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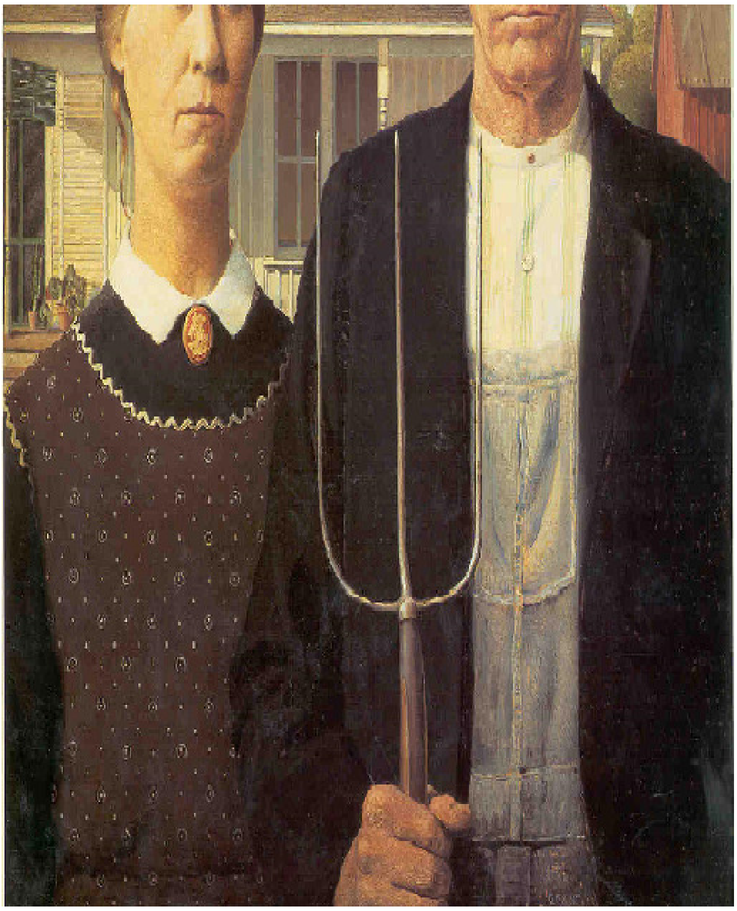

Neck Muscle Influence in Rear Impacts

A Sled Test Study using the BioRID

Master's Thesis in the Master's programme Biomedical Engineering

ISABELLE STOCKMAN

Department of Applied Mechanics
Division of Vehicle Safety
CHALMERS UNIVERSITY OF TECHNOLOGY
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ABSTRACT

Whiplash injuries resulting from foremost low velocity rear-end car impacts have become a major health problem and require expensive and time-consuming rehabilitation. Women are at greater risk of getting whiplash injuries compared to men and more prone to long-term injury. There is no direct answer to why this gender difference exists but variation in head and neck geometry and neck strength may be a possible explanation.

The aim of this Master´s Thesis was to evaluate the influence of neck muscles in rear impacts. This has been carried out in several parts. Neck strength measurements and EMG activity in the right sternocleidomastoid muscle (SCM) were studied in a group of twelve volunteers (6F, 6M). The results were used to design a muscle substitute for the BioRID dummy. A series of rear impact sled tests with and without muscle substitute and with various seat designs was performed according to EuroNCAP whiplash test protocol.

Retraction resistance for men and women were 167±38N and 86±28N respectively. Mean onset time in the right SCM was 75±11ms. In seats without WHIPS the injury risk was reduced by 44% with muscle substitute compared to reference. Female occupants seem likely to have an earlier time of contact with the head rest, higher head acceleration and a higher risk of injury. As this is confirmed by previous findings, the role of muscles may be one of the most important in understanding the difference in risk in gender. Furthermore, according to the results of this thesis females may not be at higher risk in cars equipped with WHIPS.

Key words: whiplash, neck injury, females, neck strength, electromyography, rear impact, sled test, muscle influence, sternocleidomastoid muscle, BioRID, NIC

Contents

Preface

In this Master´s Thesis work neck muscle influence in rear impacts has been investigated with the aim to further explain the gender difference in whiplash injury risk. The project has been carried out from January to June 2010 as a final step in my education at Chalmers University of Technology. The thesis has been performed at the Department of Applied Mechanics in collaboration with Autoliv.

I would like to thank my supervisor Ola Boström at Autoliv for your constant support, enthusiasm and sharing of your knowledge, and my examiner Mats Svensson at Chalmers for your help and good advice. I would also like to thank all the nice people I have had the opportunity to get to know, especially at Autoliv and at SAFER, who have showed their interest in my work and helped me during the way.

Göteborg, 2010

Isabelle Stockman

Notations

Δv	Change of velocity
a_{rel}	Relative horizontal acceleration between T1 and occipital joint
AIS 1	Minor injury according to the Abbreviated Injury Scale
Anterior	In front of…
BioRID	Biofidelic Rear Impact Dummy; represents a 50th percentile male (~1.77m and 77.7kg)
C7	The uppermost cervical vertebrae in the neck
EMG	Electromyography
EuroNCAP	European new Car Assessment Programme
Extension	Rearward bending of the neck
F_x	Shear force measured between T1 and head
Flexion	Forward bending of the neck
ISO	International Organization for Standardization
M_y	Bending moment measured between T1 and head
N_{km}	Neck Protection Criterion
NIC	Neck Injury Criterion
NIC_{max}	Max. NIC value during the first 150ms of an impact
Posterior	Behind of…
Retraction	Head moved rearward relative to the torso with no angular change
SCM	Sternocleidomastoid muscle. The muscle runs from the clavicle to the mastoid and acts to flex and rotate the head.
T1	First thoracic vertebrae
v_{rel}	Horizontal velocity between T1 and the occipital joint
WHIPS	Whiplash Protection System
WIL	Whiplash Injury Lessening

1 Introduction

1.1 Anatomy and range of motion of the neck

The spine, also referred to as the vertebral column, is formed by the vertebrae and divided into five regions: cervical, thoracic, lumbar, sacral and coccygeal. The major functions of the spine are to provide protection to the spinal cord, give support to the head, torso and upper extremities and to allow body movement in three planes. The cervical part of the spine is formed by seven vertebrae, named C1-C7. It gives support to the head and neck, protects neural and vascular structures of the neck, and protects the brain by acting as a shock absorber. The vertebrae consist of a vertebral body and a vertebral arch except from the C1 (atlas) and C2 (axis), the two topmost vertebrae which have a different structure in order to increase the range of motion of the head (Mordaka, 2004; Carlsson, 2010). Neighboring vertebrae are connected by a fibrous disc and ligaments which allow motion of the spine (Eriksson, 2004). Muscles connect between vertebrae and between vertebrae and other bones where the spinous and transverse processes serve as attachment points. The muscles between head and torso provide stability of the head and neck and affect the motion in rear-end impacts (Eriksson, 2004; Carlsson, 2010).

During a rear impact the vehicle is subjected to a forward acceleration causing a sudden velocity change. How this will affect the motion of the head and neck can be seen in Figure 1.

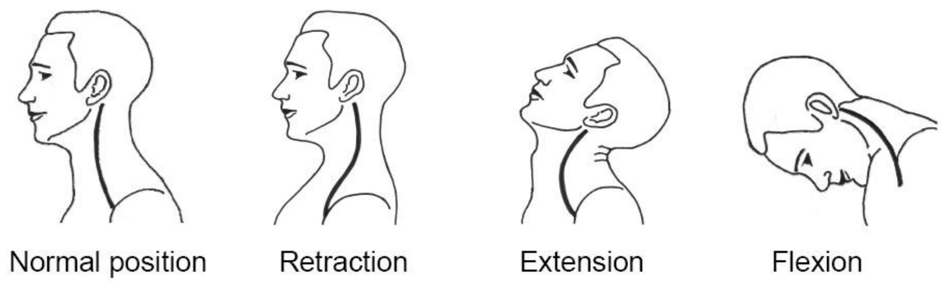

Figure 1 Whiplash motion of head and neck during a rear impact (Linder, 2001).

When the vehicle is pushed forward by the impacting car the occupant is pushed forward by the backrest of the seat. The head lags behind due to its inertia and the relative motion of the head and torso, with no angular change, leads to a retraction of the neck. In this phase the upper part of the neck is flexed and the lower part is extended which results in an S-shape of the cervical spine. The head will be bent backwards as the torso is pressed further forward and the neck will be forced into extension. This motion of the head continues until the head hits the head rest or the neck reaches its maximum range of motion. Eventually an opposite motion of the body, so called rebound, will take place. When the torso is stopped by the seat belt the head will continue forward into a flexed neck posture. This is the typically injurious motion of the neck called whiplash motion. Whiplash injuries are located in the soft tissues of the neck and are therefore not possible to detect by X-rays or Magnetic Resonance Imaging. Because of this, and due to the complicated structures of the neck, it is difficult to decide the location of the injury and the cause of the symptoms (Magnusson et al., 1999; Carlsson, 2010).

