotation. There is no hystheresis and no friction in the spine joints. The damping torque characteristics in the MADYMO spine joints were estimated to correspond to the damping of the polyurethane rubber blocks in the physical counterpart.

To model the silicon rubber characteristics of the BioRID II mechanical dummy torso, Kelvin elements (damper and spring in parallel) were placed in series along the ventral torso, connecting adjacent torso contour ellipsoids to each other. The Kelvin elements contribute to the general stiffness of the torso in the same proportion as the rubber does in the mechanical dummy.

The arms and legs of the Hybrid III dummy have been used for the MADYMO BioRID II dummy, with similar properties as in the mechanical Hybrid III. The total mass and the mass distribution of the mathematical model are the same as in the mechanical dummy.

Limitations of MADYMO model

Since in the mathematical BioRID II the forces from the cables in the neck of the mechanical BioRID II were lumped to the cervical spine joints, the MADYMO model is not valid for hyperextension of the neck. The forces from the cables mainly affect head relative T1 rotation, rather than rotations of individual joints. The lumping of these cable forces onto the cervical spine joints mainly affect rotations of individual joints.

The MADYMO BioRID II and the seats were not validated for upper neck loads, rebound phase and out of position postures.

1.4. Injury Criteria

Injury criteria are used for the evaluation of injuries. Several injury criteria for the neck region during rear-end impacts are proposed by the researchers. These injury criteria evaluate injuries during different stages of the whiplash motion, and are based on different injury mechanisms. The most common and widely accepted injury criteria are described here.

1.4.1. Biomechanical Neck Injury Predictor (Nij)

The National Highway Traffic System Administration (NHTSA 2004) has proposed a new neck injury criterion called Nij. This method combines neck axial tension/compression and neck moments (flexion/extension) into one Injury Criteria Performance Limit (ICPL). This criterion employs the summation of normalized neck axial force and normalized neck moment.

The formulation is:

$Nij = F_{NZ} + M_{NY}$,

Where:

$F_{NZ} = F_Z / F_{Z\ CRIT.}$
and $M_{NY} = M_Y / M_{Y\ CRIT.}$

F_Z = neck axial load (tension or compression).
$F_{Z\ CRIT}$ = critical force (in tension is 6806 N, and compression is -6160 N for the Hybrid III, male 50%).
M_Y = neck bending moment (flexion or extension).
$M_{Y\ CRIT}$ = critical moment (in flexion 310 Nm and extension -135 Nm for the Hybrid III, male 50%).
Nij can not exceed 1.4. (The agency is also considering N_{ij} = 1.0 as an alternative).

1.4.2. Nkm

Yang et al., (1997) based on experiments with cervical vertebra specimens, suggested that axial compression/tension forces together with shear force are responsible for the higher frequency of AISI 1 neck injuries observed in rear-end as well as frontal impacts. Partly based on these suggestions, Schmitt et al., (2001 & 2002) proposed a modification of N_{ij} for AIS 1 injuries in rear-end impacts, called Nkm. Nkm takes into account shear forces and bending moments at the occipital condyles and is suggested for evaluating possible mechanisms in the flexion phase of a rear-end impact. The limit tolerance value for this injury criterion is 0.3.

1.4.3. Intervertebral Neck Injury Criterion (IV-NIC)

Panjabi et al., (1999) hypothesized that a neck injury occurs when an intervertebral rotation exceeds its physiological limit. It is defined as the ratio of the intervertebral motion T_{trauma} under traumatic loading and the physiological range of motion $T_{physiological}$. The IV-NIC is calculated by:

$$IV\text{-}NIC = T_{trauma} / T_{physiological}$$

It has not yet been validated.

1.4.4. Neck Displacement Criterion (NDC)

NDC was proposed by Viano and Davidsson, (2002) based on rear-end impact volunteer tests, and consider the angular and linear displacement response of the head relative to the lower neck.

1.4.5. Neck Injury Criterion (NIC)

NIC was proposed by Boström et al., (1996) and is based on the injury mechanism theory of Aldman (1986) and findings of Svensson et al., (1993) and Örtengren et al., (1996). This criterion is based on the relative velocity and acceleration between the upper and lower neck.

$NIC = 0.2 \times a_{rel} + v_{rel}^2$

Where,

a_{rel} = relative acceleration between head and T1
v_{rel} = relative velocity between head and T1

The relative acceleration and velocity between the lower and upper neck used in NIC indicate an occurrence of injury in early stage of the forward motion of the torso relative to the head. Boström et al., (1996) proposed that NIC values lower than $15m^2/s^2$ do not result in soft tissue neck injuries.

In order to verify/falsify NIC as a criterion for Whiplash Associated Disorders (WAD 1-3), Eichberger et al., (1998) and Wheeler et al., (1998) performed volunteer studies. These studies verified the relevance of NIC under conditions that do not lead to long-term symptoms.

NIC, Nkm, NDC and lower neck moment were evaluated in a recent rear-end study by Kullgren et al., (2003) using a MADYMO model of the BioRID II dummy and real life crash pulses. Nkm and NIC were found most appropriate.

1.5. Whiplash Injury Protection Systems

Realizing the importance of reducing whiplash injuries, most of the automotive companies like Volvo, Toyota, Saab, Mazda, Nissan, Audi, VW, Renault, Opel, and Ford have developed whiplash injury protection systems for their vehicles. A lot of other companies and suppliers are on the way to develop or improve whiplash injury protection systems. In the present study, the risk of injury for different size occupants will be evaluated for the optimization of two of the most representative whiplash protection systems: seatback recliner and active headrest. The whiplash protection mechanisms developed by automotive manufacturers work using mainly either the seatback recliner or the active headrest principle, so it is necessary to have a notion of what is available in the market for rear-end impact protection to be familiarized with the systems' working fundamentals.

1.5.1. Toyota

Toyota front seats incorporate a "Whiplash Injury Lessening" (WIL) (Toyota, 2004) designed to minimize the risk of whiplash injuries in low-speed rear-end car collisions. WIL seatbacks frames are designed to yield in a controlled fashion in rear-end crashes to reduce the forward acceleration of occupants' torsos. This design feature helps lessen the differential motion of head and torso, which is the cause of whiplash injury. WIL seats were first developed using computer aided design, and then tested in real life crash situations.

Folksam (a Swedish Insurance Institute) in collaboration with Vägverket (Swedish National Road Administration) conducted some crash tests for the assessment of whiplash injury protection systems in rear-ends car impacts (Kraft, 2004) and the WIL system was catalogued as a Medium Risk system.

1.5.2. Nissan

The Active Head Restraint (Nissan, 2004) uses the force of the occupant's body against the seatback in a rear-end collision to move the head restraint forward instantaneously to support the head, thereby helping to reduce the impact to the neck of a front-seat occupant. The mechanism of whiplash injuries closely involves two factors resulting from the impact: the force acting to bend the neck backward and the force that causes the head to tilt rearward. Because the Active Head

Restraint is effective in controlling these two factors, it can help reduce the load on the neck at the moment of the collision.

This whiplash injury protection system was awarded as Low Risk System by Kraft, (2004).

1.5.3. Volvo

'WHIPS' – the Volvo Whiplash Protection System, first introduced at the launch of the S80 in 1998, and is now standard in all Volvo front seats (Lundell et al., 1998).

WHIPS include two components designed to limit the sudden differential motion of the head and neck. The unique feature is a recliner at the bottom of the seatback on each side that allows the seat backrest to move rearward to reduce the forward acceleration of the torso. The head restraint, which is positioned high and close to the back of the head, catches the head so that it moves forward with the torso. The combination of the slower torso acceleration and the head restraint catching the head early in the sequence means the neck changes shape less, and the change occurs more slowly than with a conventional seatback/head restraint. The result is that a whiplash injury is less likely to occur.

Recliner Design - The WHIPS Function

The recliner is the part of the seat by which the backrest is attached to the seat base. The basic function of a recliner is to facilitate adjusting the reclining angle of the backrest. In Volvo seats, there are two recliners to each seat, one on each side. In the WHIPS recliner, an impact-activated function is added.

The WHIPS recliner unit consists of two main parts (Figure 1.8): the mechanism for adjusting the static reclining angle (A) and the WHIPS system (B). These two parts are combined to form the complete WHIPS recliner unit.

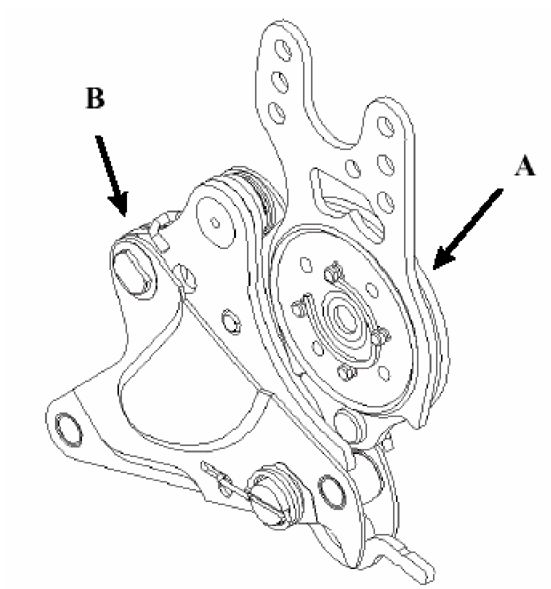


Figure 1.8: **The WHIPS recliner**

The motion of the seat may be divided in to two phases, as shown schematically in Figure 1.9.
In a rear-end impact, the seat is accelerated forward with the car. Due to the inertia of the occupant, the back of the occupant is then pressed into the seat. When the forces from the

occupant acting upon the seat backrest exceed a certain level, the WHIPS system will be activated. Hence no external sensor system is needed to activate the WHIPS system.

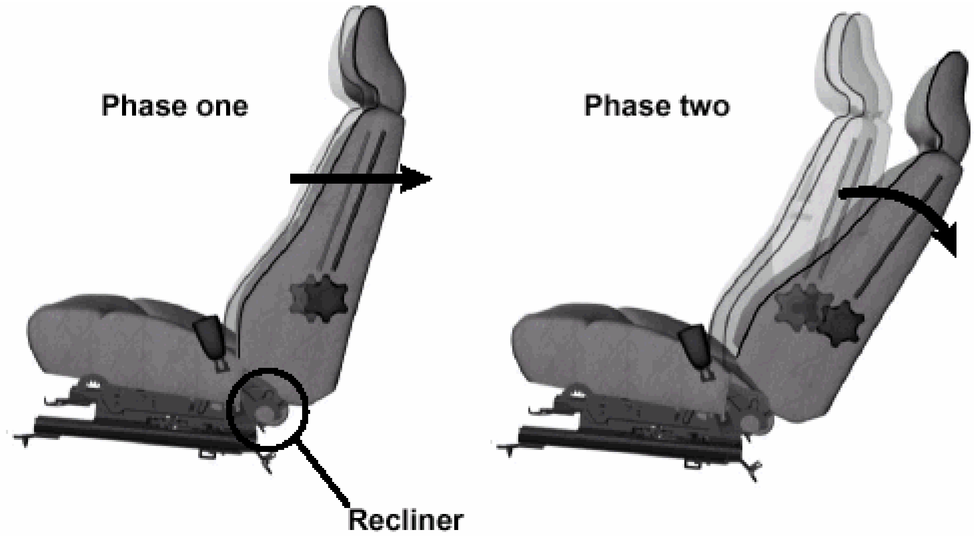


Figure 1.9: Volvo WHIPS seat motion.

The purpose of the first phase is: **1)** to let the occupant sink into the seat, thereby reducing the distance between the head and the head restraint, **2)** to create an initial rearward motion of the backrest which does not move the head restraint away from the head, and **3)** to keep occupant acceleration levels low, by letting the backrest move rearwards in a controlled way. This is accomplished by the first phase being a rearward motion of the seat backrest, the nature of this motion being essentially translational, i.e. without rotation. However, depending upon the pre-impact posture of the occupant, the motion characteristics of the backrest are to some extent adaptable and adjust to the occupant's position relative to the backrest. For example, if the occupant is leaning rearward before impact, this may give an initial tilt-forward motion of the backrest. The purpose of the second phase is to limit occupant acceleration to a low level. This is accomplished by a rearward reclining of the backrest, while absorbing energy in a controlled and gentle way.

When the backrest has absorbed the occupant's energy, and thus reclined to its rearmost position, rebound takes place. The rebound is, however, significantly reduced, compared to a conventional seat, because of the plastic energy absorption in the WHIPS recliner.

The reclining angle of the second phase is limited to approximately 15 degrees. When the maximum angle has been reached, the recliner assumes the stiffness characteristics of the existing production recliner, and the seat will perform as a seat without a WHIPS system.

The recliner is designed to operate primarily in the range of velocity change of approximately 10-20 Km/h.

Volvo's Traffic Accident Research Team shows that WHIPS reduces short-term injuries by 33 percent and long-term injuries by 54 percent. Moreover, Volvo is not alone in drawing this conclusion. Several independent surveys reveal major reductions in whiplash injuries thanks to WHIPS.

The Swedish Road Administration and Swedish Insurance Institute (Folksam) published findings of their survey and conclude that the number of whiplash incidents leading to serious injury would drop by 50 percent if all cars had the same system as that found in Volvo (Kraft, 2004).

1.5.4. Saab

In the event of a rear-end collision, the Saab Active Head Restraint System (SAHR) system is designed to limit the head movement of the occupant during the impact, helping to reduce the risk of whiplash injuries (Wiklund et al.,1998).

The system is entirely mechanical and is based on the lever principle. An upper padded support is connected to a pressure plate in the backrest of the seat. In some rear-end collisions, the occupant's body will be forced by the crash pulse into the backrest, which moves the pressure plate towards the rear. Subsequently, the head restraint is moved up and forward to "catch" the occupant's head before the whiplash movement can start. The precise activation of the system is determined by the force with which the occupant's back is forced against the backrest, the magnitude of the collision forces and by the occupant's weight.

Figure 1.10: Saab Active Head restraint System (SAHR)

A benefit of the mechanical SAHR system is that in most crashes it needs no repairs to restore it to operational condition after it has been activated. After the head restraint has constrained the movement of the head, it reverts to its initial position and is immediately ready to operate again. As whiplash injuries usually occur in low-speed collisions in which the vehicle may sustain only limited damage, the active head restraint does not increase the cost of the repairs needed after the crash.

New research from the Insurance Institute for Highway Safety (IIHS, 2001) has shown that Saab's Active Head Restraint (SAHR) system reduces neck injuries among car occupants by 43 percent. The study measured the effectiveness of the SAHR system by comparing the rates of insurance claims for driver neck injuries in rear-end crashes before and after the SAHR system was introduced.

1.6. Influence of Seat Back Cushion on Neck Injuries.

"….the seat and the head restraint are the most important facts concerning neck injury prevention": (Eichberger et al., 1996).

Hofinger et al., (1999) checked the influence of seat back cushion on neck injuries. Altogether 22 tests were performed which differ from each other by the combination of the foam, the seatback angle and the crash pulse.

They found out that by using a stiffer pelvis and a softer back cushion an earlier rotation of the torso around the pelvis was initiated. This movement with less relative displacement in the pre-contact phase decreased the gap between head and head restraint and resulted in a smaller head extension and the best acceleration and loading values in comparison to the other tests. In that test a soft cushion was used on the upper part of the seat back and a hard one on the lower part.

M. Hofinger et al., (1999) also found out that low acceleration can be achieved from the diving of the torso into the seat back combined with a rotation. Therefore the distance between head and restraint was very small when the extension of the head started and the acceleration values stayed low. Because of rotation of the torso the relative acceleration between head and T1 was very low and therefore the NIC was also low.

He also showed the importance of the cushion properties due to their behavior in reduction of severe neck injuries in rear-end impacts. The kind of cushion, its shape and position have a big influence to the seat behavior in rear-end impact.