1.2 Whiplash injury protection system

In order to increase the protection of the occupants in rear impacts and to decrease the risk of whiplash injury the main focus has been on improvements of the seat design. The amplitude of the rebound is dependent on the seat backrest properties and the position of the head rest will affect the head motion. By improved seat geometry, active devices that moves in a crash and by energy absorbers in the seat the relative motion of the head and the torso can be minimized and acceleration reduced (Jakobsson, 2004; Carlsson 2010). In this master´s thesis Volvo´s Whiplash Protection System (WHIPS) and Toyota´s Whiplash Injury Lessening (WIL) have been used.

1.2.1 WHIPS

A new seat design called Whiplash Protection System was introduced in Volvo cars in 1998. It was developed with improved distance between the head and the head rest, a more even and close support for the back, reduced occupant acceleration and lower forward rebound into the seat belt. During a rear impact the backrest of the seat first moves in translational motion and than in reclining motion (Figure 2). During this motion, deformations elements in the recliner mechanisms deform and absorb energy resulting in reduced occupant acceleration and rebound (Jakobsson, 2004).

Figure 2 Whiplash Protection System (WHIPS). Normal position → Translational motion → Reclining motion

1.2.2 WIL

The Toyota Whiplash Injury Lessening (WIL) system has no active parts. It works with improved geometry and a softer backrest. The head rest, especially the metal frame, has been moved forward and upward, and the upper part of the backrest frame has been moved rearward away from the upper torso compared to previous seats. The seat surface is remained to support the upper torso the same way as in previous seat design. During a rear impact the upper torso sinks into the backrest while the head rest meets the head of the occupant. The pelvic support at the lower part of the backrest initiates the lower part of the torso to rebound first and therefore helps to prevent neck extension (Sekizuka et al., 1998).

1.3 EuroNCAP whiplash test procedure

The mechanisms by which whiplash injuries are caused are not completely understood but it is well known that seat and head rest design can strongly influence the risk of injury. The European New Car Assessment Programme (EuroNCAP) was established in 1997 and provides assessments of the safety performance of new cars. The EuroNCAP whiplash test procedure promotes the best seat design and is based on both the geometrical aspects of the seat as well as the dynamic performance of the seat and head rest during an actual crash. This dynamic response is assessed using a seat mounted on a sled, subjected to three different pulses: low, medium and high, carried out with the BioRID 50th percentile male test dummy. The low pulse give a velocity change, Δv, of 16km/h and a peak acceleration of 5g while the medium pulse is a Δv = 16km/h pulse with a mean acceleration of 5.5g (high pulse Δv = 24km/h peak acceleration of 7.5g). The shapes of the low and medium pulses can be seen in Figure 3.

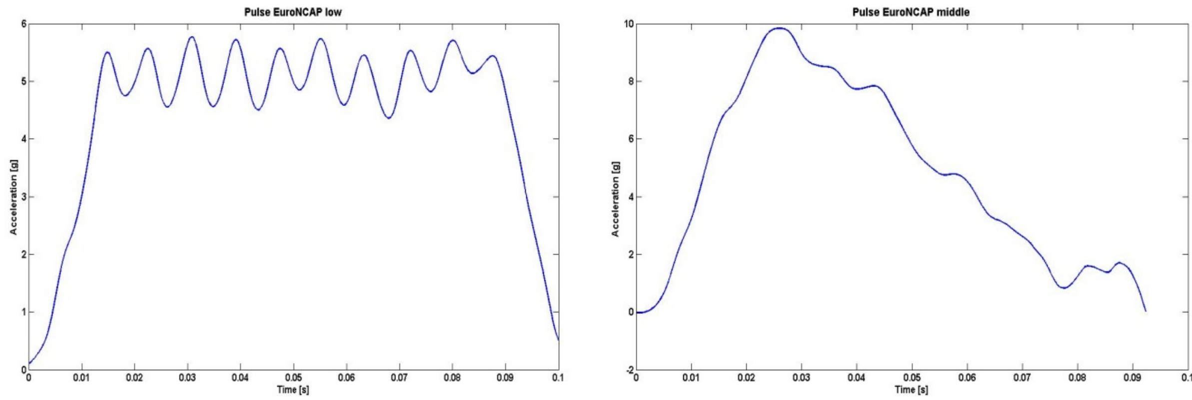

Figure 3 EuroNCAP low pulse (left) and middle pulse (right). Both with a Δv=16km/h.

The assessment is based on seven seat performance criteria:

- Neck Injury Criterion (NIC) (further explained in Section 1.4)
- N_{km} Criterion (further explained in Section 1.4)
- Head rebound velocity
- Upper neck force x-direction (shear force)
- Upper neck force z-direction (tension force)
- Head-to-head rest contact time
- T1 acceleration in x-direction

The positioning of the accelerometers and the neck load cell can be seen in Figure 4. All seat performance criteria encourage the basics of energy absorption by the seat, and short distance between the head rest and the back of the occupant´s head. There are two performance limits for each criterion: higher and lower. The higher performance limit (HPL) is the more demanding limit below which a maximum score is obtained. Above the less demanding lower performance limit (LPL) no points are scored. If the test value recorded falls between the HPL and the LPL the points score is calculated by linear interpolation. There is also the capping limit (CL) and if any single measured variable exceed this limit then a zero is recorded for the whole test. The maximum score for each criterion is 0.5 points and the score for the parameters are calculated for each of the pulses. The score for the first five criteria are summed plus the maximum score from either T1 acceleration or head-to-head rest contact time. There is a maximum possible score of three points for each of the pulses, therefore 9 for the overall series of dynamic tests. The dynamic score is combined with the result from the geometric assessment. The design of the head rest position can either add or subtract the score with maximum one point. The score can finally be reduced with three points if a excessive dynamic deflection of the backrest is observed during the high pulse test or where there is evidence of a dummy artefact (minus two points). The overall whiplash raw score is scaled to four points and included in the Adult Protection score as of January 2009. A score of 0 to 1.49 scaled points is “Poor” (coloured red), a score of 1.50 to 2.99 is “Marginal” (orange), and finally a score of 3.0 to 4.0 is “Good” (green) (van Ratingen et al., 2009; EuroNCAP, 2010).

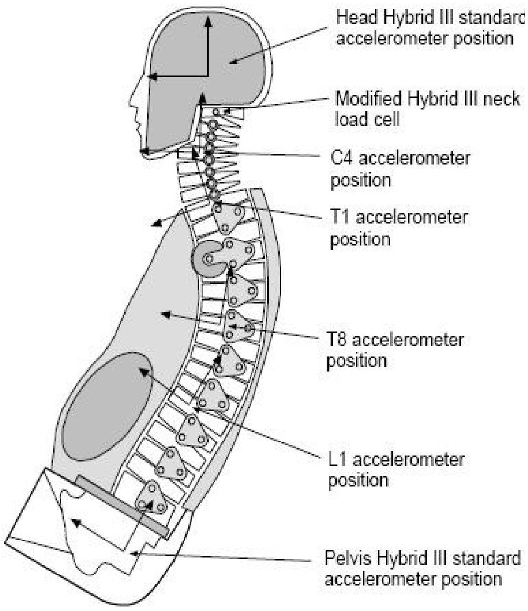

Figure 4 The BioRID and instrumentation (Davidsson, 1999).

1.4 Injury criteria

An injury criterion is a function of physical variables in the occupant related to a specific injury and generally proposed and validated based on experimental studies. For each criterion values for which no one is injured, values for which someone may be injured, and values for which all occupants are injured should be defined (Eriksson, 2004; Carlsson, 2010).

1.4.1 The Neck Injury Criterion (NIC)

The Neck Injury Criterion (NIC) was formulated by Boström et al. (1996) and is based on the relative velocity and acceleration between the upper and the lower neck. The NIC is calculated as

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

where a_{rel} is the relative horizontal acceleration between the first thoracic vertebrae T1 and the occipital joint and v_{rel} is the horizontal velocity between T1 and the occipital joint. A injury threshold was proposed; NIC values lower than $15\text{m}^2/\text{s}^2$ do not result in soft-tissue neck injuries, which has been found to work well and are still in use. NIC_{max} is defined as the peak value during the first 150ms of impact and is the most widely used criterion for soft-tissue neck injuries in rear impact tests today.

1.4.2 The Nkm Criterion

The N_{km} was proposed by Schmitt et al. (2002) to assess neck injuries in rear impacts and combines moments and shear forces. It is calculated by

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

where F_x represents the shear force, M_y the flexion/extension bending moment and F_{int} and M_{int} the critical intercept values for the force and moment, respectively. F_x and M_y are obtained from the upper neck load cell. Four different load cases can be obtained:

- N_{fa} for flexion and anterior (positive) x-direction
- N_{fp} for flexion and posterior (negative) x-direction
- N_{ea} for extension anterior x-direction
- N_{ep} for extension and posterior x-direction

Positive shear is defined as when the head is moved backwards relative to the uppermost cervical vertebra. The injury threshold value for each load case is 1.0. For $N_{km} = 0.8$ the risk of neck injury lasting more than one month is approximately 20%. For $N_{km} < 0.37$ the risk for whiplash symptoms lasting more than one month was less than 10% (Schmitt et al., 2002; explained by Carlsson, 2010).

1.5 Electromyography

Electromyography (EMG) is a technique for evaluating and recording the electrical activity produced by muscles. Surface EMG represents the sum of the electrical contributions made by active motor units (MUs). The muscle activity is detected by electrodes placed on the skin over the muscle and is often considered a global measure of motor unit activity since traditional two electrode recording cannot detect activity at the level of single MUs. The global characteristics (amplitude and power spectrum) of the surface EMG depend on the properties of the muscle fibers and on the timing of the MUs action potentials. When the muscle is voluntarily contracted action potentials begin to appear. As the muscle contraction strength is increased, more muscle fibers will produce action potentials and an increase in EMG activity will occur (Farina et al., 2004).

2 Literature Review

The purpose of this literature review was to act as a foundation to this Master´s Thesis project and to resume relevant knowledge about the neck muscles, foremost the sternocleidomastoid muscle and their behaviour and features related to a range of parameters. Collision warning systems are also of interest in this project and have been summarized very briefly.

2.1 The role of the muscles in rear impacts

The mechanical properties of muscles can be divided into active and passive phases. The passive force is generated when the non-activated muscle is stretched beyond its resting length. Active force is developed when the muscle is excited by the nervous system. One can divide the muscle activation process into three phases. When stimulus occurs, sensory receptors detect the stimulus and transfer a signal to the central nervous system. The nervous system responds and electrical activity in the muscle starts. The next phase is the time delay between stimulus and muscle onset known as reflex time. The third phase is an electromechanical delay between the onset of muscle activity and onset of muscle force generation. The time interval from muscle force onset and peak muscle force is called the rise time (Mordaka, 2004). The SCM muscles are known to generate the greatest EMG response in rear impacts and show a decrease in time to onset when the acceleration increases (Kumar et al., 2005). The onset latencies in the SCM muscles observed in whiplash studies are shorter than the voluntary latencies observed in forewarned reaction-time experiments. This indicates that the neck muscles are activated reflexively rather than voluntarily during whiplash perturbations (Siegmund et al., 2003). Brault et al. (2000) reported that the SCM muscles were more active in the retraction phase than in the rebound phase in low speed rear-end collision tests with human test subjects. The SCMs were only active in the initial portion of the rebound phase. They also found that the cervical paraspinal muscles were active during the retraction phase at a lower normalized level than the SCM muscles and then increased their activity during rebound.

2.1.1 Differences in neck muscle strength between men and women

The neck muscles generate head movements and assist in maintaining the stability of the cervical spine. The static function of a muscle, i.e. the moment-generating capacity of a muscle, is the product of its moment arm and maximum isometric force (Vasavada et al., 1998). Several studies have been done where differences in isometric neck muscle strength between men and women have been investigated. Despite nearly comparable head weights between men and women, neck muscle forces in women are only about half of those in men. Contributing factors in chronic neck pain are the sustained muscle contraction required to hold the head in different positions and fatigue caused by muscular weakness. The relatively weak neck muscles in women, compared to men, can cause muscular fatigue syndrome resulting in higher presence of chronic neck pain (Cagnie et al., 2007). Vasavada et al. (2008) studied differences in head and neck geometry and neck strength in height-matched men and women. Maximum isometric strength in flexion and extension was measured using a hand-held dynamometer, a method that has an average reliability of 0.88. Flexion strength was measured supine, extension prone, and each measurement continued for ~3 s or until the test subject could no longer resist. Based on 14 pairs of men and women who were matched by standing height and neck length they found that women had significantly smaller external neck and vertebral dimensions than men. Women did also have lower neck strength than men in both flexion and extension. Corrected neck strength was calculated which accounted for the head mass estimated from head circumference and body mass by regression equations. Neck strength for women for flexion and extension was 82±26N and 173±31N respectively.

Corrected neck strength for women was for flexion 127±25N and 219±32N for extension. For men neck strength was 149±44N in flexion and 244±69N in extension and corrected neck strength was 197±44N and 292±69N respectively. When corrected for head mass, neck strength for women in flexion were 68±25%, and in extension 80±31%, of male neck strength. Cagnie et al. (2007) did a study where ninety-six healthy volunteers divided into four age groups (20-29; 30-39; 40-49; 50-49) each consisting of 12 men and 12 women, and a group of 30 women with chronic neck pain participated. Maximal isometric strength of the cervical muscles was tested for flexion and extension by using a dynamometer, and moments were resolved about axes through the midpoint of the line between the sternal notch and the C7 spinous process (referred to as C7-T1). The test subjects were positioned prone for flexion measurements and supine for extension. A seat belt at shoulder height was used to prevent any extra strength effect from trunk musculature and in the supine position test subjects crossed their arms to prevent movements of the thorax. The average maximum moments resolved at C7-T1 for women were 16.6±3.6Nm for flexion and 26.5±6.2Nm for extension. The average maximum moments resolved at C7-T1 for men were 24±6.0Nm for flexion and 36.4±7.7Nm for extension. A level of significance of $P<0.05$ was used for the analyses. No significant difference was found in muscle strength with different age groups. The average maximum moment for extension was lower in the group of women with chronic neck pain (22.3±5.6Nm) compared with the healthy female test subjects, but no significant differences in flexion strength were found.

Larochelle et al. (2009) did a study where the purpose was to investigate the influence of test position on muscle fatigue and strength in neck extensors and flexors in a group of twenty-five women without neck pain. Two different test positions, sitting and lying, were tested. The test subjects sustained isometric contractions during 20s and 10s at 25% and 75% of their maximal voluntary contraction (MVC) respectively. Surface electromyography was used to measure the change over time of the median frequency of the power spectrum of the myoelectric signal of SCM. In sitting position a dynamometric device was used and the output was the force produced (N). In sitting and supine position the test subjects had their arms crossed over the chest which was stabilised by two belts crossing the sternum. In the lying position subjects laid on a table with a load cell able to record forces in both flexion and extension. The load cell was fixed under the headrest which was linked to the table by a hinge. A headrest offered support of the head and resistance in neck extension. A non-elastic strap placed above the forehead stabilised the head and provided resistance during neck flexion. The effect of gravity was considered by subtracting the weight of the head in extension measurements and adding in flexion. The strength output was the torque produced (Nm). Torque measurements were converted into forces by division of the test subjects´ external lever arms to be comparable with the strength output in sitting position. $P<0.05$ was considered statistically significant. Fatigue of the SCM was not significantly affected by test position at either low or high loads. Mean strength values for flexion were for sitting position 85±33N and for lying position 97±22N. Mean strength values for extension were 137±42N for sitting position and 163±38N for lying position. In both flexion and extension neck strength was significantly higher in lying compared to sitting.

Vasavada et al. (2001) measured maximum moments in 11 men and 5 women to analyse how neck muscle moments vary along the cervical spine. The test subjects were seated upright with their heads linked to a 6-axis load cell by a device with eight pads that were tightened around the head. The moments produced when the test subject pushed against the pads in different directions were resolved about axes through the midpoint of the line between the C7 spinous process and the sternal notch (C7-T1). In addition to C7-T1 two other points were used to calculate the variation of maximum moments with vertical distance along the spine.

The first point, referred to as the mastoid is the midpoint of the mastoid processes approximately at the level of the skull and the C1 joint. The second point, C4, is the equivalent center of rotation calculated from the intervertebral kinematics of a biomechanical model. C4 is located in the C4 vertebral body in the middle between the C7-T1 point and the median plane (i.e. the imaginary plane that goes vertically from the top to the bottom of the body, dividing it into left and right parts) projection of the tragus of the ear. The average maximum moments resolved at C7-T1 for the men were 52±11Nm for extension and 30±5Nm for flexion and for the women 21±12Nm and 15±4Nm respectively. The data for moments resolved at different levels along the cervical spine were normalized by the magnitude of the moment resolved at C7-T1. The data was also linearly regressed according to vertical distance along the cervical spine where C7-T1 was defined as 0 and the mastoid process defined as 1. Maximum moments were divided by head mass or inertia in order to take the test subject size into consideration. Adjusted moments generated by the women still were only 40-60% of the moments generated by the men. The differences in maximum moments between genders do not correspond to differences in the demands made on the neck muscles by gravitation and inertia forces. The mass and inertial properties are only slightly greater in men than in women, still maximum moments generated by men are more than two times those generated by women. This suggests that mechanical demands on the neck muscles in women may be closer to their maximum moment-generating capacity. When operating closer to their maximum functional capacity, fatigue in the neck muscles could increase and the ability to actively stabilise the cervical spine could decrease. In both flexion and extension moments resolved at C4 were 30-40% lower than those resolved at C7-T1 whereas moments resolved at the mastoid were 50-60% lower. The results confirmed a linear decrease of moment magnitude as the vertical position of the point about which moments were resolved varied from the lower cervical spine to the mastoid process.