Croft, (1998) confirmed the risk of cervical injuries at the moment of the first contact between head and head restraint. Such injuries can occur if the restraint is properly positioned. Immediately following head contact, the upper cervical spine will be forced into acute flexion as the inertia of the neck continues to draw it rearward, since there is no contact with either seat back or head restraint (Geigl, 1997).

1.7. Influence of Crash Pulses on Neck Injuries.

The crash pulse influence on neck injuries should be carefully analyzed in order to define appropriate rear-end impact tests. Kraft et al., (1998) showed that the shape of real-life crash pulses varies to a large extent and that the total speed change, $\Delta v_{tot\ pulse}$ is not a good measure to predict the duration of the occupant's symptoms.

Eriksson and Boström, (1999) also showed that total speed change and peak acceleration were not good NIC_{max} predictors. Rather, speed change during a limited time period, 70 to 110 ms, of the impact, equivalent to mean acceleration during the same period, showed to predict high NIC_{max} values well.

Zuby and Farmer, (2003) tested the effect of different acceleration pulse characteristics on BioRID responses. Bimodal acceleration pulses were compared with unimodal crash accelerations; early-peak acceleration pulses were compared with late-peak acceleration pulses. Neither the magnitude nor timing of peak BioRID responses was affected by the pulse shape difference, bimodal vs. unimodal. However, delayed peak sled acceleration (peak occurring at 72 ms compared with 18 ms) did affect the timing and magnitude of some BioRID responses. The head restraint contact and peak BioRID responses occurred later in the late-peak tests than in the early-peak and bimodal tests, although the differences were only significant for contact time, NIC, maximum Fx (longitudinal force), minimum Fx, maximum Fz (vertical force), and minimum Fz. The late peak acceleration pulse also tended to produce BioRID responses that were lower in magnitude compared with early-peak and bimodal pulses, however, only Nkm, maximum My (flexion/extension moment), and minimum Fz were significantly different from responses in the other tests.

1.8. Dummy Scaling

Mathematical models are very useful for generating dummies of different size due to the reduced development time and cost compared to physical dummies. The number of available sizes of mechanical dummies is still limited. Therefore, scaling procedures have become increasingly important in order to develop different size models from already available mathematical dummies.

1.8.1. Scaling Method

The scaling method has been described by Hapee et al., (1998). A general overview of the scaling procedure is presented.

The first step in this scaling method consists in generating a set of target anthropometry parameters from a relevant population. The corresponding parameters have also been evaluated for the reference models to be scaled. After generating the anthropometry parameters, the scaling is made by taking the ratio between the anthropometric parameters of the reference dummy and the values from the desired anthropometry, and from this comparison, different scaling factors are obtained for separate body parts for x, y and z directions.

The second part of the method consists of a correction process, in which the previously obtained factors are applied to the standard model and then the mass and main dimensions of the resulting model are checked. Since the mass is only an indirect result from the scaling, therefore it will deviate slightly from the specified mass. The correction is performed, optimizing the prediction of mass, erect standing height, seated height and shoulder width. In this correction phase only the geometry scaling factors are optimized.

For each body segment, the scaling factors l_x, l_y and l_z are defined. The geometry scaling factors are defined by a target (the desired dimension (l_{tx}, l_{ty}, l_{tz})) and a reference value (dimension from the reference dummy (l_{rx}, l_{ry}, l_{rz})).

$$l_x = \frac{l_{tx}}{l_{rx}}; l_y = \frac{l_{ty}}{l_{ry}}; l_z = \frac{l_{tz}}{l_{rz}}$$

The method scales mass assuming same density for the target and reference dummy:

$$\frac{m_t}{m_r} = l_x l_y l_z$$

For non-linear stiffness, the displacement and forces are scaled separately with different scaling factors:

$$\frac{d_t}{d_r} = l_z; \frac{F_t}{F_r} = l_x l_y$$

The resulting scaling factors are applied to the reference model. Complex and confidential non-linear scaling methods are applied. These methods enable scaling of all mechanical parameters, including joint stiffness and damping.

The method also performs scaling of:

- geometry,
- sensor Locations,
- reference length for the VC criterion,
- masses and moments of inertia,
- protected joint models,
- joint characteristics (stiffness, friction, damping, hysteresis),
- ellipsoids and contact characteristics,

- force models.

This method does not take into account varying tissue properties depending on size, since the scaling is based on geometry. This method is correct when scaling is done within the adult population, because equal tissue properties can be assumed.

1.8.2. Influence of dummy size on the estimated risk of whiplash injury

There is not much data available for tests done on 5th or 95th percentile dummies. However, in one study by DeRosia et al., (2004), tests performed on 5th percentile female and 95th percentile male postmortem human subjects at 15 and 25 km/h showed high NIC values of 19-37m^2/s^2. This study increased the concern of researchers when evaluating the risk or whiplash injury, stressing the fact that whiplash tests with body sizes different from the average male lead to increased injury parameter values.

To investigate the influence of body size on the injury parameter values, there have been some tests done with Hybrid III dummies concerning rear-end impact (DeRosia et al., 2002), because of the availability of the 5th percentile female, 50th percentile and 95th percentile male dummies in the Hybrid III family. Even though the Hybrid III dummy has been proven to be not biofidelic for low speed rear-end impacts, important information has been obtained from these studies:

- The 5th percentile female Hybrid III showed much higher values of injury related parameters (NIC, Nij and Nkm). This is because of the significantly different position of the head with respect to the seatback.
- This dummy also exhibited the highest peak values of all dummies in neck tension and shear forces, flexion moment in the neck region, head and T1 accelerations.

Even though these studies were made with Hybrid III dummies, they are very valuable since they attempt to compare between the three anthropometries. According to DeRosia et al., (2004), the values may not be close to what a more biofidelic model would give as a result, but since the testing procedure was constant, it is possible to successfully observe the influence of anthropometry.

From their study it seems to be clear that the lack of biofidelity of the Hybrid III affects the results, so more biofidelic models are needed.

Based on described studies one can say that there is a clear need to evaluate the risk of whiplash injury for different body sizes using more biofidelic models than the Hybrid III dummy, preferably mathematical models due to the lack of biofidelic mechanical dummies with different body size. This is also supported by clinical data which indicates the influence of body size on the risk of neck injuries during rear-end accidents.

1.9. Aim of the Study

Automotive companies perform tests and computer simulations in order to optimize and design safety devices in their vehicles, including seats for rear-end impact protection. Generally, these tests are performed using only one dummy size that is considered to be most representative. This leads to uncertainties in the performance of rear-end impact protection devices for different size occupants. In order to analyze how the optimization of automobile seats based on one size dummy affects the risk of whiplash injury for different size occupants, the present study was divided into the following investigations:

- The influence of crash dummy (BioRID-II) size and weight (95th percentile male, 50th percentile male and 5th percentile female) on whiplash injury related parameters in rear-end car impacts.

- The influence of crash dummy size and weight in optimization of seat construction i.e. analysis of the whiplash injury protection systems e.g., active head restraint system and seat back recliner system.

2. METHODOLOGY

Considering the available time and resources, the present study is based on computer Simulations. Computer models have great advantages over mechanical models, which have been already discussed (Paragraph 1.3.2). The software packages EASi-CRASH (MAD) and MADYMO are used for the modeling and simulation. In order to determine the risk of whiplash injury for different size occupants, the study is divided in four cases which are simulated and analyzed.

2.1. Modeling Requirements

To perform simulations for the analysis of rear-end impact scenarios, three elements were required to build the model of the system in MADYMO: dummy models, a seat system model and crash pulses.

2.1.1. Dummies

The model used in the current study is the MADYMO BioRID II MBS model (Figure 1.7). In addition to the superior biofidelity of the BioRID II dummy, it has the advantage of being a mathematical model, which is considerably easier to modify. This is essential for the current study, because different size dummies were needed and only BioRID II 50th percentile dummies (both mechanical and mathematical) exist. To develop mechanical BioRID II dummies of different sizes would have required a considerably longer time (years), than scaling the available 50th percentile mathematical model to generate the other dummy sizes that were needed. A description of the basic model was presented in paragraph 1.3.2.1.

2.1.1.1. Scaling Procedure

One of the tasks of the current study is to determine whether occupant body size is an important factor influencing injury severity during rear-end car impacts. To accomplish this, it is necessary to generate dummy models for smaller and larger body sizes.

Since there was only a BioRID II 50th percentile male dummy model available, this dummy was scaled to various body sizes by using the scaling methods already discussed in paragraph 1.8.1.

To perform the scaling, a parameterized BioRID II MADYMO MBS model dummy file has been written based on the original BioRID II model, which is based on 50th percentile male anthropometry. This file contains 35 anthropometry values (35 values define a specific anthropometry, see Appendix-IV), and a subset within these values is used to generate the scaling factors, by comparing the 50th percentile set of values with the set of values from the desired anthropometry, which are generated by MADYSCALE.

This parameterized dummy file was used as the reference dummy model for scaling and generating the dummies with the desired anthropometry. After the comparison and optimization process, the different dimensions and properties of the 50th percentile dummy are multiplied by the combinations of scaling factors generated by MADYSCALE, and a new dummy file with scaled properties is generated (For a flowchart of the scaling process, see Appendix-V).

To illustrate more clearly this methodology, the scaling procedure of the mass of a given body section (the head) will be described.

A set for the reference anthropometry and for the desired anthropometry is defined (Appendix-IV). After generation of the scaling factors by the use of the parameterized 50th percentile dummy and the desired anthropometry, a matrix of scaling factors is obtained (Table 2.1).

Table 2.1: Scale Factors for the 95th percentile male dummy.

	Lx	Ly	Lz	Lxyz
Pelvis	1.01761	1.03348	1.14884	1.06507
Lumbar Spine	0.99572	1.03002	1.14884	1.0562
Abdomen	1.03436	1.06696	1.14884	1.08233
Thoracic Spine	0.97496	1.08814	1.14884	1.06818
Ribcage	0.97496	1.05723	1.14884	1.05796
Neck	1.01103	1.01103	1.14884	1.05502
Head	0.96845	1.08731	1.02858	1.02697
Clavicles	0.97496	0.95863	1.14884	1.024
Upper Arm	1.04321	1.04321	1.18633	1.08889
Lower Arm	0.9992	0.9992	1.20821	1.0645
Hand	1.00379	0.88741	1.19851	1.02205
Upper Leg	0.99902	0.99902	1.21802	1.06726
Lower Leg	1.02275	1.02275	1.07877	1.04109
Feet	0.99412	1.03533	1.07877	1.0355

The value that corresponds to the head (row 7 in Table 2.1) is used to multiply the values of the 50th percentile dummy.

The center of gravity is scaled by simple multiplication of each dimension with its respective scaling factor. The mass and the inertia are scaled by combining the scaling factors with equations derived by TNO:

$$Mass = 4.4 \cdot X_{7,1} \cdot X_{7,2} \cdot X_{7,3}$$

$$Inertia = \begin{bmatrix} 0.0204 * (X_{7,1} \cdot X_{7,2} \cdot X_{7,3}) * \left(\dfrac{\dfrac{X_{7,2}^{2}}{2} + X_{7,3}^{2}}{1.5} \right) \\ 0.0211 * (X_{7,1} \cdot X_{7,2} \cdot X_{7,3}) * \left(\dfrac{1.8133\ X_{7,2}^{2} + X_{7,3}^{2}}{2.8133} \right) \\ 0.0143 * (X_{7,1} \cdot X_{7,2} \cdot X_{7,3}) * \left(\dfrac{1.1029\ X_{7,2}^{2} + X_{7,3}^{2}}{2.1029} \right) \end{bmatrix}$$

$$COG = (0.0203 \cdot X_{7,1}, 0 \cdot X_{7,2}, 0.0292 \cdot X_{7,3})$$

$X_{7,1} \cdot X_{7,2} \cdot X_{7,3}$ are the x, y and z scaling factors that correspond to the head, 7th row of the scaling factors matrix (Table 2.1). The 50th percentile value for the mass of the head is 4.4 kg, the center of gravity is located at (0.0203, 0.0, 0.0292) m, and the inertia is Ixx=0.0204 kg*m^2, Iyy=0.0211 kg*m^2, Izz=0.0143 kg*m^2.

In a similar way, all properties of the dummy have been multiplied by their correspondent scaling factors and/or combinations of them.

In addition to the 50^{th} percentile dummy, a 5^{th} percentile female dummy and a 95^{th} percentile male dummy are used in the study. These are considered as the two extreme anthropometry dummies within the adult population. The parameterized BioRID II dummy has been used to generate a 5^{th} percentile female dummy and a 95^{th} percentile male dummy (Figure 2.2).

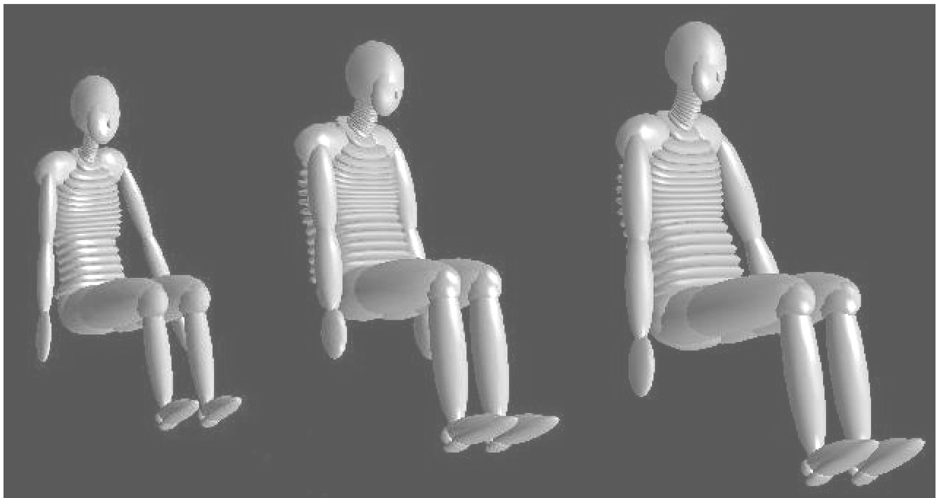


Figure 2.2: From left to right: generated 5^{th} percentile female, original 50^{th} percentile male and generated 95^{th} percentile male BioRID II MBS models.

It is very important to note that the intercept values for the Nkm injury criterion should be scaled for each corresponding dummy. The Nkm criterion includes in its equation some parameters which depend on the dummy size used, while the NIC injury criterion is independent from dummy size.

Kleinberger et al., (1998) used geometric scaling factors to scale Nij, assuming equal material properties between the dummies. The method used to scale Nkm is analogous, since the function variables to be scaled are similar: they are also force and moment values. Forces are scaled according to the cross-sectional area of the neck length, represented by the second power of the circumference. Bending moments are scaled according to the third power of the neck length, represented by the third power of the neck circumference (Table 2.2). Circumference measurements are used to quantify neck length because it is a simple measurement to record. The final scaled intercepts are finally obtained by multiplying the reference value (50^{th} percentile male value) by the calculated scale factors (Table 2.3).

Table 2.2: Scale factors for the three BioRID II dummies used in the present study. Neck circumference values were obtained from the parameters in MADYSCALE.

Dummy	Neck Circumference [m]	Neck Length Scale Factor $?_L$	Shear Load Scale Factor $?_L^2$	Bending Moment Scale Factor $?_L^3$
BioRID II 5^{th} %tile female	0.3214	0.869	0.755	0.655
BioRID II 50^{th} %tile male	0.370	1.000	1.000	1.000
BioRID II 95^{th} %tile male	0.4021	1.087	1.181	1.284

Table 2.3: Scaled intercepts for the Nkm criteria.