2.1.2 Electromyographic response time and impact awareness

Awareness of an impact can affect the kinematics of head movement in a simulated rear end impact and refers to the anticipation of an event. Such anticipation can be divided into temporal, event and amplitude awareness. Temporal awareness refers to whether the subject knows about the exact timing when an event will occur. Event awareness and amplitude awareness describes whether the subjects knows an event will occur and the magnitude of the event respectively (Hernández et al., 2005). Awareness affects the muscle contraction timing in which aware occupants contract muscles prior to impact and unaware occupants contract their muscles reflexively in response to impact (Stemper et al., 2006). Magnusson et al. (1999) studied cervical electromyographic activity during low-speed rear impacts in a group of eight male subjects. The subjects were seated on a car seat mounted on a sled with their legs and feet in driving position and hands in the lap. The forward acceleration of the sled was given by a spring under tension which was set not to exceed 0.5g (4.9m/s^2). The acceleration of the sled generated a centrally triggered response when the back hit the backrest. Surface EMG activity was measured over trapezius and SCM. Each subject underwent four rear impacts: two expected and two unexpected. For the expected impacts a countdown was done and the subjects were free to prepare any way they wanted. The unexpected impacts were applied at irregular times and the test subjects, equipped with earplugs, knew that an impact would come but not when. The reaction times of the muscles were referred to the onset of the sled movement. They found that there was a significant difference between trunk and head acceleration and the acceleration of the head was more than twice the magnitude of that of the sled. The average muscle reaction time from sled acceleration was 73.3±17.7ms and from head acceleration 20.4±19.6ms for SCM. There was no significant difference ($P<0.05$ was considered statistically significant) between the expected and unexpected impacts. The duration from onset to peak of the first EMG burst was on average 42.8±29.9ms for SCM and the number of peaks during the duration of muscle activity was 1.8±0.7. Another similar study was performed by Hernández et al. (2006) where twenty-nine adults (17M, 12F) were exposed to three rear impacts: two unexpected impacts causing chair accelerations of $4.5 m/s^2$ and $10.1 m/s^2$, and one expected impact ($10.1 m/s^2$). For the unexpected impacts the subjects were aware that there would be an impact but they did not know the timing or the magnitude of the impact. For the expected impact the timing of the impact and the magnitude, in terms of slow or fast impact, were given. Surface EMG electrodes were used to measure muscle activity in both SCM muscles. Onset was relative to the onset of the sled and defined as the time in which 5% of the peak value occurred. Each test subject performed a maximum isometric flexion force and the corresponding peak value of the generated EMG was recorded. The EMG amplitudes for the SCM muscles recording in the sled tests were normalized against the maximum value and expressed as percentage of maximum voluntary contraction. Normalized EMG activity was 2-3 times higher in women than in men. They found an increased EMG activity with an increase in impact magnitude. The magnitude of normalized EMG activity was 3-4 times higher in the fast unexpected impacts compared to the slow unexpected impacts. There were no differences in the magnitude of the muscle response due to impact awareness. Mean onset time of the muscle activity in SCM was 131±132ms in the slow unexpected impact, 99±51ms in the fast unexpected impact and 87±50ms in the fast expected impact. There were no significant differences in muscle response regarding temporal awareness.

Brault et al. (2000) assessed the potential for cervical muscle injury from rear-end automobile collision in a study where forty-two human subjects (21M, 21F) were exposed to rear-end collisions of 4km/h and 8km/h speed change on the target vehicle. Kinematic response of head and torso and surface EMG in the SCM was measured. The subjects were instructed to relax prior to impact and were not aware of the timing or magnitude of the impact. All visual and auditory cues of the impending impact were eliminated. Muscle onset time in SCM from bumper contact was for females 87±10ms with a speed change of 4km/h and 79±9ms with a speed change of 8km/h. For males the onset time was 95±8ms and 83±8ms respectively. Onset was defined as the time when muscle activity reached 10% of the peak EMG. The shorter response time at 8km/h could be explained by the shorter time interval between bumper contact and occupant acceleration at higher speed change. The body was accelerated sooner by the backrest at higher speed change collision. The first phase of muscle activity was co-contraction of flexor and extensor muscles and occurred in the retraction phase. This may be a reflexive attempt to stiffen the neck in order to minimize the relative motion between head and neck. The rearward retraction of the head relative to the C7/T1 resulted in lengthening of the activated SCM. The magnitude of the muscle response increased with increasing speed change.

Siegmund et al. (2003) observed changes in head kinematics and EMG muscle activity in aware and unaware subjects exposed to multiple whiplash-like perturbations. Forty-four test subjects (21M, 23F) participated in the study. Surface electrodes were used to record muscle activity in the SCM and paraspinal muscles. The horizontal speed change of the sled was 0.5m/s over 60ms with a peak acceleration of 1.51g. Each subject underwent 11 perturbations. Half of the subjects were temporally aware and received a count-down for each perturbation. EMG onset was defined as the time when muscle activity reached 10% of the maximum value and maximal contractions were performed in flexion and extension to generate normalizing data for the muscles. Multiple perturbations did not affect the onset latency but produced an average SCM EMG amplitude decrement of 41-54%. The onset of the SCM was around 70ms. The retraction ranged from 19.6-23.9mm and was larger in the last trials compared to the first. Kumar et al. (2005) found that the head displacement is greater when the impact is unexpected and both head velocity and head acceleration are increased which may affect the injury risk. They also saw a trend towards increased amount of time before EMG onset in the expected impacts compared to the unexpected. Stemper et al. (2006) implemented a head-neck computational model subjected to rear impacts with a speed change of 10.5km/h in unaware and aware condition. In the aware case the occupant reached maximum muscle contraction prior to impact. In the unaware case the occupant implemented contraction levels prior to impact to maintain upright position. Both the aware and the unaware occupant demonstrated retraction, extension and rebound phase but in the aware occupant S-shaped spinal curvature was not present in the retraction phase and head to T1 extension was decreased by 63%.

2.2 Pre-crash safety systems for rear-end collisions

In 2002 Mercedes-Benz marketed their preventive occupant protection system PRE-SAFE® which activates before the actual impact occurs and improves the safety for the car occupants. In 2009 Bogenrieder et al. published a paper about the Mercedes-Benz approach to integrate rear-end accidents into the PRE-SAFE® system by mounting a radar sensor in the rear bumper. The sensor should focus mainly on the area right behind the car due to the fact that in most cases the impacting vehicle approaches in the same lane. There are two main aspects that have to be considered. First, whether the system shall be designed to send any warning signals to either the vehicle occupants or to the approaching vehicle, or not. And second, the time period that the system needs to activate its functions. An issue is that Mercedes-Benz studies with driving simulator test showed that in order to have enough time for average reaction delay and appropriate and effective reaction a warning signal must be triggered 2.6 seconds before the predicted impact. In speeds of 40-50km/h the warning has to start when the distance to the approaching vehicle is 29-36m, and in city traffic this will probably result in very frequent warning and “false alarm”. Due to this warning dilemma Mercedes-Benz approach to improve safety in rear-end impacts is not by sending warning signals instead they focus on increase brake force, reversible seat belt tensioning, and activation of the active headrest. In the seat belt tensioning case an electric motor in the belt system tightens the belt and fixes the occupants closer to the seat. This system should be activated at approximately 100ms before the collision in order to leave enough time for the seat belt tensioners.

Matsubayashi et al. (2007) presented Toyota´s approach to a pre-crash safety system for rear-end collision. The system was developed to alert the driver of the approaching vehicle and to reduce whiplash injury. With millimeter-wave radar installed in the rear bumper, the system can detect a vehicle approaching closely from behind. If its judgement is that the risk of impact is high a hazard light are flashed as a warning signal to the driver of vehicles approaching from behind, and the headrest is moved forward toward the head of the occupant before the collision occurs. There is a head detection sensor in the surface layer of the headrest to ensure that the head is not pushed more than necessary. The headrest can return to its original position after been activated and can be re-used. Tests have been done to verify the effect of the pre-crash system. In a sled test with the BioRID II dummy (Δ v = 16km/h) the NIC value was reduced by approximately 50% with the pre-crash headrest. Verification of the effect of the pre-crash hazard lights was performed by a test where two vehicles with a speed of 45km/h and a distance of 18m apart were driven one behind the other. The reaction time of the driver in the following vehicle was reduced with 20% (from ~1550ms to ~1250ms) with automatic flashing of the hazard lights.

2.3 A comparison of visual, auditory and tactile warning signals

Scott et al. (2008) did a study where they examined the effectiveness of visual, auditory and tactile warning signals as a function of warning timing, relative no warning in rear-end collision prevention. The visual warning was a triangular array (5*5cm) of red optoelectronic light emitting diodes located on the simulated dashboard. The auditory warning was a 75dB, 2000Hz auditory tone from three speakers located on the dashboard, and the tactile warning was delivered via a waist belt to simulate that the stimuli originated from the driver´s seatbelt. Sixteen drivers experienced the four warning conditions (no warning, visual, auditory and tactile) in a driving simulator where they had to follow a lead car on a two-lane road. The warnings were activated when the time-to-collision between the two vehicles fell below a threshold of either 3 or 5 seconds. With the early warning timing (5s TTC) the driver response time, defined as the time from the warning to deceleration initiation, was the lowest with the tactile warning and the highest with no warning. The tactile warning did also have a significant advantage over visual warning, and any warning was better than no warning. With late warning (3s TTC) there was no statistically significant advantage of the three warning modalities over the no-warning, nor did one modality have an advantage over another.

3 Method and Material

3.1 Retraction resistance measurements and EMG recordings

3.1.1 Neck strength measurements

A group of twelve test subjects (6F, 6M), where one female had a history of back- and neck pain participated voluntarily in the experiment. Each subject was seated in an upright neutral position in a Volvo car seat without head rest and an angle close to 90° between the seat and the backrest to provide support for the upper part of the body. An inelastic 25mm wide band was strapped around the test subjects head immediately above the eyebrows. A digital hand-held dynamometer with a hook in one end was attached to a hole in the inelastic band and used to determine the magnitude of the retraction resistance (N). The test subjects were requested to keep their head in the initial neutral position even when a horizontal backward force via the dynamometer was applied. They were not allowed to move their heads forward to help resisting the applied force. The maximal retraction resistance was defined as the force measured just before the subject moved its head backwards and no longer could resist the force, or when the subject said stop because he or she experienced discomfort. All subjects were given a second trial if they wished, and if that was the case the maximum value of the two trials was chosen as maximum retraction resistance. Mean value and one standard deviation was calculated for men and women separately and compared with results from neck strength maximum voluntary contractions for flexion in sitting position presented in the literature review.

3.1.2 Surface EMG measurements

Electromyographic activity in the right sternocleidomastoid (SCM) muscle was recorded using pairs of disposable surface electrodes. The same group of volunteers as in the neck strength measurements was used. Also the same seat was used but with a larger angle between seat and backrest (approximately 110°) to mimic the seat position in sled tests with the BioRID dummy. The test subjects were asked to turn their head to the left in order to help the positioning of the electrodes by making the SCM more apparent. Two electrodes were placed beside each other perpendicular to the direction of the muscle fibres and over the right SCM muscle (see Fig. 5). A ground electrode was placed over the vertebral column approximately at T1 level. The three electrodes were connected to a MyoTrac EMG biofeedback system. The test subjects pushed their forehead against their hand palm to confirm that the electrode position was correct and that the signal was strong and readable. The MyoTrac was connected to a computer recording system and the EMG signal was sampled with a sampling frequency of 10 kHz. A high speed video camera was used. The frame rate was 1000 frames per second and the camera recorded the first 600ms after start-up. The test subjects were instructed to adopt a comfortable seated posture with their back against the backrest, face forward and relax their face and neck muscles. A non reflective surface was in front of them making it impossible to see what was behind them. The subjects were exposed to the sound from a balloon exploding about 1 meter behind them. The sound from the balloon triggered a sound-trigger system which started the high speed video camera and the recording of the EMG signal. The signal was processed and filtered with Channel Frequency Class 50 (CFC50). Onset time for the EMG was determined. Onset was relative to the explosion of the balloon and was defined as the time in which a distinct increase of the EMG signal occurred. Data acquisition was restricted to the first 1000ms after the start-up triggered by the balloon. Two trials were performed. Since there was a difference between trials the values for both trials were used. Mean value and one standard deviation was calculated.

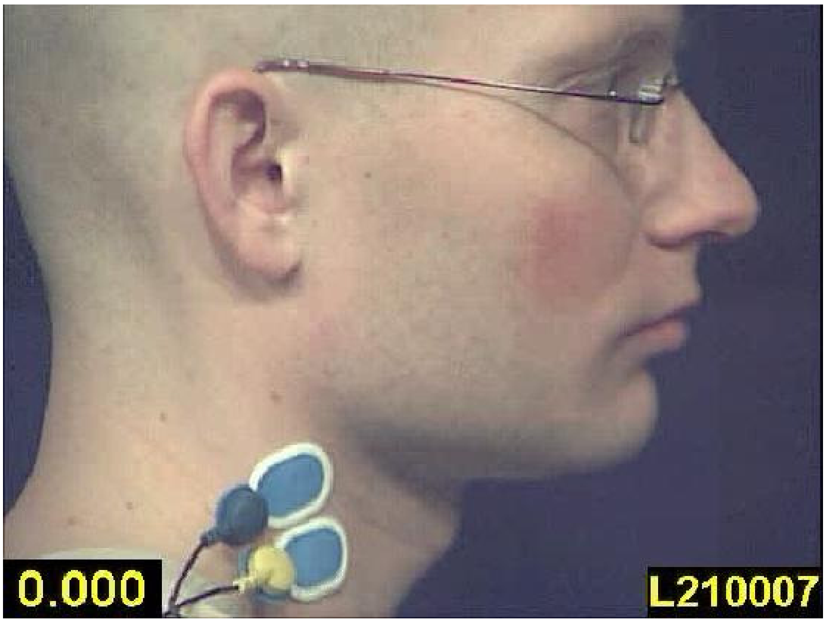

Figure 5 Position of the surface EMG electrodes over the right SCM.

3.2 Muscle substitute

A muscle substitute for the BioRID II dummy was designed with the purpose to prevent or decrease the retraction motion of the dummy´s head relative T1 in a rear impact. Aluminum was used as material because of its low weight properties. The muscle substitute can be divided into three parts: (i) the head part following the motion of the head (ii) the T1 part following the movement of the lower neck and upper torso and (iii) the part connecting (i) and (ii). The design can be seen in Figure 6. Part (i) has threads in one end and was kept in place with a screw nut inside the head of the dummy. Part (ii) was attached to the two uppermost stainless steel torsion pins with which the vertebrae are coupled together, and fixed with a screw nut on each side of the neck. Inside the cylinder (part iii) was a stainless steel extension spring. The spring had an initial length of L_0=142mm and an initial force, i.e. the force required before the spring started to extend, of F_0=57 N. The spring constant was c=2.31N/mm. The muscle substitute was designed to correspond to a force just below the average value for men in the retraction resistance experiment. The spring properties are shown graphically in Figure 7. The cylinder was used to extend the spring by 36mm from its initial length to start at F_{start}=140N, i.e. a force of more than 140N was required to extend the spring further. The loops in the ends of the spring was threaded through slots in circular stops and locked by u-shaped metal bars. This resulted in a robust construction and the distance between parts (i) and (ii) was always 188mm. The total weight of the muscle substitute was about 0.7kg.

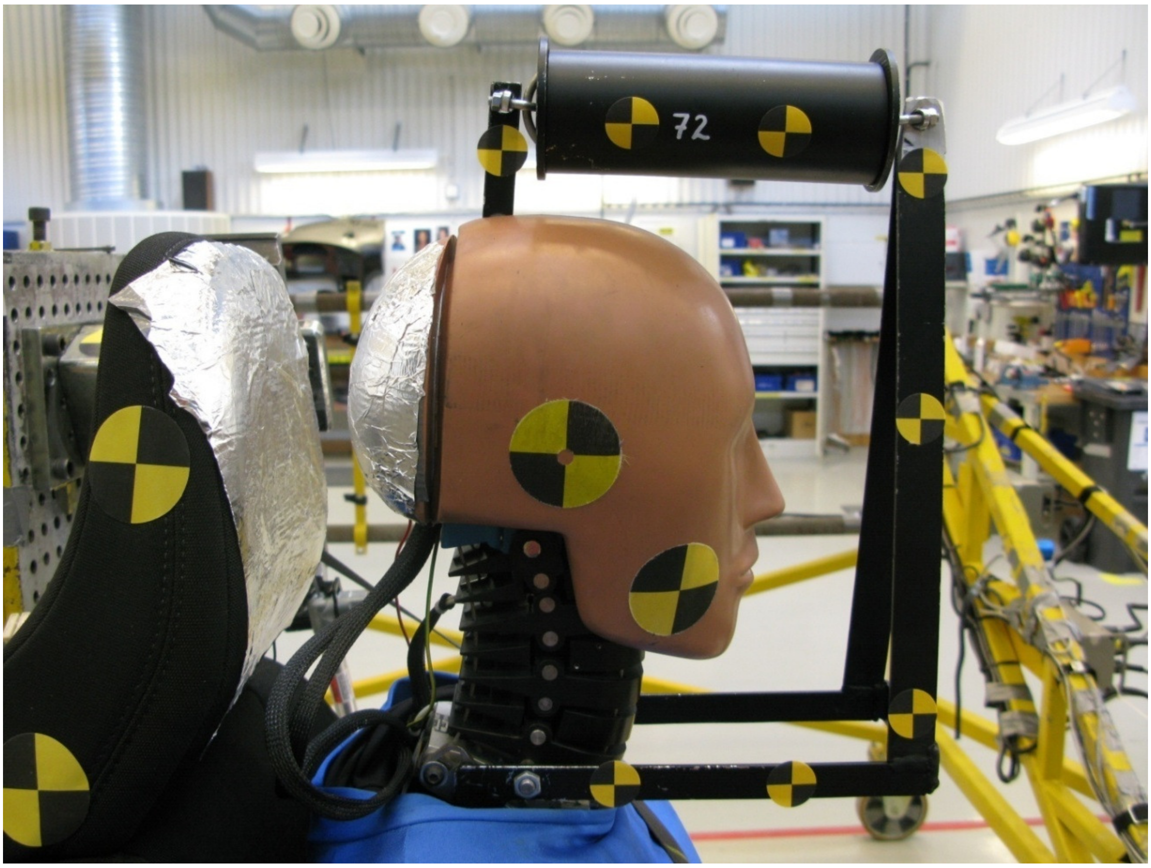

Figure 6 BioRID II dummy with muscle substitute.