Dummy	Shear [N]	Moment Flexion/Extension [Nm]
BioRID II 5th %tile female	638	58/31
BioRID II 50th %tile male	845	88.1/47.5
BioRID II 95th %tile male	998	113/61

2.1.2. Seat systems

A Toyota Yaris seat was used in current study. This seat was used due to the following reasons:

- The MADYMO model of the seat was readily available for the study purpose.
- The seat is a typical example of a modern car seat.
- Seat has been validated for rear-end car impacts by Eriksson, (2000).
- The seat got a high rating in rear-end impact injury protection by the German Automobile Club (ADAC) (CarPages, 2004).
- On rear-end impact the seat backrest has a translational motion, which let the occupant sink into the seat.

The seat was modified for the analysis of the optimization of the active headrest system and the seatback recliner system. The seat was adjusted to give a torso angle of 25° as described in the *Static Evaluation of Head Restraints Procedure* by RCAR, (2001).

The head restraint was positioned in the “Good” zone (see Appendix-I) with respect to the height and head to headrest distance for the 5th percentile female, 50th percentile male and 95th percentile male dummies. This was done to eliminate the effect of headrest height on the test results. A separate case was performed to analyze this effect (this is explained in paragraph 2.1.4).

The exact modeling of the two whiplash injury protection systems was beyond the scope of this project. It would have required the correct seat properties and this information was lacking.

Only one seat (Toyota Yaris) model was used for the analysis of whiplash injury protection systems with active headrest and with seatback recliner. By doing this, the effect of having different seat properties (seat stiffness, seat damping properties, cushion stiffness, head restraint geometry, seat back geometry, mechanical and physical properties of the whiplash injury protection systems) on the whiplash injury related parameters is eliminated, and only the influence of the dummy size can be observed.

2.1.3. Crash Pulses

The shape and the severity of crash pulse are very important to assess risk for whiplash injury. This phenomenon is discussed in paragraph 1.7. In order to do proper analyses of whiplash injury risk for dummies of various size, it is necessary to perform simulations with crash pulse of different severities, because whiplash injury parameter values of each dummy size vary by varying severity of the crash pulses. Three crash pulses (high, medium and low) which are proposed by EEVC, (2004) are used, because EEVC recommend using these pulses with the BioRID II dummy model, and are under consideration to be a standard for future rear-end impact testing.

The three acceleration pulses have the same shape with maximum acceleration occurring at 27ms and delta-V duration of 91 ms, differing only in acceleration magnitude (Table 2.4). The shape of medium severity pulse is presented in Figure 2.3.

Table 2.4: Crash pulses used in the study.

Pulse Severity	Maximum Acc. [g]	Delta-V [km/h]
Low	5	10
Mid	10	16
High	15	26

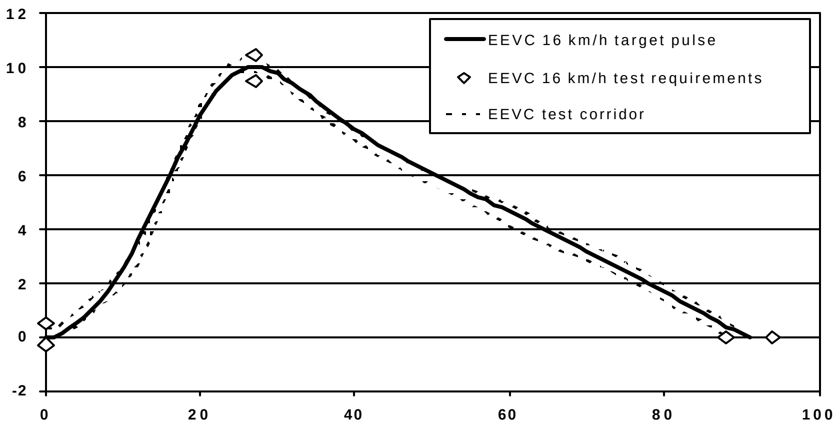


Figure 2.3: Medium Severity Crash Pulse

2.2. Simulation Procedure

To investigate the influence of seat optimization based on one dummy size on the risk of whiplash injury related parameters for different occupant sizes, a specific procedure was defined.

As the exact injury mechanism has not yet been established and different researchers have suggested several mechanisms. It can be observed that the injury mechanism is dependent on the:

- occupant acceleration;
- relative movements between adjacent vertebrae;
- the forward rebound.

Therefore, injury related parameters which we analyzed are: relative acceleration and velocity between the head and upper neck, linear acceleration of the head, chest, lower neck and pelvis in the longitudinal direction, head to headrest contact time, forces generated by the dummy on the seatback. NIC and Nkm are used as the main neck injury criterion. Forward rebound is not considered in the study because:

- no seat belt is used in the model;
- rebound is highly dependent on the seat mechanical and physical properties;
- BioRID II MADYMO model is not validated for rebound.

In order to fulfill the aims of the study, the simulations were performed of four cases:

2.2.1. Case-1: Analysis of the influence of the crash pulses on whiplash injury related parameters for various body sizes.

Tools:

Crash Dummies: BioRID II 95th percentile male, 50th percentile male and 5th percentile female.
Seat system: Toyota Yaris seat with the headrest in "Good" position (Appendix-I).
Crash pulse: Low, Medium and High severity.

In this case the original seat (without any modification regarding whiplash injury protection system) was used. This was done in order to examine the influence of the crash pulse severity on different body sizes with out any whiplash injury protection device.

2.2.2. Case-2: Analysis of the influence of head to headrest distance on whiplash injury related parameters for various body sizes.

Tools:

Crash Dummies: BioRID II 95th percentile male, 50th percentile male and 5th percentile female.
Seat system: Toyota Yaris seat with the headrest in "Good" position (Appendix-I) tilted forward 8.6 degrees.
Crash pulse: Low, Medium and High severity.

By the tilted position the distance between the head and the headrest has reduced which is the primary objective of the active headrest system. The whiplash injury related parameters of various body sizes are analyzed. The whiplash injury related parameters are also compared with Case-1 where there is no modification in the seat.

The limitation of such approach is that, as the exact active headrest system was not modeled, only the main effect of the active headrest system was considered (reduction of head to headrest distance), so only the change in the trend of whiplash injury related parameters could be observed instead of the whole time-history behavior of the active headrest system (i.e., the movement of headrest caused by the rearward movement of the dummy).

2.2.3. Case-3: Analysis of the influence of seatback recliner stiffness on whiplash injury related parameters**.**

Tools:

Crash Dummies: BioRID II 95th percentile male, 50th percentile male and 5th percentile female.
Seat system: Toyota Yaris seat with the headrest in "Good" position (Appendix-I), and with two different stiffness of the recliner joint. One stiffness is 25 % more than the original, and one is 25% reduced.
Crash pulse: High severity.

In this case the influence of the seatback recliner (i.e., variation of the stiffness of the recliner joint) was analyzed in comparison with the injury related parameters of various body sizes. Reducing the stiffness allows the seatback to recline more. It should be noted that motion of the recliner was divided into two phases:
1) Rearward translational motion of the seat backrest, which will let the occupant sink into the seat, thereby reducing the distance between the head and the headrest.
2) Rotational motion of the seat backrest which will control the occupant overall acceleration. The reclining angle of the second phase is not limited.

The limitation of this case is that, the model used is not able to absorb any energy by plastic deformation of any part. However, plastic deformation is used to minimize rebound, which is beyond the scope of the present study. With the model used, the effect of changing the recliner stiffness on the whiplash injury related parameters for the optimization of a seat recliner system such as the WHIPS can be observed, fulfilling the aims of the study. The whiplash injury related parameters are also compared with Case-1 where there is no modification in the seat.

2.2.4. Case- 4: Analysis of the effect of lowering the headrest height for 95th percentile male dummy on whiplash injury related parameters.

Tools:
Crash Dummies: BioRID II 95th percentile male.
Seat system: Toyota Yaris seat with the headrest height of 7cm.
Crash pulse: High severity.

The headrest for the 95th percentile male dummy was placed in the "Good" region (see Appendix-I) in the above cases. To accomplish this, the headrest had been raised 15cm which is unrealistic in present cars, where the maximum vertical displacement of the headrest is around 7-8 cm. In Case-4 the headrest is raised only 7cm (the maximum obtainable headrest height in the Toyota Yaris seat) and the influence of whiplash injury parameters was analyzed.

2.2.5. Seat backrest forces and recliner moment analysis

The influence of dummy size and recliner stiffness on active headrest system and the seat recliner system activation is an important part of the current study. Seatback forces and recliner moments activate the active headrest and the seat recliner systems respectively; therefore the influence of dummy size on forces on the seatback and on moments on the seat recliner is analyzed.

Forces on an area of the Toyota Yaris seat defined by the location of the SAHR system pressure plate have been calculated, in order to analyze the activation properties of a system with such pressure plate location (Figure 2.5). The contact area was defined by selecting the areas of the dummy's back that contacted the area defined on the seat at the start of the simulation. The advantage of this approach is the possibility of measuring forces on the dummy, and the possibility of measuring forces on the seat without modifying the model. The main disadvantage of this approach is that if the position of the dummy's back with respect to the seatback changes considerably (as in the possible case of ramping), the measured forces will not correspond to the forces on the pre-defined area, but to another section of the seatback. However, the vertical displacement of the dummy with respect to the seatback was observed to be small enough so the dummy's back still contacted a good part of the region during the simulations, so a clear trend can still be observed.

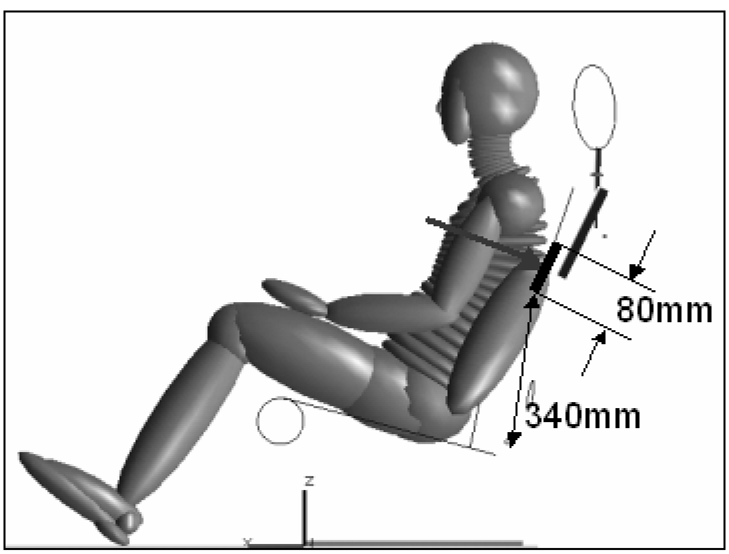


Figure 2.5: Active Headrest System pressure plate location.

3. RESULTS AND DISCUSSIONS

The current study is divided into four clearly distinct cases, in which each one of them meets a specific purpose in finding the factors that influence whiplash injury related parameters and seat optimization for low speed rear-end impacts. The results of the cases are presented and discussed in this section.

3.1. Case 1: Analysis of the influence of the crash pulses on whiplash injury related parameters for various body sizes

The highest injury values of NIC are observed in the simulations with the 5^{th} percentile female dummy (Figure 3.1). This is due to the lower mass of the small female dummy, which is accelerated to a higher extent than the male dummies. As is seen in the Appendix-II, all acceleration values in the simulation of the 5^{th} percentile female are higher, as well as the relative accelerations.

Except for the medium pulse, NIC values of the 5^{th} percentile female and the 95^{th} percentile male are higher than those from the simulation of the 50^{th} percentile dummy. A clear increase of NIC is seen when the low and high pulse was used ion the simulation of the 95^{th} percentile male when compared to the average male dummy. This is because the higher mass and larger size of the dummy increases the injury value in a less obvi ous way than in the 5^{th} percentile female: the higher mass of the dummy does not allow it to accelerate as a lighter dummy would be expected to accelerate, so contact with the seat and the headrest is delayed (Figure 3.2), increasing head to lower neck relative accelerations (Appendix-II). Later in current study it will be analyzed how that reduction of headrest contact time influences the injury values.

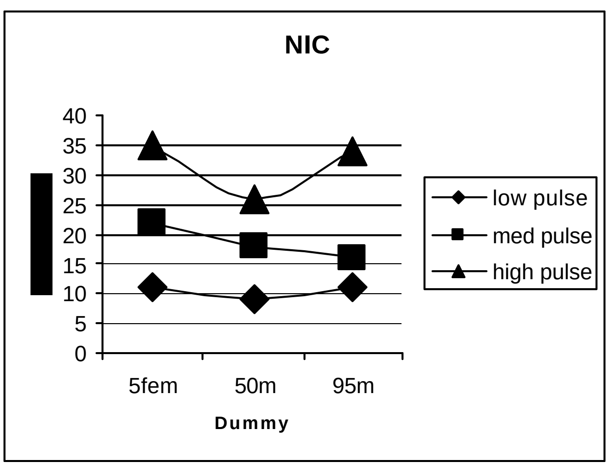


Figure 3.1: NIC and Dummy Size

Figure 3.2: Headrest contact time and dummy size

The larger size of the dummy also does not permit optimum interaction with the seat. In contrast with the 5^{th} percentile female, which sinks completely into the seat, the 95^{th} percentile male interacts only partially with the seatback (Figure 3.3). The body is too wide and too tall in relation to the seat; therefore it cannot properly fit into the seat. This in turn, is an obstacle for the head of the dummy to contact the headrest quickly. This influence is seen clearly for the low severity crash pulse, due to the size and mass of the large male dummy, it takes more time to completely interact with the seat, in comparison to the smaller dummies.

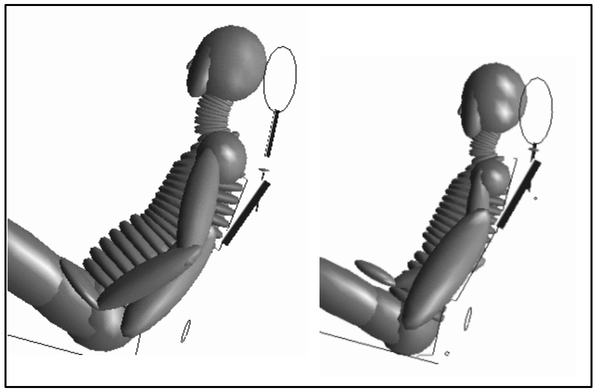


Figure 3.3: Difference of interaction with the seat: 95^{th} percentile male (left), 5^{th} percentile female (right).

An analogous trend is observed on the Nkm injury criterion. The highest injury values of Nkm are also seen in the situation where the 5^{th} percentile dummy is used and the lowest when the 50^{th} percentile male is used (Figure 3.4).

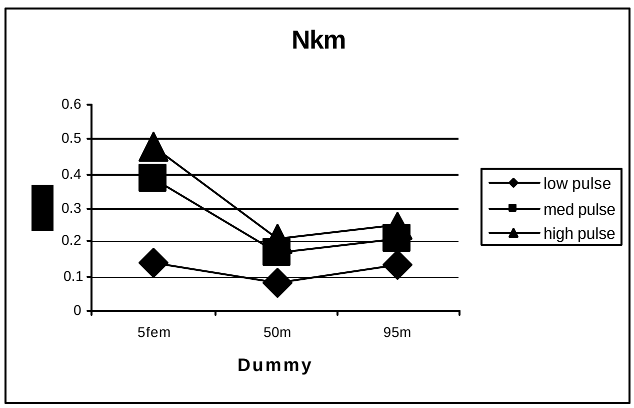


Figure 3.4: Nkm and dummy size.

The higher accelerations of the small female dummy due to its lower mass cause higher levels of moments and shear forces in the upper neck.

The 95^{th} percentile dummy's anthropometry generates higher values for the joint stiffness scaling factors, therefore making the 95^{th} percentile dummy's joints stiffer. Stiffer joints suffer higher moments and shear forces, increasing Nkm values.