Figure 7 The initial force at the initial length is F_0=57N. The spring was extended by 36mm from its initial length to start at F_{start}=140N i.e. a force of more than 140N was required to extend the spring further. This point is where the vertical and horizontal lines meet in the graph. The lower cross shows the average retraction resistance for women and the upper cross represent the average retraction resistance for men.

3.3 Rear impact sled tests

A series of rear impact sled tests with the BioRID II dummy in EuroNCAP low- and medium pulse was performed at Autoliv, Vårgårda. Both pulses resulted in a change of velocity Δv=16km/h. The shape of the pulses can be seen in Figure 1. The dummy was seated in front passenger position (hands in lap) restrained by a lap and shoulder belt. The positioning was performed by specialists from Autoliv AB according to ISO (International Organization for Standardization) and EuroNCAP protocols. Five different seat designs with and without whiplash injury protection system were compared with and without muscle influence.

- Volvo welded: a Volvo seat with welded WHIPS and the WHIPS no longer active.
- Volvo P2x welded: an older Volvo seat model where the WHIPS was welded and no longer active.
- Volvo P2x: seat with active WHIPS.
- Hybrid Toyota-Volvo: a Volvo seat with WHIPS and a Toyota backrest and head rest with WIL.
- Hybrid Toyota-Volvo, Volvo head rest: a Volvo seat with WHIPS, a Toyota backrest with WIL and a Volvo head rest.

The seats where mounted on a still standing target sled that was impacted from the rear. Two high speed video cameras mounted on the sled was used. One monitored an overall view and the other monitored a view of the dummy´s head and the head rest. The different signals were filtered and processed, and 7 seat performance criteria were calculated: NIC; N_{km}; T1 acceleration in x-direction; head to head rest contact time; Upper neck force in x- and z-direction and rebound velocity. The filtered signals were plotted in MATLAB and Excel. NIC was recalculated to injury risk (%). Also head acceleration in x-direction was compared between tests with and without the muscle substitute, and for different seat design.

4 Results

4.1 Retraction resistance measurements and EMG recordings

4.1.1 Neck strength measurements

Retraction resistance expressed in N was summarized for men and for women. Mean value and one standard deviation was calculated. The value for the woman with a history of neck- and back pain is included in the calculations. A comparison of the results from this current study and maximum isometric strength in flexion from literature (Vasavada, 2008) can be seen in Figure 8. The mean value (SD) for the females when the woman with history of neck pain was excluded was 92(27)N.

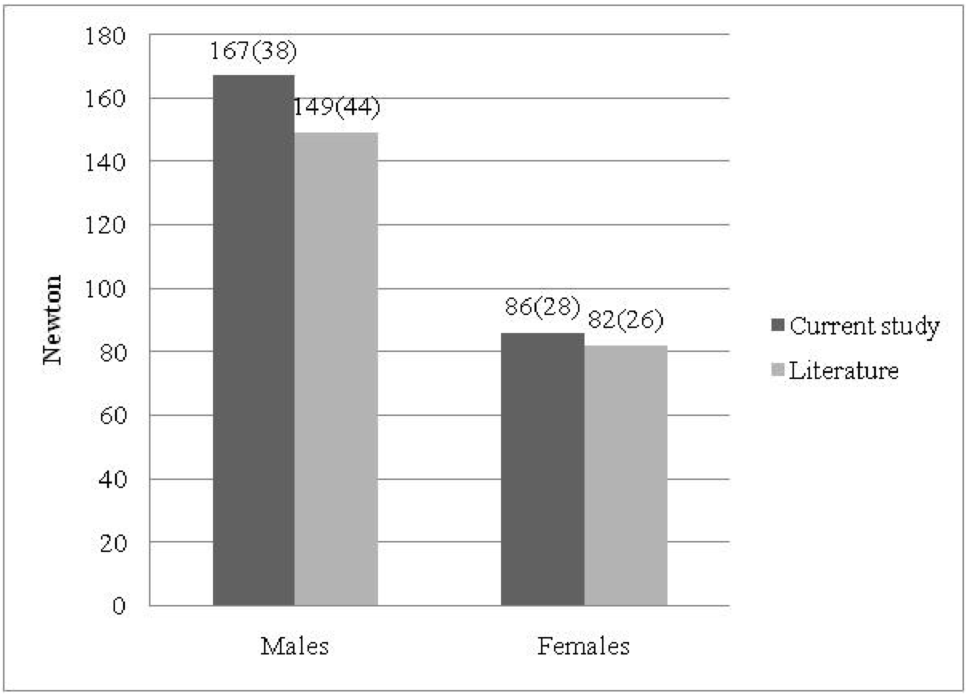

Figure 8 Comparison of retraction resistance results from current study and maximum isometric strength in flexion from literature (Vasavada, 2008). Mean values (SD) expressed in Newton.

4.1.2 Surface EMG measurements

A summary of the right SCM muscle onset can be seen in Table 1.

Table 1 Summary of muscle (right SCM) reaction time from balloon pop for two trials. Male (M) Female (F)

Test person	Trial	Activation time [ms]
1 (M)	1	74
	2	63
2 (M)	1	61
	2	68
3 (M)	1	85
	2	74
4 (M)	1	85
	2	125
5 (F)	1	65
	2	66
6 (F)	1	18
	2	20
7 (F)	1	75
	2	63
8 (F)	1	79
	2	76
9 (F)	1	87
	2	91
10 (M)	1	66
	2	70
11 (F)	1	73
	2	98
12 (M)	1	86
	2	162

Muscle activation time was defined as the time from the explosion of the balloon (t=0) to the time where a distinct rising of the EMG output signal started. This can be seen at t=0.066 s in Figure 9.

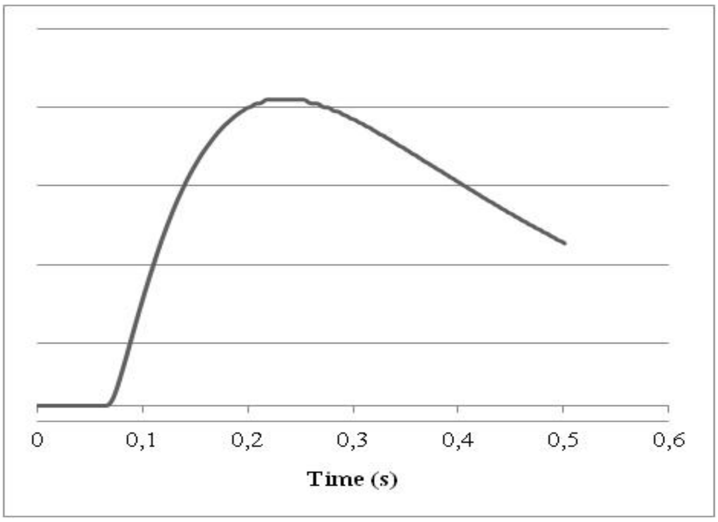

Figure 9 Example of MyoTrac output signal after filtering with CFC 50.

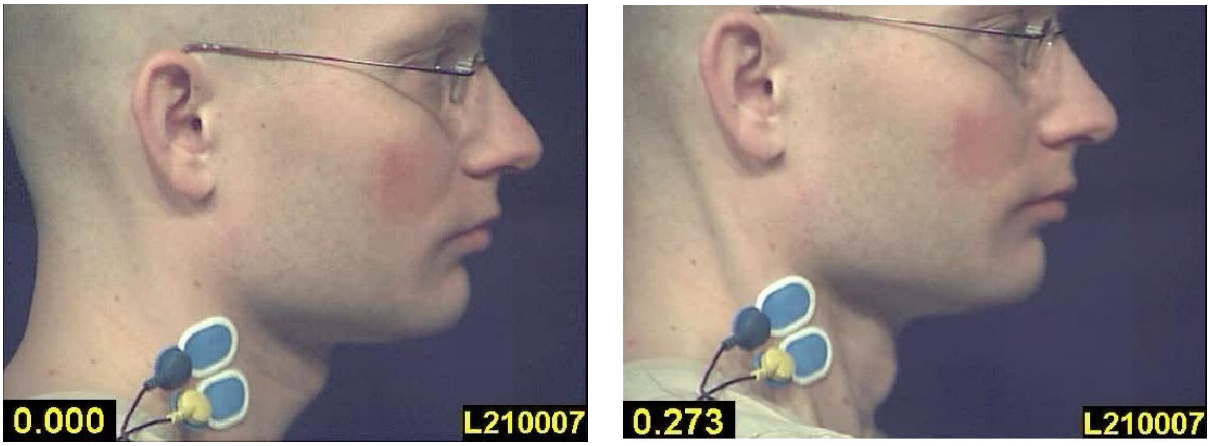

Figure 10 Comparison of SCM relaxed (left) and contracted (right). The time in the lower left corner is the time in seconds.

A group of 12 test subjects and two recordings per subjects resulted in 24 values (Table 1). Four values differed a lot from the rest (18; 20; 125; 162 ms). These four values were removed and a normal probability plot was done for the remaining twenty values to determine whether the data set is approximately normally distributed or not. The result is shown in Figure 11.