3.2. Case-2: Analysis of the influence of head to headrest distance on whiplash injury related parameters for various body sizes

In this case the influence of head to headrest distance on whiplash injury related parameters for various body sizes was considered. All the results are presented in Appendix-II (Case-2). Only the influence of the high severity crash pulse is presented and discussed here. The influence of reducing the head to headrest distance on each body size is discussed separately, and the time history behaviour of NIC for each body size can also be found in Appendix-III.

5th Percentile Female.

The maximum NIC value in Case-1 for a 5th percentile female size is 35 m^2/s^2 at 64 ms, Nkm is 0.48 at 73 ms, and the head to headrest contact time is 55 ms. The values of other injury related parameters are also given in Table 3.1. Using these values from Case-1 as the baseline, the influence of reducing the head to headrest contact distance on the whiplash injury related parameters is analyzed. In Case-2, occupant lower neck and chest accelerations are reduced when compared to Case-1 (Table 3.1). This is directly related to the reduction of relative velocity and acceleration between head and the lower neck. The head to headrest contact time has also reduced because of the reduction of the head to headrest distance. The Nkm value has also improved.

Table 3.1: whiplash injury parameters for 5th percentile female for Case-1 & 2.

Case	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max)Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Lower neck (max) [m/s^2] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
1	35@ 64	0.48@ 73	55	152@ 64	2.37 @66	276 @ 64	246@7 5	236@ 65	146@ 57
2	21.6 @64	0.275 @83	48	101 @ 64	1.47 @ 66	263 @ 65	202 @ 74	233 @ 65	127 @ 57

50th Percentile Male.

The maximum NIC value in Case-1 for the 50th percentile male dummy is 26 m^2/s^2 at 79ms, Nkm is 0.207 at 97 ms, and the head to headrest contact time is 81 ms. Using these values from the results of Case-1 as the reference, the influence of reducing the head to headrest contact distance on the whiplash injury related parameters is observed.

The occupant (50th percentile male) maximum lower neck and chest accelerations are reduced (Table 3.2). The lower value of the NIC (16.5m^2/s^2 at 78 ms) is the result of the reduced relative movement (velocity) between the head and the lower neck. The head to headrest contact time has greatly reduced from 81 to 50 ms. Nkm values were reduced even thought they were under the tolerance limit of 0.3.

Table 3.2: whiplash injury parameters of 50th percentile male for Case-1 & 2.

Case	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max)Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Lower neck (max) [m/s^2] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
1	26@ 79	0.207 @97	81	107@ 78	2.62 @89	108 @ 78	229@ 104	170@ 80	103@ 74
2	16.5 @78	0.156 @103	50	71 @ 74	1.92 @ 85	98 @ 77	201 @ 101	170 @ 79	99 @ 74

95th Percentile Male.

The maximum NIC value in Case-1 for the 95th percentile male dummy is 34 m^2/s^2 at 91 ms, Nkm is 0.25 at 105 ms, and the head to headrest contact time is 94 ms. The values of other injury related parameters area also given in Table 3.3. It is interesting to see that Nkm gives a very low value. Using these values as the base line we will now observe the influence of reducing the head to headrest contact distance on the whiplash injury related parameters.

The dummy (95 percentile male) lower neck and chest accelerations are reduced (Table 3.3). The lower value of the NIC ($17.6 m^2/s^2$ at 95 ms) also shows the reduced relative velocity between the head and the lower neck. The head to headrest contact time has greatly reduced to 40 ms.

Table 3.3: whiplash injury parameters for 95^{th} percentile male for Case-1 & 2.

Case	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max)Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Lower neck (max) [m/s^2] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
1	34@ 91	0.25@ 105	94	148@ 90	2.86 @105	147 @ 91	176@ 125	167@ 76	83@ 81
2	17.6 @95	0.184 @86	40	70 @ 85	2.18 @ 98	126 @ 96	142 @ 123	116 @ 76	80 @ 80

Dummy Size Influence Analysis

Once having observed the behavior of NIC and Nkm when headrest distance was reduced for each size dummy, to fulfill the aim of the current study it is necessary to analyze how dummy size influences the effects of reduction of head to headrest distance on injury related parameters.

For all dummies, it is observed that NIC is dependent on both relative velocity and acceleration between head and the lower neck, and relative velocity is dependent on the head to headrest contact time. Relative velocity decreases with the reduction of head to headrest contact time, and in consequence, NIC also decreases (Appendix-II).

Considerable improvement in NIC was observed for all dummies by the reduction of head to headrest distance, but it is important to note that the 95^{th} percentile male dummy showed the largest improvement (Table 3.4). The 5^{th} percentile female has higher injury risk in all cases, and the seat has already been optimized for the 50^{th} percentile dummy, so there is less room for improvements for the 50^{th} percentile male situation.

For the Nkm criterion, the largest improvement is seen in the 5^{th} percentile female meaning that the active headrest system is very effective in reducing high neck forces (Table 3.4). The 50^{th} percentile male shows again the lowest improvement, since the seat has been optimized for the average male.

For both NIC and Nkm, the situation where the 50^{th} percentile male is used gives the lowest values (Table 3.4). This also confirms the observation that the Toyota Yaris seat has been designed for rear-end impacts with the 50^{th} percentile male in consideration.

It is also observed that NIC is also dependant on the time difference between the occurrence of head-T1 maximum relative acceleration and relative velocity. However, the time gap between relative acceleration and relative velocity is not the only factor affecting NIC.

Table 3.4: Injury criteria values in Case-2 and improvement in Case-2 relative to Case-1

Dummy	NIC	Nkm	NICmax improvement [%]	Nkm improvement [%]
5 fem	21.6	0.275	38	43
50 male	16.5	0.156	37	24
95 male	17.6	0.184	50	26

3.3. Case-3: Analysis of the influence of seatback recliner stiffness on whiplash injury related parameters

In this case the influence of changing the recliner stiffness on whiplash injury related parameters for various body sizes was observed. All the results are presented in Appendix-II (Case-3). Only the influence of the high severity crash pulse is presented and discussed here. The influence of reduced recliner stiffness (25 percent less than the original) on each body size is discussed and compared first with Case-1 because it is the case where the seat reclines more than the designed (optimized) seat and behaves more like a WHIPS system. The time history behaviour of NIC for each body size can also be observed in Appendix-III.

5th Percentile Female

Occupant lower neck and chest accelerations are reduced when reducing recliner stiffness, but there is an increase of the relative acceleration and velocity between head and the lower neck, which in turn has increased the NIC value from 35 to 41m^2/s^2. It is observed that the head to head rest contact time has also increased when lowering the recliner stiffness (Table 3.5).

Table 3.5: whiplash injury parameters for 5th percentile female for Case-1 & 3 (reduced stiffness).

Case	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max)Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Lower neck (max) [m/s^2] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
1	35@ 64	0.48@ 73	55	152@ 64	2.37 @66	276 @ 64	246@7 5	236@ 65	146@ 57
3	41@ 65	0.49@ 74	59	179@ 65	2.49 @68	250 @ 65	251@ 65	230@ 65	128@ 58

50th Percentile Male

The occupant maximum lower neck and chest accelerations are reduced (Table 3.6) when decreasing recliner stiffness. The relative acceleration between the head and the lower neck is reduced which resulted NIC value reduction (23m^2/s^2 at 91 ms), and the head to headrest contact time is increased. Although the head to headrest contact time is increased, the reduction in the occupant maximum lower neck and chest accelerations and the relative acceleration between the head and the lower neck helped in the reduction of the NIC. Nkm shows a very slight increase.

Table 3.6: whiplash injury parameters of 50 percentile male for case-1 & 3 (reduced stiffness).

Case	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max)Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Lower neck (max) [m/s^2] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
1	26@ 79	0.207 @97	81	107@ 78	2.62 @89	108 @ 78	229@ 104	170@ 80	103@ 74
3	23@ 91	0.216 @108	93	81@ 91	2.87 @99	90 @ 93	229@ 117	168@ 79	93@ 75

95th Percentile Male

The lower neck acceleration is increased and delayed but the chest acceleration is reduced slightly (Table 3.7). The higher value of the NIC (37m^2/s^2 at 105 ms) is the result of increased relative velocity between the head and the lower neck. The head to headrest contact time has also increased to 106 ms. Nkm increases when seat recliner stiffness is reduced.

Table 3.7: whiplash injury parameters for 95 percentile male for Case-1 & 3 (reduced stiffness).

Case	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max)Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Lower neck (max) [m/s^2] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
1	34@ 91	0.25@ 105	94	148@ 90	2.86 @105	147 @ 91	176@ 125	167@ 76	83@ 81
3	37@ 105	0.29@ 121	106	145@ 104	3.26 @114	154 @ 104	162@ 139	159@ 83	77@ 82

Influence of changing the recliner stiffness on whiplash injury related parameters of different body sizes.

It is an important part of the aim of the current study to examine the influence of changing the stiffness of the recliner joint on whiplash injury related parameters of different body sizes.

1). For the 5th percentile female, chest acceleration decreases as the stiffness decreases (Table 3.8). However, Head-T1 relative velocity and acceleration increase, which results in the increase of NIC. Head to headrest contact time increases with the reduction of recliner stiffness.

Reducing the recliner stiffness does not help in the reduction of injury related parameters for the 5th percentile female as it increases the head to headrest contact time (first contact). This in turn increases the relative movement (velocity) between the head and lower neck.

Table 3.8: Whiplash injury parameters for the 5th percentile female.

Rec. Stiff	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max) Head-T1 [m/s^2] @ [ms]	Vrel (max)Head-T1 [m/s] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
Low	41@65	0.49@74	59	179@ 65	2.49@ 68	251@ 65	230@ 65	128@ 58
Med	35@64	0.48@73	55	152@ 64	2.37@ 66	246@ 75	236@ 65	146@ 57
High	32@63	0.48@73	54	137@ 63	2.24@ 65	279@ 73	240@ 65	157@ 57

NIC and Nkm both have an increasing trend with the reduction of the stiffness of recliner joint (Figure 3.8 & 3.9). However, the change in Nkm is very small.

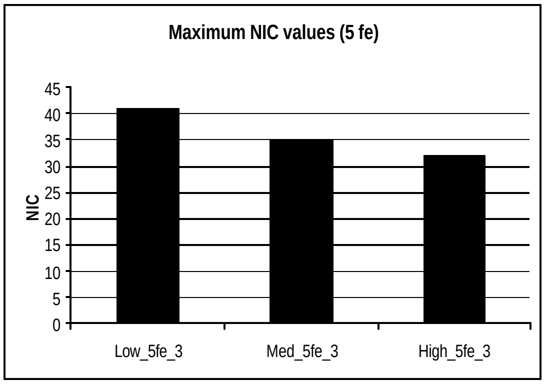


Figure 3.8: NIC and recliner stiffness (5%ile female)

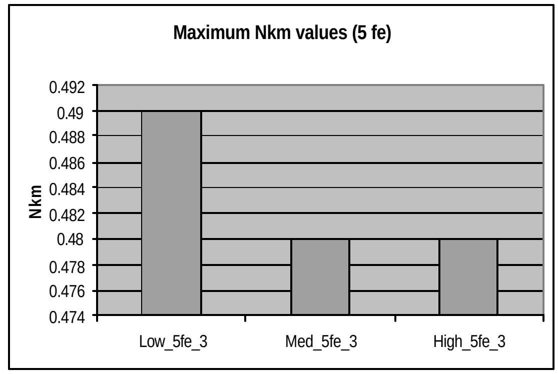


Figure 3.9: Nkm and recliner stiffness (5%ile female)

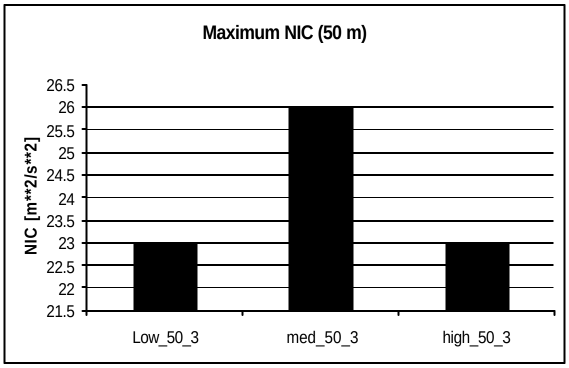
2). For the 50th percentile male, the lower neck and chest accelerations decrease with the decrease of recliner stiffness (Table 3.9 and Figures 3.10, 3.11). Relative velocity between head and lower neck increases. The lower value of the NIC is the result of lower relative acceleration between head and lower neck.

However, when recliner stiffness is increased, NIC is decreased as in the case of reducing recliner stiffness. Even though the Head-T1 relative acceleration is increased with the increase of recliner stiffness, a decrease of Head-T1 relative velocity is observed, which explains the reduction of NIC in this situation.

Reduction of the recliner stiffness decreases the lower neck and chest accelerations. Relative acceleration between head and the lower neck is also reduced. Nkm has an opposite trend to the NIC trend, since it gives the best value with medium recliner stiffness and increases when the recliner stiffness is changed. It is also interesting to observe that Nkm shows very little change in the injury related parameters when changing the recliner stiffness, and all the values of Nkm are under the tolerance limit of 0.3 (Table 3.9 and Figures 3.10, 3.11).

Table 3.9: Whiplash injury parameters for the 50th percentile male.

Rec. Stiff	**NIC [m^2/s^2] @ [ms]**	**Nkm @ [ms]**	**Head to HR time [ms]**	**Arel (max) Head-T1 [m/s^2] @ [ms]**	**Vrel (max)Head-T1 [m/s] @ [ms]**	**Ax head (max) [m/s^2] @ [ms]**	**Ax pelv (max) [m/s^2] @ [ms]**	**Ax chest (max) [m/s^2] @ [ms]**
Low	23@91	0.216@ 108	93	81@ 91	2.87@ 99	229@ 117	168@ 79	93@ 75
Med	26@79	0.207@ 97	81	107@ 78	2.62@ 89	229@ 104	170@ 80	103@ 74
High	23@76	0.212@ 92	72	96@ 71	2.47@ 83	243@ 99	173@ 78	111@ 70



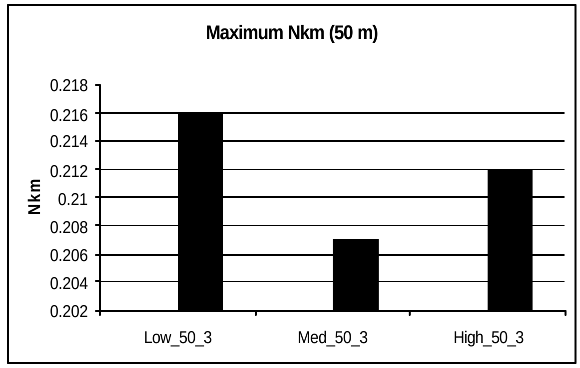
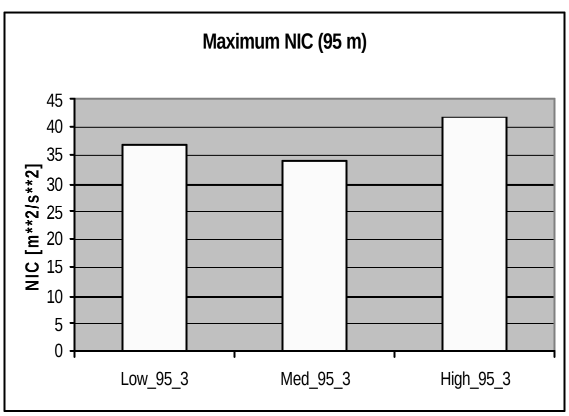


Figure 3.10: NIC and rec. stiffness (50 male) **Figure 3.11:** Nkm and rec. stiffness (50 male)

3). For the 95th percentile male the chest acceleration decreases as the recliner stiffness decreases but lower neck acceleration has a different trend. It decreases as the recliner stiffness decreases up to a certain limit and after that it starts increasing (Table 3.10). NIC increases as the recliner stiffness increases (Figure 3.12). Nkm has also the same trend as NIC but it is interesting to know that Nkm indicates very low injury risk for the 95th percentile male size as the value is under the tolerance limit of 0.3 for all recliner stiffnesses (Figure 3.13).