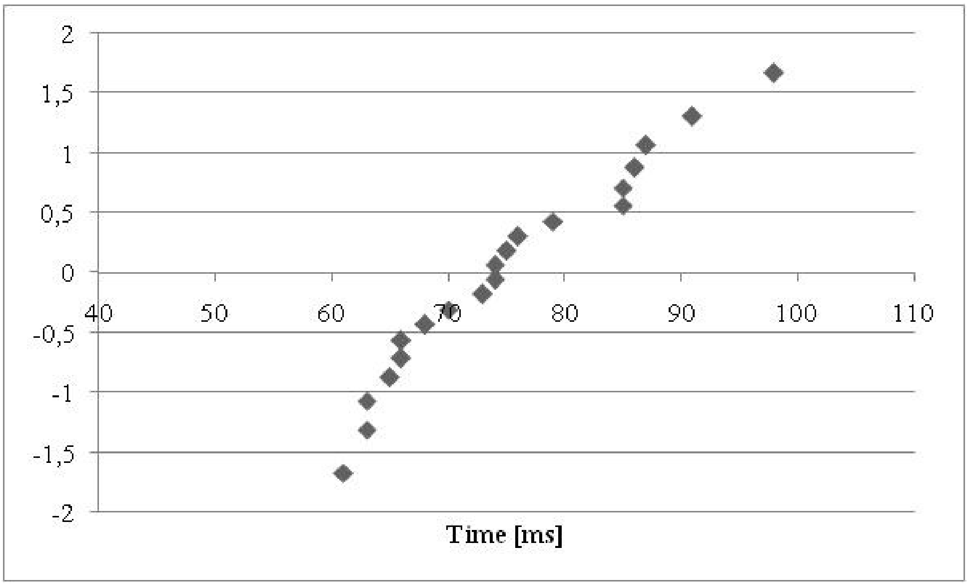

Figure 11 Normal probability plot for muscle onset times.

The probability density function can be seen in Figure 12. Mean value for the data set and one standard deviation has been calculated. Mean value was 75.3±10.5ms.

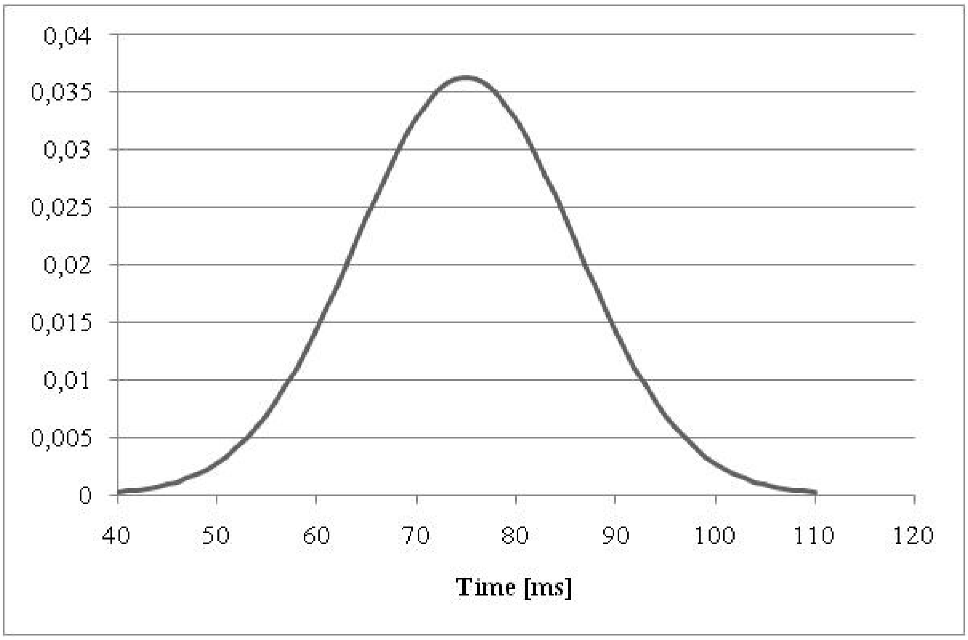

Figure 12 The probability density function for muscle activation time. Mean value (SD) = 75(11) ms.

4.2 Rear impact sled tests

The difference in reclining motion between a Volvo welded seat with active WHIPS and a Volvo welded seat can be seen in Figure 13. The marks on the yellow part of the sled just behind the seat can be used as reference points.

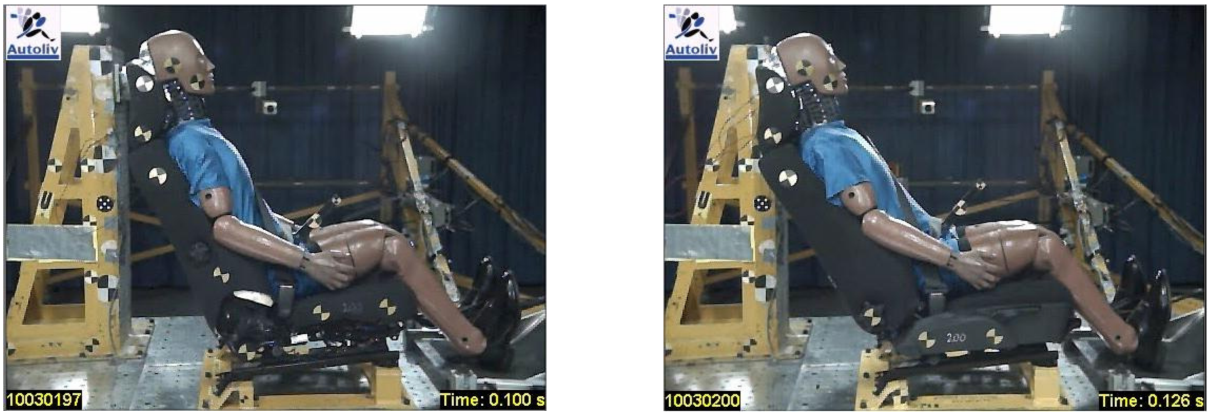

Figure 13 Difference in motion of a welded Volvo seat (left) and a Volvo seat with active WHIPS (right).

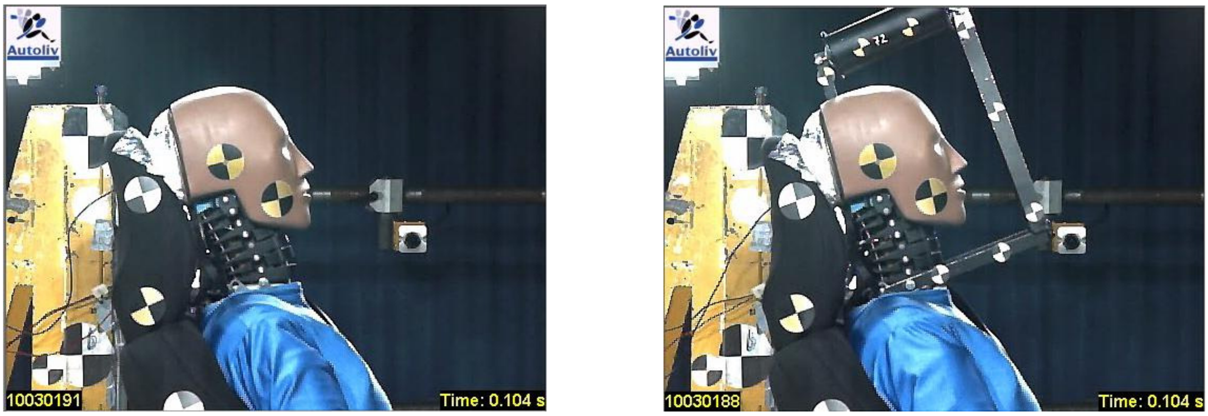

Figure 14 Difference in motion of the head and neck without (left) and with muscle substitute (right) in a welded Volvo seat.

The difference in relative motion between the head and the torso without muscle influence compared to with the muscle substitute can be seen in Figure 14.

The seat performance criteria result from the sled tests for the different seat types can be seen in Table 2 (medium pulse) and table 3 (low pulse).

Table 2 Result from sled test for the various seat types, medium pulse. Performance limit range is the interval from high performance to low performance according to EuroNCAP. The performance limit range interval for "Head contact time" is the interval for head to head rest contact time.

Seat type	Muscle	NIC max $[m^2/s^2]$	Nkm max	T1 X max acc, [g]	Head contact time [ms]	Upper Neck Force X [N]	Upper Neck Force Z [N]	Head rebound velocity [m/s]
Volvo welded	0	16.7	0.26	12.0	61-149	58.5	639.3	5.6623
Volvo welded	X	12.5	0.20	11.7	71-163	16.8	656.8	4.8303
Volvo P2x welded	0	16.2	0.15	12.1	56-147	5.7	649.0	5.1574
Volvo P2x welded	X	13.9	0.27	10.8	58-151	23.6	913.3	4.8356
Volvo P2x	0	12.3	0.25	7.4	83-189	18.1	323.3	3.2032
Volvo P2x	X	13.4	0.35	7.8	85-198	34.0	538.6	3.0999
Hybrid Toyota/Volvo	0	15.2	0.17	7.0	87-189	57.2	373.6	3.4366
Hybrid Toyota/Volvo	X	11.2	0.28	7.0	93-196	53.7	449.7	3.4709
Hybrid, Volvo HR	0	11.6	0.12	8.1	58-184	0.2	175.4	2.9429
Hybrid, Volvo HR	X	11.5	0.39	7.6	81-188	12.6	335.5	3,1499
Performance limit range middle pulse		11-24	0.15-0.55	9.3-13.10	57-82	30-190	360-750	3.2-4.8

Table 3 Result from sled test for the various seat types, low pulse. Performance limit range is the interval from high performance to low performance according to EuroNCAP. The performance limit range interval for "Head contact time" is the interval for head to head rest contact time.