Table 3.10: Whiplash injury parameters for the 95th percentile male.

Rec. Stiff	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max) Head-T1 [m/s^2] @ [ms]	Vrel (max)Head-T1 [m/s] @ [ms]	Ax head (max) [m/s^2] @ [ms]	Ax pelv (max) [m/s^2] @ [ms]	Ax chest (max) [m/s^2] @ [ms]
Low	37@ 105	0.29@ 121	106	145@ 104	3.26@ 114	162@ 139	159@ 83	77@ 82
Med	34@91	0.25@ 105	94	148@ 90	2.86@ 105	176@ 125	167@ 76	83@ 81
high	42@85	0.28@97	84	191@ 85	2.69@ 91	175@ 119	172@ 79	91@ 79



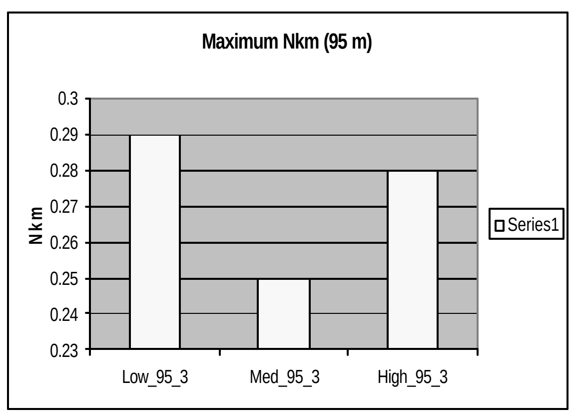


Figure 3.12: NIC and rec. stiffness (95 male) **Figure 3.13:** Nkm and rec. stiffness (95 male)

It is also observed that 95th percentile male dummy in the simulation with a high crash pulse, generates a torque of 2492 N at the recliner which displaced the seatback 29 degrees (see Appendix-III). The angle of inclination of the seatback is very high, and an excessively high inclination angle of the seatback is undesirable, because of the possibility of injuring occupants on the rear seats. The maximum angle of recliner should also be optimized for each dummy size, so it does not exceed the allowable limits.

The stiffness of the recliner must be optimized and limited in order to get the reduced acceleration. Otherwise a very low stiffness will result in a very high recliner angle of the seat back, and also the head to headrest contact time increases, which may result in increased injury parameter values. The optimized stiffness should be used in order to get the optimized maximum angle of the recliner. The maximum angle of recliner should also be limited so it does not harm the rear passengers.

It is observed that for a 5th percentile female high recliner stiffness gives the best results. The maximum recliner angle in this situation is 7.43 degrees (Appendix-lll, ID-21).

For the 50th percentile male, both high and low stiffness reduces NIC and the change in Nkm values is negligible. The maximum recliner angle for low stiffness is 24 degrees and for the high stiffness is 15 degrees (Appendix-lll, ID-22).

For 95th percentile medium stiffness gives the best results. The maximum recliner angle is 24 degrees (Appendix-lll, ID-26).

The chest acceleration decreases by the decrease of the stiffness but lower neck accelerations have a different trend. This is due to the fact that the 95th percentile dummy has is too big to fit into the seat. The seat has been optimized for a 50th percentile human, so a 95th percentile dummy cannot properly fit in to the seat. The shoulders and the lower neck area are out of seat. That's why they cannot penetrate in to the seat and properly transmit all the generated forces to the seat. The chest area transmits the forces to the seat effectively, therefore decreasing chest acceleration.

3.4. Case 4: Analysis of the effect of lowering the headrest height for 95th percentile male dummy on whiplash injury related parameters

Injury values appear to increase (Figure 3.14 and 3.15), but the increase is not very significant. Analyzing the possible causes, it was found that the height of the headrest was now located in the boundary between what is considered a "good" and a "poor" headrest height (Appendix-I). When the headrest is lowered more, NIC and Nkm values start to increase considerably.

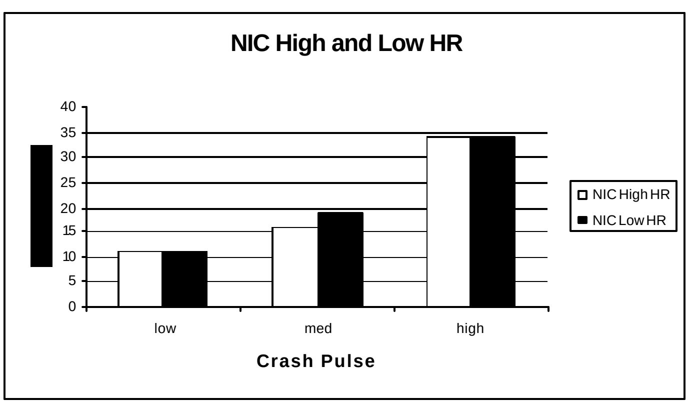


Figure 3.14: NIC high headrest and low headrest comparison for 3 crash pulses.

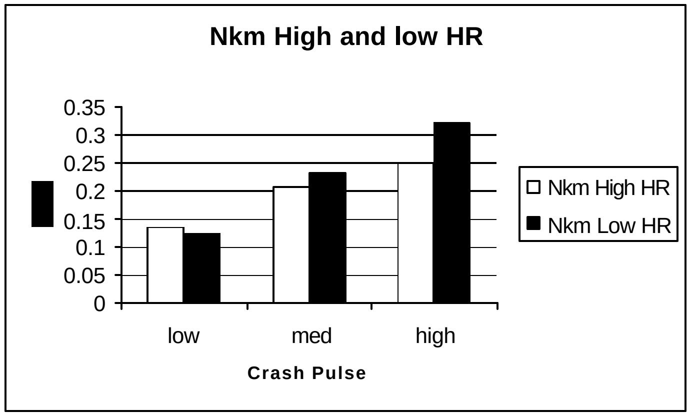


Figure 3.15: Nkm high headrest and low headrest comparison for 3 crash pulses.

A clear increasing trend in the relative head-T1 velocity is seen in all crash pulses (Figure 3.17). This means that the lowering of the headrest increases the risk of injury, even if there is not a large change in the NIC or Nkm values.

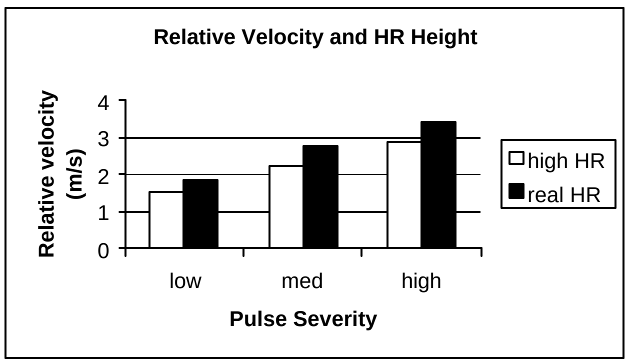


Figure 3.17: Relative Velocity Head-T1 and headrest height for 3 crash pulses.

This relative velocity increase is linked to the increased head to headrest distance (therefore the headrest contact time) as it has been previously mentioned in cases 2 and 3. NIC values do not change considerably because the time gap between maximum acceleration and maximum relative velocity is increased (Appendix-II, Case1- ID 7-9 and Case-4 ID 28-30).

4. SEATBACK FORCES AND RECLINER MOMENT ANALYSIS

The intent of this part of the study is to analyze the influence of dummy size on the activation of whiplash injury protection systems. The activation of these devices is directly related to the forces and moments on the seat, which are generated by the occupant's interaction with the seat.

4.1. Seatback Forces Analysis

It has been mentioned in Chapter-1 (Paragraph 1.5.4) that the activation of the Saab Active Headrest System is related directly to the force applied to the pressure plate (Figure 4.1). Once a certain level of force is reached, the headrest is brought forward. Forces on the seatback have been measured in order to observe its levels, but most importantly, to observe the time history behavior of these forces. The Toyota Yaris MADYMO model's seat back is divided into three sections, which facilitate quantifying forces independently for the lower, middle and upper back. This is important because the force behavior is influenced by its vertical location on the seat.

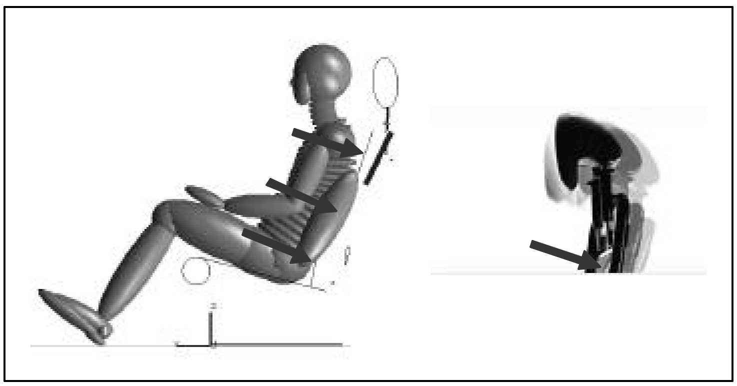


Figure 4.1: Seatback forces measured in the model (left) and force on pressure plate on the SAHR system (right).

Forces on the lower section of the seat are the largest for all the dummy sizes, but have some delay in time response (Figures 4.2 and 4.3). Middle back forces increase rapidly and reach a high level. The pressure plate on the SAHR is located on an area around the middle back section and the upper back section.

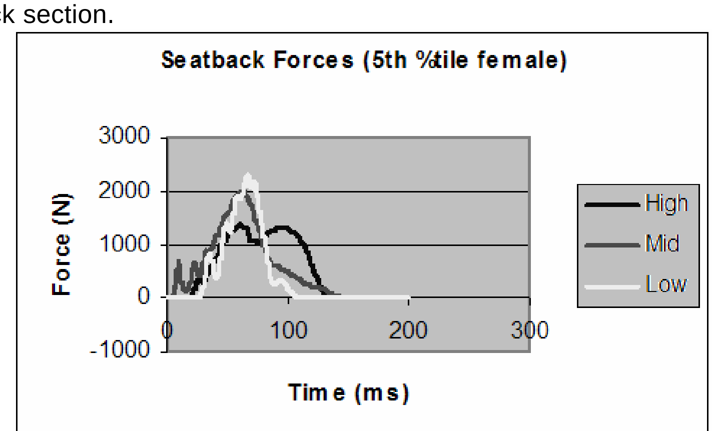


Figure 4.2: Seatback section forces for the 5th percentile female dummy

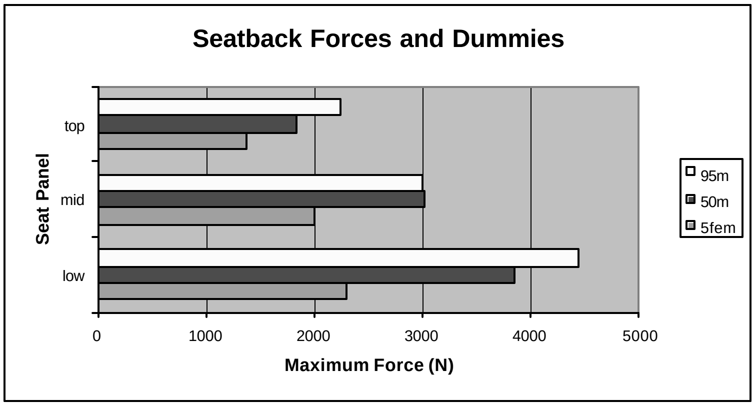


Figure 4.3: Seatback forces and Dummy size.

The middle panel forces for the three dummy sizes clearly indicate how an active headrest system might activate due to the different anthropometry (Figure 4.4). Results have been presented in a time history form, since the time history of the forces that might activate the whiplash protection systems should be analyzed in order to be able to compare activation performance in a more visible manner than just analyzing maximum values.

The 5th percentile female reaches the maximum force level faster than the other dummies because of its smaller mass (higher acceleration levels are reached as explained in Case 1), but the 95th percentile male is rapid to reach a certain force level, because of the higher mass. The deficient interaction with the seat of the 95th percentile male is also observed since the large male reaches forces similar to those of the 50th percentile male. The 95th percentile male is capable of applying higher forces on the seat because of the higher mass, but the forces on the seat are lower because the dummy's back cannot interact with the seatback entirely because of to large dimensions.

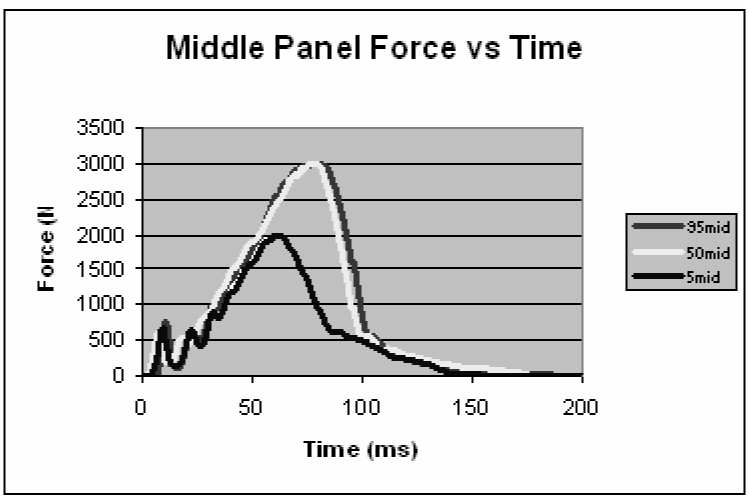


Figure 4.4: Middle panel force for the three dummy sizes.

When the seat recliner stiffness was changed (Case-3) no significant influence on time response for the maximum force value was observed (Table 4.1). Increasing recliner stiffness increases the level of force only to a small percentage compared to the increase of stiffness (25% stiffness increases the force only by 4%).

Table 4.1: Seat recliner influence on the middle panel force magnitude and time response.

Seat Recliner Stiffness	Middle Panel Force Maximum Value (N) @ Time [ms]
Low	2828 @ 78
Medium	3008 @ 78
High	3127 @ 77

The 5th percentile female reaches its own maximum in less time than the larger dummies, and the 95th percentile male reaches a pre-established force level faster than the smaller dummies (Figure 4.5). The 95th percentile male's performance is limited by the great size of the dummy compared to the car seat. Force measurements on the pre-defined area can be improved by modifying the mathematical model of the seat, by adding the pressure plate and its mechanical properties, but this would imply construction of a completely different seat model, which is out of the scope of this study.

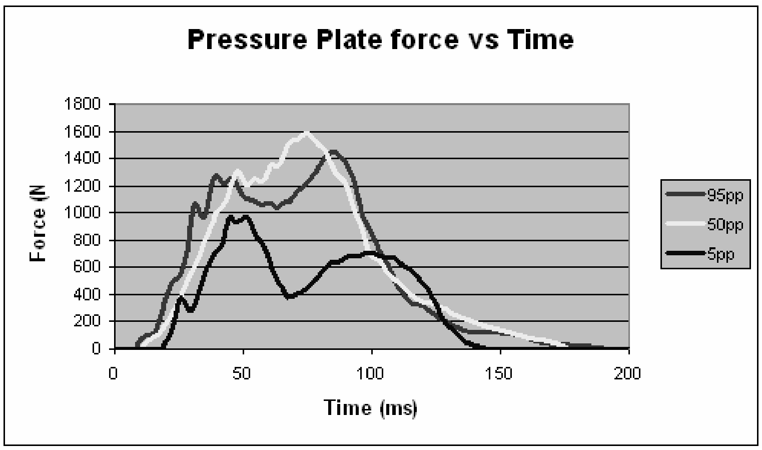


Figure 4.5: Pressure plate force and dummy size.

4.2. Seatback Recliner Moment Analysis

Activation of a seat recliner system is related to the moment on the recliner joint of the seat (Figure 4.6). Therefore, the influence of dummy size and seat recliner stiffness on recliner moment has been analyzed.

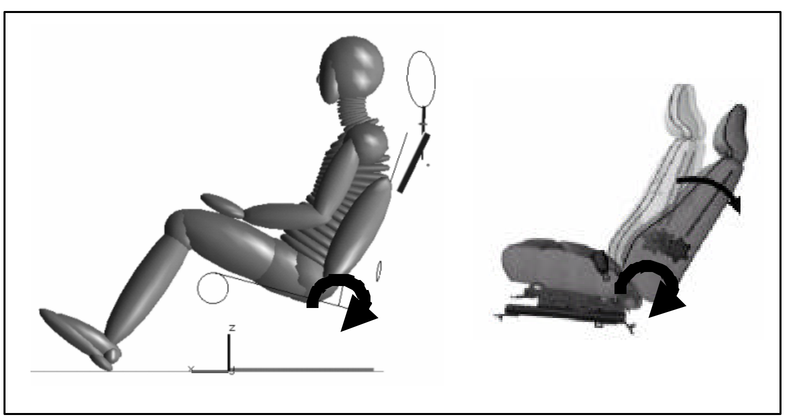


Figure 4.6: Setup for calculation of recliner moment.

The small female dummy reaches its own maximum value in less time than the larger dummies due to its smaller mass as explained in Case-1 (Figure 4.7). The 95th percentile male reaches a certain moment value in less time than the smaller dummies. When increasing seat stiffness the moment values became larger, but the increase is not as significant as the increase of stiffness.

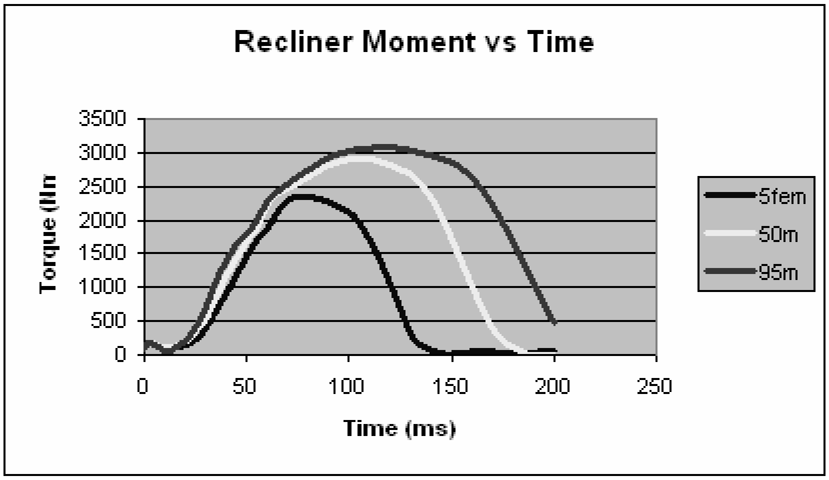


Figure 4.7: Recliner moment and dummy size.

From the time history, it is seen that the Toyota Yaris seat model allows the dummy to sink to the seat for about 20 ms with the high severity pulse, before it actually starts reclining.

5. CONCLUSIONS

Case-1: Analysis of the influence of the crash pulses on whiplash injury related parameters for various body sizes.

- The simulation of small female dummy shows the highest whiplash injury related parameters values. This is due to the very clear influence of the smaller mass. In the simulation of 95th percentile male dummy the injury related values are higher than in one of the average male dummy, because of the not so obvious but still significant reasons: less than optimum seat interaction and stiffer joints.

- The lowest values of injury related parameters are in the simulation of the 50th percentile male dummy. This means that the seat used in current study has been designed and optimized for the average male dummy.

Case-2: Analysis of the influence of head to headrest distance on whiplash injury related parameters for various body sizes.

- It is observed from simulations that changing the position of the head rest gives best results for all three dummy sizes. We could observe a reduction of:

  - head to head rest contact time;
  - relative velocity between head and the lower neck;
  - overall occupant acceleration.

- Reduction of relative velocity between the head and lower neck is more influential on the reduction of NIC, than the deceleration of the occupant.

- NIC and Nkm have the same trend for all three dummy sizes when three crash pulses were investigated. Their values increase by the increase of the pulse severity. Both indicated the higher whiplash injury risk when the female dummy was used in the simulation.

- Nkm indicated great reduction of injury risk parameters for all dummies due to the change of the headrest position (Figure 5.1), while NIC indicated higher risk for all dummy sizes only when the high severity pulse was used (Figure 5.2). Both injury criteria seem to be sensitive to the change of head rest position.

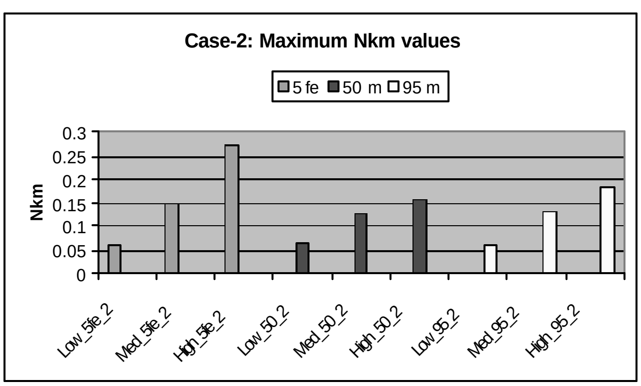


Figure 5.1: Maximum Nkm values of case-2

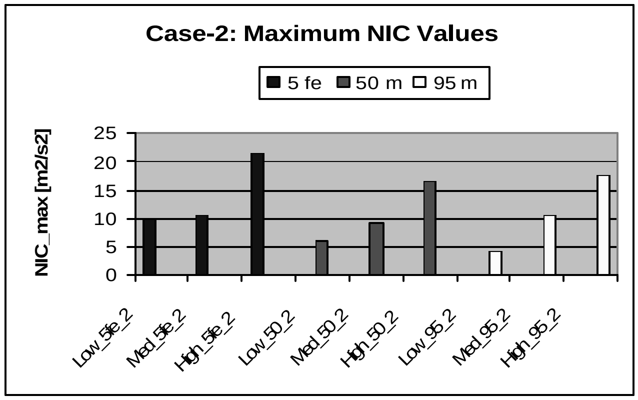


Figure 5.2: Maximum NIC values of case-2

- Systems with active headrest movements are designed to control the relative velocity between head and neck. These systems have good control over the relative movement between head and neck and helps in the reduction of whiplash injury risk for all three dummy sizes. It is also easy to optimize the system because of its simple operation principle and even if the system is optimized for one occupant size, it will most probably give satisfactory results if used by other occupant sizes.

Case-3: Analysis of the influence of seat back recliner stiffness on whiplash injury related parameters for various body sizes.

- Reduction of the seat stiffness does not cause the reduction of NIC when a 5th percentile female dummy was used (Figure 5.3). Reduction of the seat stiffness resulted in the increase of relative velocity and acceleration between the head and lower neck, although the occupant lower neck and chest accelerations are reduced (Table 4.6).

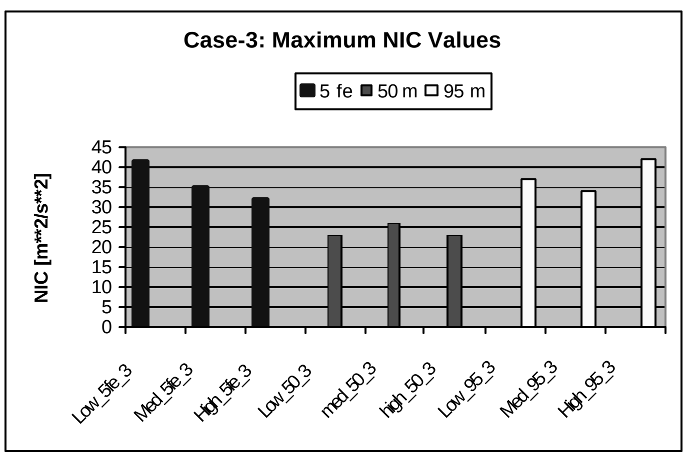


Figure 5.3: Maximum NIC values in Case-3

- For 95th percentile male size reducing the recliner stiffness resulted in the increase of lower neck acceleration although the chest acceleration is reduced (Table 3.7).

- It is also observed that variation (increase and decrease) of the recliner stiffness indicated higher NIC for all dummy sizes (Figure 5.3). NIC indicated the best results for a

50th percentile male, due to the optimization of the seat for 50th percentile male size by the manufacturer.

- Nkm indicated only the higher whiplash injury risk for the small female (Figure 5.4). No risk for a 50th and 95th percentile male sizes are observed. It is also observed that Nkm is sensitive to the dummy size but not much to the variation of the recliner stiffness. It can be also observed from Figure 5.4 that the change in magnitude of Nkm is very small when the recliner stiffness was modified for each dummy size.

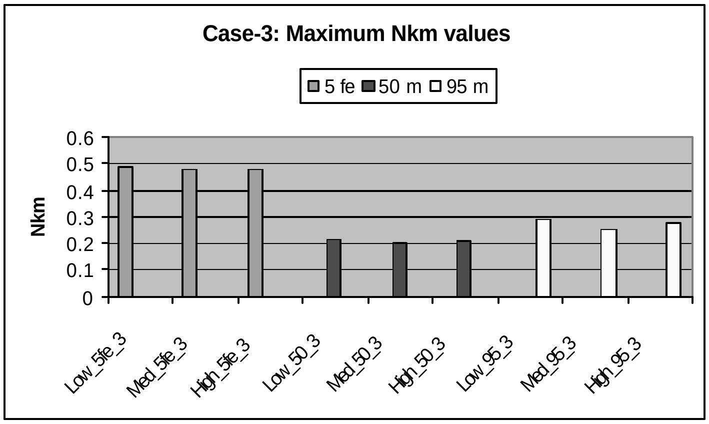


Figure 5.4: Maximum Nkm values in Case-3

- The system with seat back recliner is designed to control the occupant overall acceleration, as well as the relative velocity between the head and lower neck. This kind of systems helps in the reduction of acceleration for 5th and 50th percentile dummies but not for 95th percentile (Table 3.7). 95th percentile male has a big size and can't properly fit in to the seat. On the other hand, increase in the head to head rest contact time increases the relative velocity and also the NIC. The Toyota Yaris seat allows the occupant to sink into the seat for around 20 ms with high severity pulse (Phase-I, see Chapter-2 under Paragraph 2.2.3). However, this it is not enough to obtain a significant reduction of head to headrest contact time. The results might be better with medium and low severity pulses, because of the increase in the time for the dummy to sink into the seat.

- In order to get the best results, the whiplash injury protection system with seat recliner needs to be optimized (the optimization of recliner stiffness and maximum allowable angle of recliner, for each dummy size as discussed in Chapter 3 under section 3.3 for each dummy size and pulse. Whiplash injury protection systems with seat recliner, if optimized for one occupant size may not be suitable to be used for other occupant sizes because as it has been shown in the present study, the behavior and performance of the seat recliner system varies considerably with each dummy size when the seat has been optimized for the 50th percentile male.

Case-4: Analysis of the effect of lowering the headrest height for 95th percentile male dummy on whiplash injury related parameters.

- When the headrest is lowered from the ideal position, there is still a fairly broad area in which the headrest still functions correctly. The headrest rating diagram (Appendix-I) clearly illustrates this, and it has been proven in current investigation. There is a limit in which the headrest ceases to work correctly and injury values start increasing if the headrest is lowered above that limit. Also, the importance of observing different injury related variables is important when some results from injury criteria are not conclusive.

Seatback Forces and Recliner Moment Analysis

- The largest forces were found on the section of the seat corresponding to the lumbar area of the dummy, but the fastest force time response is observed for the middle panel (the thoracic section of the seatback) force for all dummy sizes.
- The 95th percentile male dummy's generated forces and moments reach a certain level in less time than in the cases where the smaller dummies were used; on the other hand, the 5th percentile female dummy's generated forces and moments are the slowest to reach a pre-defined force or moment level. However, the dummy's interaction time with the seat is shorter, reaching its own force/moment peak value faster than the larger dummies.
- The change of recliner stiffness does not affect the whiplash injury protection systems' activation significantly.
- The force and moment time-history behavior (which is directly related to rear-end impact injury protection) should be considered independently for each dummy size for the optimization of whiplash injury protection systems.

6. GENERAL CONCLUSIONS

The influence of dummy size on the optimization of seats for rear-end impact situations was studied. A Toyota Yaris seat was used as a representative of a typical, modern car seat. To simulate the car occupants in rear-end car collusion the most advanced currently available mathematical model of the BioRID ll dummy was selected. We conclude:

- The small female is at highest whiplash injury risk.
- The large male is at higher whiplash injury risk than the 50th percentile dummy.
- Active Headrest Systems decrease whiplash injury risk for varying anthropometry even with the use of the 50th percentile-optimized seat.
- Seat recliner systems are more complex to optimize since each dummy size and crash pulse have to be considered independently.
- Seat recliners optimized for one dummy size and pulse might not give acceptable results with other size of dummies and crash pulses.
- The seat used in current study is over-optimized for the 50th percentile dummy.
- NIC and Nkm are sensitive to active head rest systems (i.e., the whiplash injury protection systems that reduce the occupant relative velocity between lower neck and head). Both indicated the change in the whiplash injury risk parameter values by reduction of head to headrest contact time.
- NIC is sensitive to seat recliner systems (i.e., the whiplash injury protection systems that try to control the occupant overall acceleration as well as the relative velocity between lower neck and head), however Nkm seems to be less sensitive to the variation of recliner stiffness. NIC indicated higher risk for all occupant sizes when recliner stiffness was changed and Nkm indicated high risk for the small female only.
- Different body sizes have to be considered in the optimization of the seats as well as the whiplash injury protection systems. Presently one body size cannot be used as a representative for all body sizes. Regarding the whiplash injury the seats optimized for one body size cannot give satisfactory protection of occupants having other body sizes.
- The cooperation of the car manufactures or suppliers is needed to develop mathematical models of active headrest system and seat backrest recliner system. Also in future studies, in crash simulations the seatbelt and forward rebound should be considered.

Recommendations

The current study was an attempt to investigate the influence of seat optimization based on one dummy size on the risk of whiplash injury related parameters for different size occupants. Based on the analysis performed and considering the limitations of the study we can also conclude that:

- More work is needed to develop mathematical models of the active headrest system and seat backrest recliner system. Also in the modeling of the crash the seatbelt and forward rebound should be considered. To consider rebound, it is necessary to validate the dummy models for this situation.

- Optimization of the seats as well as the whiplash injury protection systems should be done with dummies representing occupants of various body sizes.

- Other injury criteria can be considered, but to accomplish this, injury criteria should be validated and verified. In the validation of injury criteria, different body sizes should be used.

- The cooperation with the car industry is really important in order to develop the mathematical models of the whiplash injury protection systems.

List of References

Aldman B, (1986). An Analytical Approach to the Impact Biomechanics of the Head and Neck Injuries. Proc. Of 30th Annual AAAM Conference, pp 439-454., LC 64-1965.

Allan F Tencer, Sohail Mirza, Philippe Huber (2001). The Response Of The MADYMO 50th Percentile Male Human Model (V 5.4), Human Volunteers, And Cadaveric Specimens: A Comparison In Rear-end End Impact. 9th International MADYMO User Conference pp. 1-9.

Boström O, Svensson MY, Aldman B, Hansson HA, Håland Y, Lövsund P, Seeman T, Suneson A, Säljö A, Örtengren T. (1996) A New Neck Injury Criterion Candidate – Based On Injury Findings In The Cervical Spinal Ganglia After Experimental Neck Extension Truma, Autoliv Research, Chalmers University Of Technology, University Of Göteborg, National Defense Research Establishment, Sweden. IRCOBI conference 1996, pp 123-136.

CarPages (2004), www.carpages.co.uk.

Chapline JF, Ferguson SA, Lillis RP, Lund AK, Williams AF, (2000). Neck Pain And Head Restraint Position Relative To The Driver's Head In Rear-end End Collisions. Accident Analysis And Prevention 32. pp 287-297.

Davidsson J., Lövsund P., Ono K., Svensson M., Inami S. A. (1999). Comparison Between Volunteer, Biorid P3 And Hybrid III Performance In Rear-end Impacts. IRCOBI Conference – Sitges (Spain), September 1999, pp. 165–178.

Derosia J, Yoganandan N, Pintar FA. (2004). Rear-end Impact Responses Of Different Sized Adult Hybrid III Dummies. Traffic Inj Prev. 2004 Mar;5(1):50-5. PMID: 14754675 [Pubmed - Indexed For MEDLINE].

Derosia J, Yoganandan N, Pintar FA (2002). Small Female And Large Male Responses In Rear-end Impact, Department Of Neurosurgery, Medical College Of Wisconsion. VA Medical Centre, Milwaukee, WI. 46th Annual Proceedings, Association for The Advancement Of Automotive Medicine, 2002.

EEVC (2004). EEVC Working Document 54, Low Speed Rear-end Impact Test Procedure, April 12 2004.

Eriksson L. (2000). Mathematical Modeling Of Low Speed Rear-end Impacts – Development Ad Validation Of MBS-Models, And Influence Of Risk Factors On NIC, Crash Safety Division, Department Of Machine And Vehicle Systems, Chalmers University Of Technology, Göteborg, Sweden.

Eriksson L, Boström O. (1999). Assessing The Influence Of Crash Pulse, Seat Force Characteristics, And Head Restraint Position On NICmax In Rear-end End Crashes Using A Mathematical BioRid Dummy, Autoliv Sweden, Autoliv Research, Chalmers University Of Technology, Sweden, IRCOBI Conference –Spain 1999, pp 213-230.

Eriksson L. (2000). Three-Dimensional Mathematical Models Of The Biorid I And Car Seats, For Low-Speed Rear-end Impacts. Chalmers University Of Technology, Department Of Machine And Vehicle Design, Autoliv Sweden Pp. 1-17.

Erichberger A, Steffan H, Geigl B, Svensson M, Boström O, Leinzinger PE,Darok M. (1998) Evaluation Of The Applicability Of The Neck Injury Criterion (NIC) In Rear-end End Impacts On The Basis Of Human Subject Tests. Proc. Of IRCOBI Conference, 1998, pp 321-333.

FMVSS 202 (2001). Department Of Transportation, National Highway Traffic Safety Administration, 49 CFR Part 571, Federal Motor Vehicle Safety Standards; Head Restraints; Proposed Rule Pp 968-999.

Happee R., Haaster R. van, Michaelsen L., Hoffmann R. (1998), "Optimisation of vehicle passive safety for occupants with varying anthropometry", ESV Conference 1998, Paper 98-S9-O-03.

Hofinger M, Mayrhofer E, Geigl B, Moser A, Steffan H. (1999) Reduction Of Neck Injuries B Improving The Occupant Interaction With The Seat Back Cushion, Institute Of Mechanics, University Of Technology Graz, DSD Linz, Austria, IRCOBI Conference- Spain 1999, pp 201-212.

IIHS (1999). IIHS (Insurance Institute for Highway Safety), STATUS REPORT Vol. 34, No. 5, May 22, 1999.

IIHS (2001). IIHS (Insurance Institute For Highway Safety),STATUS REPORT Vol. 36, No. 9 , Oct 6, 2001.

Jakobsson L. (2004). Whiplash Associated Disorders In Frontal And Rear-end Car Impacts – Biomechanical Guidelines And Evaluation Criteria Based On Accident Data And Parameter Studies , Crash Safety Division, Department Of Machine And Vehicle Systems, Chalmers University Of Technology, Göteborg, Sweden.

Jakobsson L. (2004). AIS 1 Neck Injuries In Rear-end End Car Impacts– Biomechanical Guidelines And Evaluation Criteria Based On Accident Data And Occupant Modeling , Crash Safety Division, Department Of Machine And Vehicle Systems, Chalmers University Of Technology, Göteborg, Sweden.

Krafft M, Kullgren A, Tingvall C, (1998). Crash Pulse Recorders In Rear-end Impacts- Real Life Data, Folksam Research And Karolinska Institute, Swedish National Road Administration, Sweden, Paper No. 98-S6-0-10.

Krafft M, Kullgren A, Ydenius A, Tingvall C. (2002). Influence Of Crash Pulse Characteristics On Whiplash Associated Disorders In Rear-end Impacts - Crash Recording In Real Life Crashes. Traffic Injury Prevention 3(2), pp 141-149.

Krafft M, Kullgren A, Lie A, Nygren Å, Tingvall C. (1997). Soft Tissue Injury Of The Cervical Spine In Rear-end End Car Collisions. Journal Of Traffic Medicine 25(3-4).

Kullgren A, Eriksson L, Boström O, Krafft M. (2003). Validation Of Neck Injury Criteria Using Reconstructed Real-Life Rear-end Crashes With Recorded Crash Pulses, Folksam Research, Autoliv Research, Autoliv Sweden , Chalmers University Of Technology. Paper No. 344. Proc. of the 18th Int. ESV Conference, 2003.

Krafft M, Kullgren A, Lie A, Tingvall C. (2004). Assessment Of Whiplash Protection In Rear-end Impacts - Crash Tests And Real Life Crashes, Folksam Research, Swedish National Road Administration, June 2004.

Linder A. (2001). Neck Injuries In Rear-end Impacts – Dummy Neck Development, Dummy Evaluation And Test Condition Specifications, Crash Safety Division, Department Of Machine And Vehicle Systems, Chalmers University Of Technology, Göteborg, Sweden.

Lundell B, Jakobsson L, Alfredsson B, Lindström M, Simonsson L. (1998). The WHIPS Seat - A Car Seat For Improved Protection Against Neck Injuries In Rear-end End Impacts, Volvo Car Corporation And Autoliv, Sweden. 16th ESV Conference, Canada. Paper Number 98-S7-O-08.

Lundell B, Jakobsson L, Alfredsson B, Jernström C, And Hellman I. (1998). Guidelines For And The Design Of A Car Seat Concept For Improved Protection Against Neck Injuries In Rear-end End Car Impacts, Volvo Car Corporation, Volvo Data Corporation, International Congress And Exposition Detroit, Michigan, 1998.

Lövsund P, Nygren Å, Salen B, Tingvall C. (1988). Neck Injuries In Rear-end End Collisions Among Front And Rear-end Seat Occupants. Proc. Of Int. IRCOBI Conference, 1988, pp 319-325.

Morris AP, Thomas P. (1996). A Study Of Soft Tissue Neck Injuries In The UK. Paper No. 96-S9-O-08. Proc. Of The 15th Int. ESV Conference, 1996, pp 1412-1425.

Mertz J.H. And Patrick, L.M. (1971) Strength And Response Of The Human Neck. Proceedings of The 15th Stapp Car Crash Conference, 1971.

Mertz, H, Prasad, P, Irwin, A. (1997). Injury Risk Curves For Children And Adults In Frontal And Rear-end Collisions, Proceedings Of The 41st Stapp Car Crash Conference, SAE Paper 973318, 1997.

M. Muser (Zurich), H. Zellmer (Elmshorn), F. Walz (Zurich),In Collaboration With W. Hell And K. Langwieder (Munich) (2001). Test Procedure For The Evaluation Of The Injury Risk To The Cervical Spine In A Low Speed Rear-end End Impact, Working Draft #5 pp. 1-12.

National Highway Traffic Safety Administration, US Department of Transportation, USA. http://www.nhtsa.dot.gov/

Nissan Motors, (2004), www.nissan-global.com

Nygren Å, (1984). Injuries To Car Occupants – Some Aspects Of The Interior Safety Of Cars – A Study Of Five Years material From An Insurance Company. Acta Oto- Laryngological – Suppli.395, ISSN 0365-5237, 1984.

Ono K, Kaneoka K, Wittek A, Kajzer J. (1997). Cervical Injury Mechanism Based On The Analysis Of Human Cervical Vertebral Motion And Head-Neck-Torso Kinematics During Low Speed Rear-end Impacts, Japan Automobile Research Institute, University Of Tsukuba Nagoya University, 1997.

Otte D, Pohlemann T, Blauth M. (1997). Significance Of Soft Tissue Neck Injuries AIS 1 In The Accident Scene And Deformation Characteristics Of Cars With Delta-V Up To 10 Km/H. Proc. Of Int. IRCOBI Conference, 1997, pp. 265-283.

Olsson I, Bunketor O, Carlsson G, Gustafsson C, Planth I, Norin H, Ysander L. (1990). An In Depth Study Of Neck Injuries In Rear-end End Collisions. Proc. Of Int. IRCOBI Conference, 1990, pp 269-280.

Otoophtalmical Neurophysiology, Buenos Aires, Argentina. http://vertigo-dizziness.com/english/whiplash_injury.html

Panjabi MM, Wang JL, Delson N. (1999). Neck Injury Croiteria Based On Intervertebral Motions And Its Evaluation Using An Instrumented Neck Dummy. Proc. Of Int. IRCOBI Conference, 1999, pp. 179-190.

Prasad P, Kim A, Weerappuli DPV, Roberts V, Schneider D. (1997). Relationships between Passenger Car Seat Back Strength And Occupant Injury Severity In Rear-end End Collisions: Field And Laboratory Studies. SAE Paper No. 973343, Proc. of 41st Int. Stapp Car Crash Conference, 1997.

Parkin S, Mackay GM, Hassan A.M, Graham R. (1995). Rear-end End Collisions And Seat Performance – To Yield Or Not To Yield, Proc. 39th Annual AAAM Conference, Chicago, Illinois, 1995.

RCAR, (2001), Research Council for Automobile Repairs, A Procedure for Evaluating Motor Vehicle Head Restraints, pp. 2.

Sekizuka, M. (1998). Seat Designs for Whiplash Injury Lessening. Toyota Motor Corporation.

Schmitt KU, Muser MH, Niederer P. (2001). A New Neck Injury Criterion Candidate for Rear-end End Collisions Taking Into Account Shear Forces And Bending Moments. Paper No. 124, Proc. Of Int. 17th ESV Conference, 2001.

Schmitt KU, Muser MH, Niederer P, Waz FH. (2002). A Proposal For A Neck Protection Criterion For Low Speed Rear-end End Impacts. Traffic Injury Prevention 3(2), 2002.

Shin M, Park K, Park G. (2003). Occupant analysis and seat design to reduce neck injury from rear-end end impact. International Journal of Crashworthiness, Volume 8, Issue 6. pp 573-581.

Svensson M. Y. (1993). Neck Injuries In Rear-end End Car Collisions - Sites And Biomechanical Causes Of The Injuries, Test Methods And Preventive Measures, Department Of Injury Prevention, Chalmers University Of Technology, Göteborg, Sweden.

Svensson M. Y. (2003) Injury Biomechanics of the Cervical Spine in Car Collisions - Some Needs for Further Research. Crash Safety Division, Chalmers University Of Technology, Göteborg, Sweden. Pain Res Manag. 2003 Spring;8(1):44-6.

Siegmund GP, King DJ, Lawrence JM, Wheeler JB, Brault JR, Smith TA. (1997) Head/Neck Kinematics Response of Human Subjects In Low Speed Rear-end End Collisions. Paper No. 973341, Proc. Of 41st Int. Stapp Car Crash Conference, 1997, pp 357-385.

Sturzenegger M, Radanov BP, Di Stefano G. (1995). The Effect Of Accident Mechanisms And Initial Findings On The Long Term Course Of Whiplash Injury. J. Or Neurology 242.

Tanaka M, Yoshida H, Tsutsumi S. (2003). A New Seat System Developed By K-D Neck Model to Reduce Whiplash Injuries. 18th ESV Government Focal Points 2003.

Temming J, Zobel R. (2000). Neck Distortion Injuries in Road Traffic Crashes (Analysis of the Volkswagen Database). In: Frontiers In Whiplash Trauma, N. Yoganandan And F.A.Pintar, IOS Press, pp 118-133.

Thomson RW, Romilly DP, Narvin FDP, Macnabb MJ. (1993). Dynamic Requirements Of Automotive Seatbacks, International Congress And Exposition, SAE Paper No. 930349.

TNO (2004). TNO MADYMO BV, Madymore. Madymo Newsletter 24 June 2004.

TNO (2003). TNO MADYMO Utilities Manual Version 6.1 April 2003.

Toyota Motor Corporation, 2004, www.conceptcar.co.uk/ news/technology/cardesignnews16 php)

Viano DC, Davidsson J. (2002). Neck Displacements Of Volunteers, BioRid P3 And Hybrid lll In Rear-end Impacts: Implication To Whiplash Assessment By A Neck Displacement Criterion (NDC). Traffic Injury Prevention 3(2), 2002, 105-116.

Viano D.C, Olsen S, Locke G.S, Humer M. (2002). Neck Biomechanical Responses With Active Head Restraints – Rear-end Barrier Tests With BioRid And Sled Tests With Hybrid lll, Saab Automotive, General Motors, Chalmers University Of Technology, Lear Corporation, SAE 2002 World Congress Detroit, Michigan, 2002.

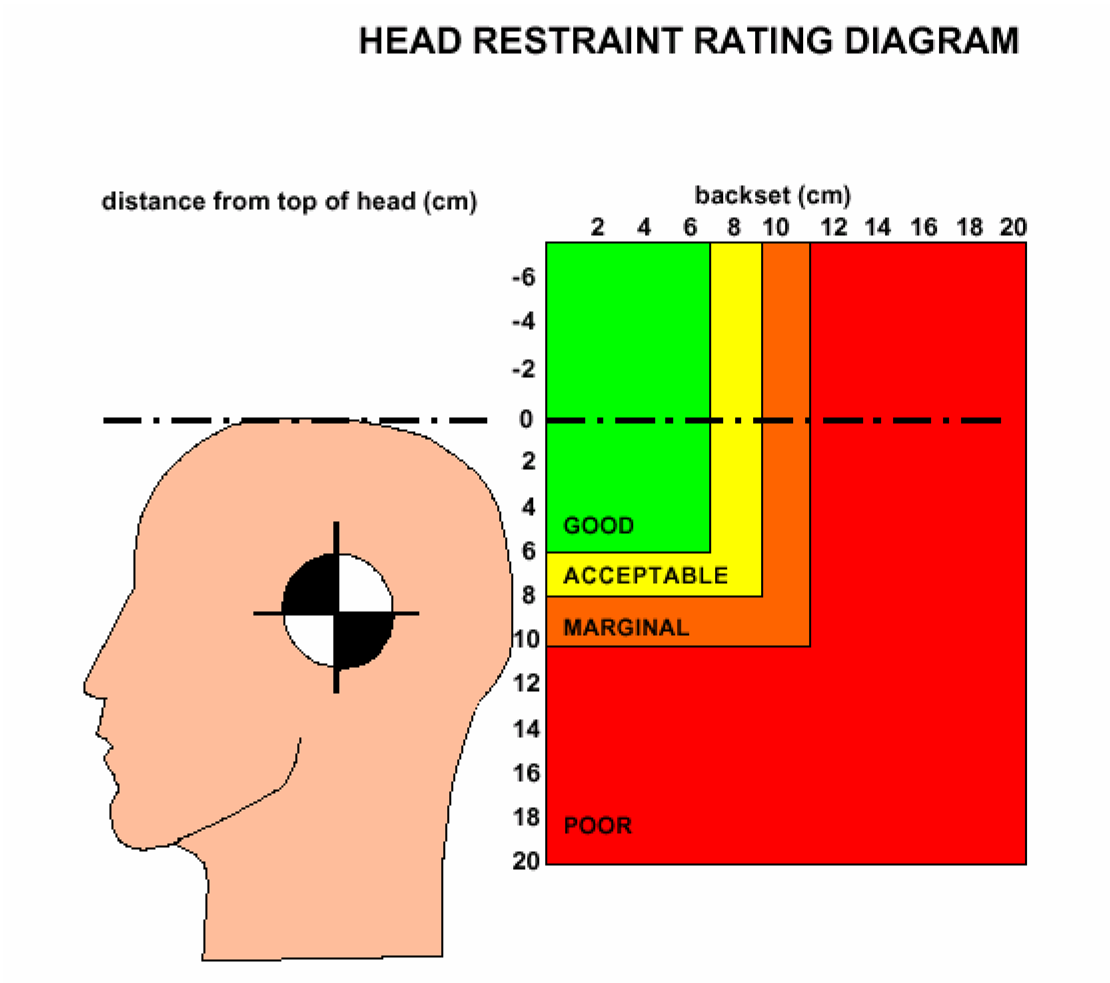
Wheeler, J. B., Smith, T. A., Siegmund, G.P., Brault, J. R., King, D.J. (1998) Validation Of The Neck Injury Criterion (NIC) Using Kinematics And Clinical Results From Human Subjects In Rear-end End Collisions. Proc. Of Int. IRCOBI Conference, 1998, pp 335-348.

Wiklund K, Larsson H. (1998). Saab Active Head Restraint (SAHR) – Seat Design To Reduce The Risk Of Neck Injuries In Rear-end Impacts, International Congress And Exposition, Detroit, Michigan, 1998.

Yang, J. (1997), Injury Biomechanics In Car Pedestrian Collisions: Development, Validation And Application Of Human-Body Mathematical Models, Department Of Injury Prevention, Chalmers University Of Technology, Göteborg, Sweden.

Zuby D.S, Farmer C.M, Avery M, (2003). The Influence Of Crash Pulse Shape On BioRid Response, Insurance Institute Of Highway Safety, Motor Insurance Repair Centre – Thahcham, IRCOBI Conference- Portugal, 2003.

Appendix-I


HEAD RESTRAINT RATING DIAGRAM
distance from top of head (cm)
backset (cm)
2 4 6 8 10 12 14 16 18 20
-6
-4
-2
0
2
4
6
8
10
12
14
16
18
20
GOOD
ACCEPTABLE
MARGINAL
POOR

Appendix-II

Results: Whiplash Injury Related Parameters for case-1,2,3 & 4.

Case-1: Analysis of the influence of the crash pulses on whiplash injury related parameters for various body sizes.

ID	Dummy %ile	crash pulse	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max) Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Head (max) [m/s^2] @ [ms]	Ax Pelvis (max) [m/s^2] @ [ms]	Ax Chest (max) [m/s^2] @ [ms]
1		low	11@85	0.139@ 109	86	47@ 76	1.50@ 91	146@ 107	61@ 73	48@ 74
2	5	med	22@62	0.387@ 82	63	105@ 62	1.99@ 74	245@ 82	153@ 73	107@ 62
3		high	35@64	0.48@73	55	152@ 64	2.37@ 66	246@75	236@ 65	146@ 57
4		low	9@89	0.08@ 101	97	38@ 89	1.44@ 107	125@ 124	45@ 91	45@98
5	50	med	18@80	0.17@ 113	82	76@ 80	2.22@ 92	195@ 108	113@ 77	75@ 76
6		high	26@79	0.207@ 97	81	107@ 78	2.62@ 89	229@ 104	170@ 80	103@ 74
7		low	11@97	0.134@ 137	98	48@ 94	1.54@ 111	101@ 136	48@ ‘99	40@ 95
8	95	med	16@91	0.21@ 115	93	66@ 91	2.22@ 104	139@ 130	120@ 84	60@ 50
9		high	34@91	0.25@ 105	94	148@ 90	2.86@ 105	176@ 125	167@ 76	83@ 81

Case-2: Analysis of the influence of head to headrest distance on whiplash injury related parameters for various body sizes.

ID	Dummy %ile	crash pulse	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max) Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Head (max) [m/s^2] @ [ms]	Ax Pelvis (max) [m/s^2] @ [ms]	Ax Chest (max) [m/s^2] @ [ms]
10		low	9.7@74	0.06@8 7	75	45 @ 74	1.05 @ 82	122 @ 99	61 @ 73	47 @ 74
11	5	med	10.5@62	0.15@6 7	55	49 @ 62	1.11 @ 68	194 @ 78	150 @ 72	97 @ 60
12		high	21.6@64	0.275@ 83	48	101 @ 64	1.47 @ 66	202 @ 74	233 @ 65	127 @ 57
13		low	6@69	0.067@ 89	77	28 @ 68	0.85 @ 90	92 @ 110	44 @ 91	40 @ 97
14	50	med	9.1@79	0.125@ 104	57	39 @ 79	1.39 @ 88	149 @ 105	113 @ 77	71 @ 76
15		high	16.5@78	0.156@ 103	50	71 @ 74	1.92 @ 85	201 @ 101	170 @ 79	99 @ 74
16		low	4.12@64	0.06@1 36	59	20 @64	0.49 @ 107	64 @ 136	47 @ 98	35 @ 96
17	95	med	10.5@87	0.132@ 91	46	48 @ 86	1.5 @ 102	121 @ 135	119 @ 83	59 @ 83
18		high	17.6@95	0.184@ 86	40	70 @ 85	2.18 @ 98	142 @ 123	116 @ 76	80 @ 80

Case-3: Analysis of the influence of seatback recliner stiffness on whiplash injury related parameters for various body sizes.

ID	Dummy %ile	Rec. Stiff	NIC $[m^2/s^2]$ @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max) Head-T1 $[m/s^2]$ @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Head (max) $[m/s^2]$ @ [ms]	Ax Pelvis (max) $[m/s^2]$ @ [ms]	Ax Chest (max) $[m/s^2]$ @ [ms]
19		low	41@65	0.49@74	59	179@65	2.49@68	251@65	230@65	128@58
20	5	med	35@64	0.48@73	55	152@64	2.37@66	246@75	236@65	146@57
21		high	32@63	0.48@73	54	137@63	2.24@65	279@73	240@65	157@57
22		low	23@91	0.216@108	93	81@91	2.87@99	229@117	168@79	93@75
23	50	med	26@79	0.207@97	81	107@78	2.62@89	229@104	170@80	103@74
24		high	23@76	0.212@92	72	96@71	2.47@83	243@99	173@78	111@70
25		low	37@105	0.29@121	106	145@104	3.26@114	162@139	159@83	77@82
26	95	med	34@91	0.25@105	94	148@90	2.86@105	176@125	167@76	83@81
27		high	42@85	0.28@97	84	191@85	2.69@91	175@119	172@79	91@79

Case-3 (cont..)

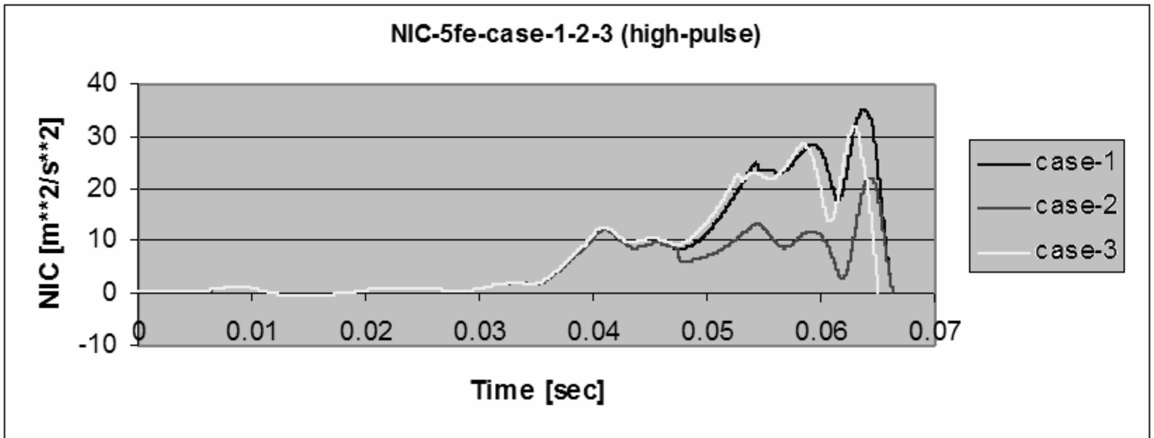
ID	Dummy %ile	Rec Stiff	Tmax recliner [N]	SB dis [deg]	Fmax SB Up [N]	Fmax SB Mid [N]	Fmax SB Dwn [N]	PP MaxF [N]
19		Low	1887@88	12.2@103	1211@62	1934@61	2297@66	953@52
20	5	Med	2348@78	9.3@93	1375@60	1996@61	2294@66	968@45
21		High	2833@76	7.43@87	1521@59	2061@61	2292@66	986@45
22		Low	2351@95	24@135	1752@95	2828@78	3861@80	1490@80
23	50	Med	2911@105	19.2@122	1828@81	3008@78	3851@80	1585@75
24		High	3311@96	15.9@115	2056@75	3127@77	3828@80	1563@73
25		Low	2492@96	29@158	2311@106	2886@78	4442@84	1448@92
26	95	Med	3079@117	23.6@145	2245@101	2996@80	4432@84	1452@85
27		High	3541@115	19.6@133	2362@100	3125@79	4407@84	1313@82

Case- 4: Analysis of the effect of lowering the headrest height for 95th percentile male dummy on whiplash injury related parameters.

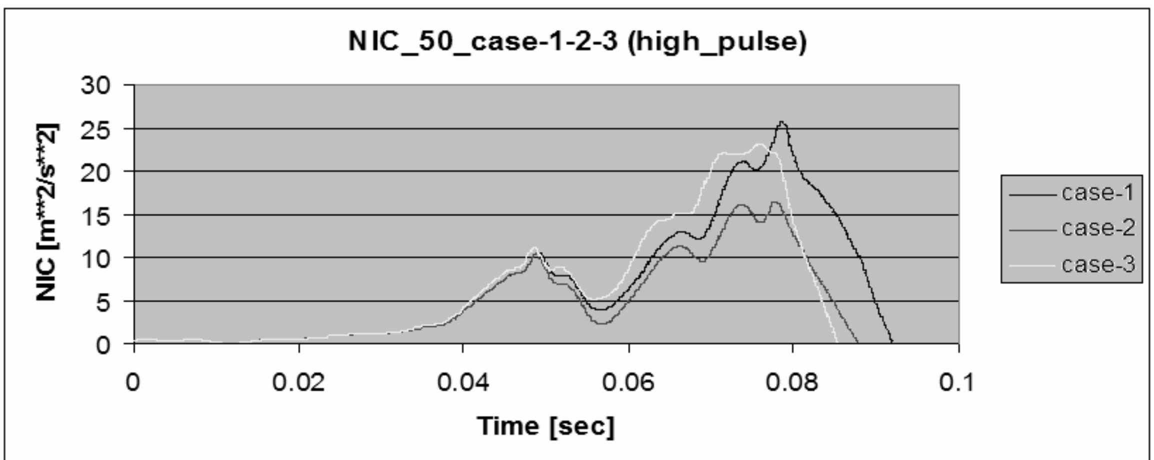
ID	Dummy %ile	crash pulse	NIC [m^2/s^2] @ [ms]	Nkm @ [ms]	Head to HR time [ms]	Arel (max) Head-T1 [m/s^2] @ [ms]	Vrel (max) Head-T1 [m/s] @ [ms]	Ax Head (max) [m/s^2] @ [ms]	Ax Pelvis (max) [m/s^2] @ [ms]	Ax Chest (max) [m/s^2] @ [ms]
28		Low	11@ 107	0.124@ 137	108	48@ 94	1.85@ 116	127@ 137	48@ 99	40@ 95
29	95	med	19@ 99	0.233@ 118	101	70@ 99	2.76@ 110	180@ 128	120@ 83	60@ 50
30		high	34@ 91	0.32@ 115	101	148@ 91	3.4@ 109	219@ 121	167@ 76	83@ 81

Appendix-III

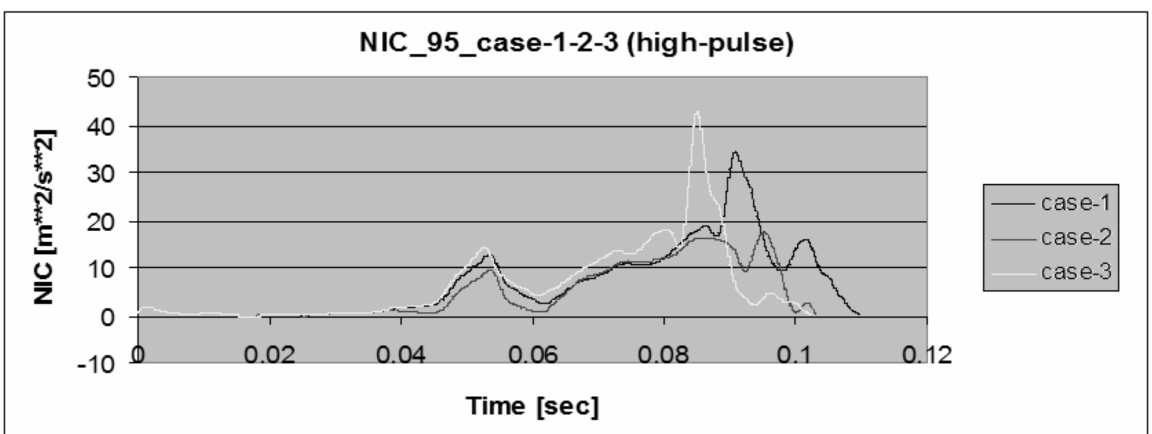
Time History of the Neck Injury criteria for all occupant sizes in case-1,2 & 3 (only high severity pulse)



NIC 5th percentile female Case-1-2-3 (high pulse)



NIC 50th percentile male Case-1-2-3 (high pulse)



NIC 95th percentile male Case-1-2-3 (high pulse)

Appendix-IV

Scaling Parameters used by MADYMO/Scaler

Parameter	Remark
1. Weight	Used to calculate correction factors
2. Standing height	Used to calculate correction factors
3. Shoulder height	NOT used to calculate first scaling factors
4. Armpit height	NOT used to calculate first scaling factors
5. Waist height	NOT used to calculate first scaling factors
6. Seated height	Used to calculate correction factors
7. Head length	
8. Head breadth	
9. Head to chin height	
10. Neck circumference	
11. Shoulder breadth	Used to calculate correction factors
12. Chest depth	
13. Chest breadth	
14. Waist depth	
15. Waist breadth	
16. Buttock depth	
17. Hip breadth standing	
18. Shoulder to elbow length	
19. Forearm-hand length	
20. Biceps circumference	
21. Elbow circumference	NOT used to calculate first scaling factors
22. Forearm circumference	
23. Wrist circumference	NOT used to calculate first scaling factors
24. Knee height seated	
25. Thigh circumference	NOT used to calculate first scaling factors
26. Upper leg circumference	
27. Knee circumference	NOT used to calculate first scaling factors
28. Calf circumference	
29. Ankle circumference	NOT used to calculate first scaling factors
30. Ankle height, outside	NOT used to calculate first scaling factors
31. Foot breadth	
32. Foot length	
33. Hand breadth	
34. Hand length	
35. Hand depth	

Appendix-V

Flowchart of MADYMO/Scaler