Seat type	Muscle influence	NIC max $[m^2/s^2]$	Nkm max	T1 X max acc, [g]	Head contact time [ms]	Upper Neck Force X [N]	Upper Neck Force Z [N]
Volvo P2x welded	0	8.7	0.29	10.1	60-166	8.2	292.6
Volvo P2x welded	Male	9.2	0.23	10.1	64-168	13.4	771.7
Performance limit range low pulse		9-15	0.12-0.35	9.4-12.0	61-83	30-110	270-610

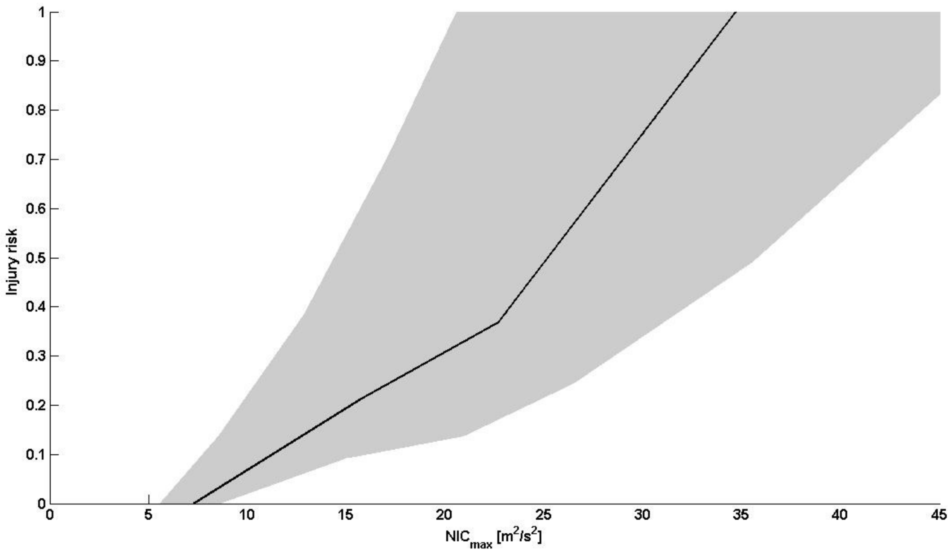

Figure 15 Long-term neck injury risk versus $NIC_{max.}$ (Eriksson, 2004). The injury risk curve (black line) is established from the relation between NIC_{max} values and the duration of AIS 1 neck injury outcome. The spread of the risk (grey area) is based upon the variation in seat geometry and seating posture.

Based on the injury risk curve established by Eriksson and Kullgren a similar curve (Fig. 15) was plotted in MATLAB to determine the injury risk for the NIC_{max}. The result is summarised in Table 4 and in Figure 16.

Table 4 Risk reduction (%) with muscle substitute calculated from the change in injury risk.

Seat type	Muscle influence	NIC max [m^2/s^2]	Injury risk	Risk reduction [%]
Volvo welded	0	16.7	0.234	- 44
Volvo welded	Male	12.5	0.131	
Volvo P2x welded	0	16.2	0.223	- 24.5
Volvo P2x welded	Male	13.9	0.166	
Volvo P2x	0	12.3	0.126	+22.2
Volvo P2x	Male	13.4	0.154	
Hybrid Toyota/Volvo	0	15.2	0.199	- 50.75
Hybrid Toyota/Volvo	Male	11.2	0.098	
Hybrid Toyota/Volvo, Volvo HR	0	11.6	0.108	+1.85
Hybrid Toyota/Volvo, Volvo HR	Male	11.5	0.106	

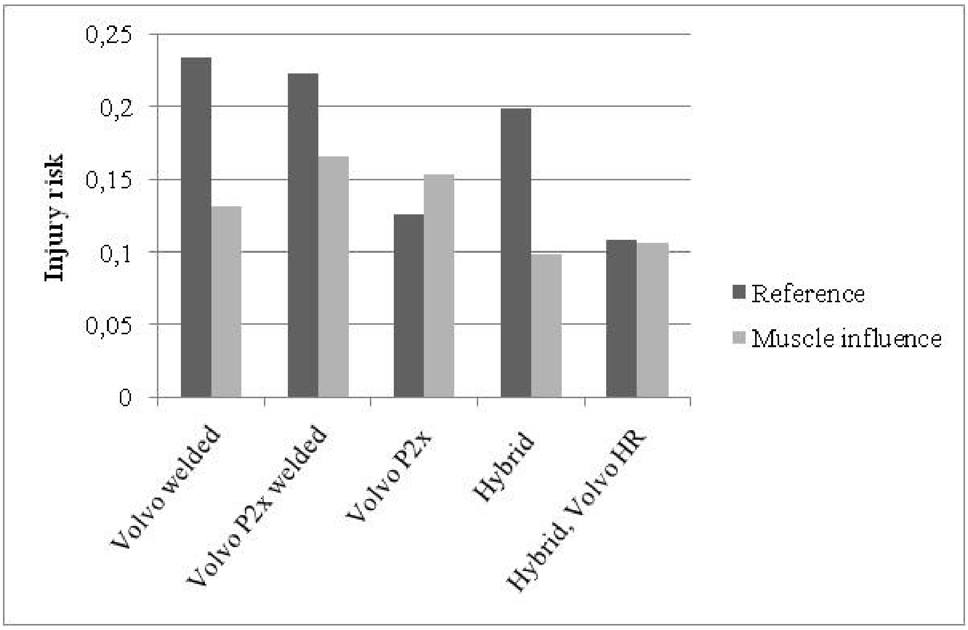

Figure 16 The injury risk based on Figure 10 for reference test versus muscle substitute test for each of the five seat types.

Comparison of EuroNCAP seat performance criteria between tests without (reference) and with muscle substitute can be seen in Figure 17-21. Results for Upper neck force in x- and z-direction can be found in Appendix.

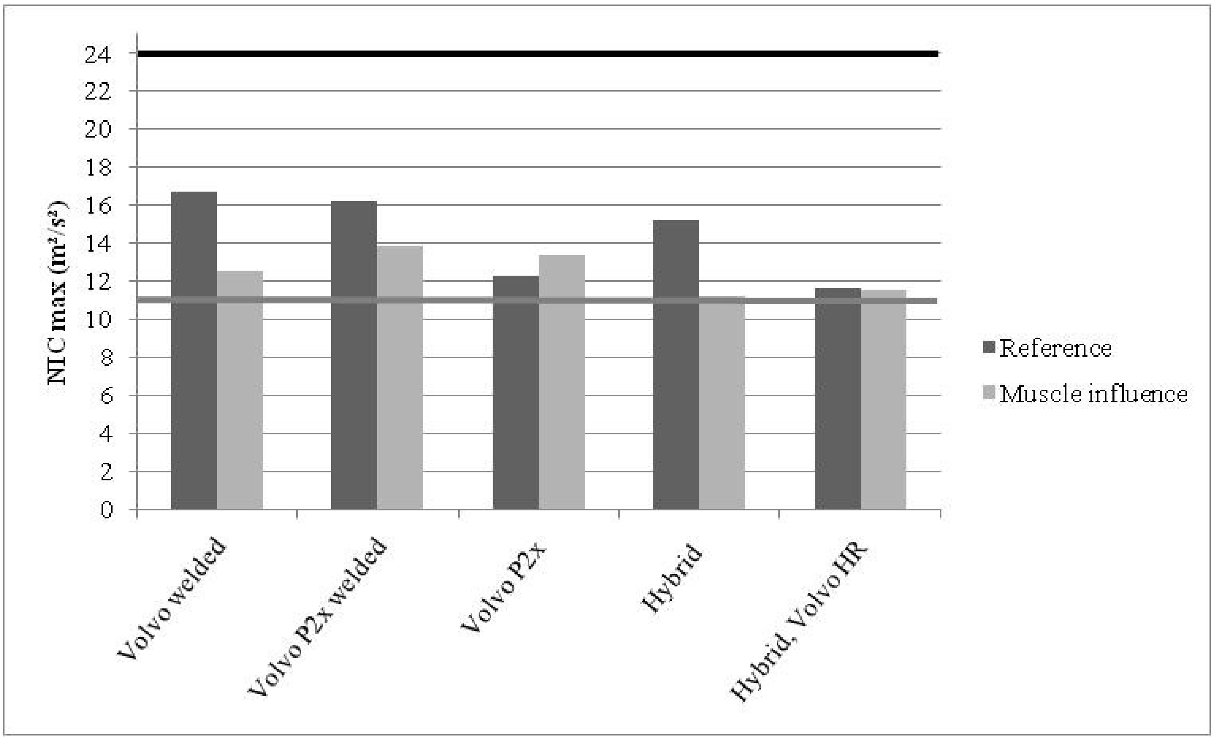

Figure 17 NIC_{max} comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

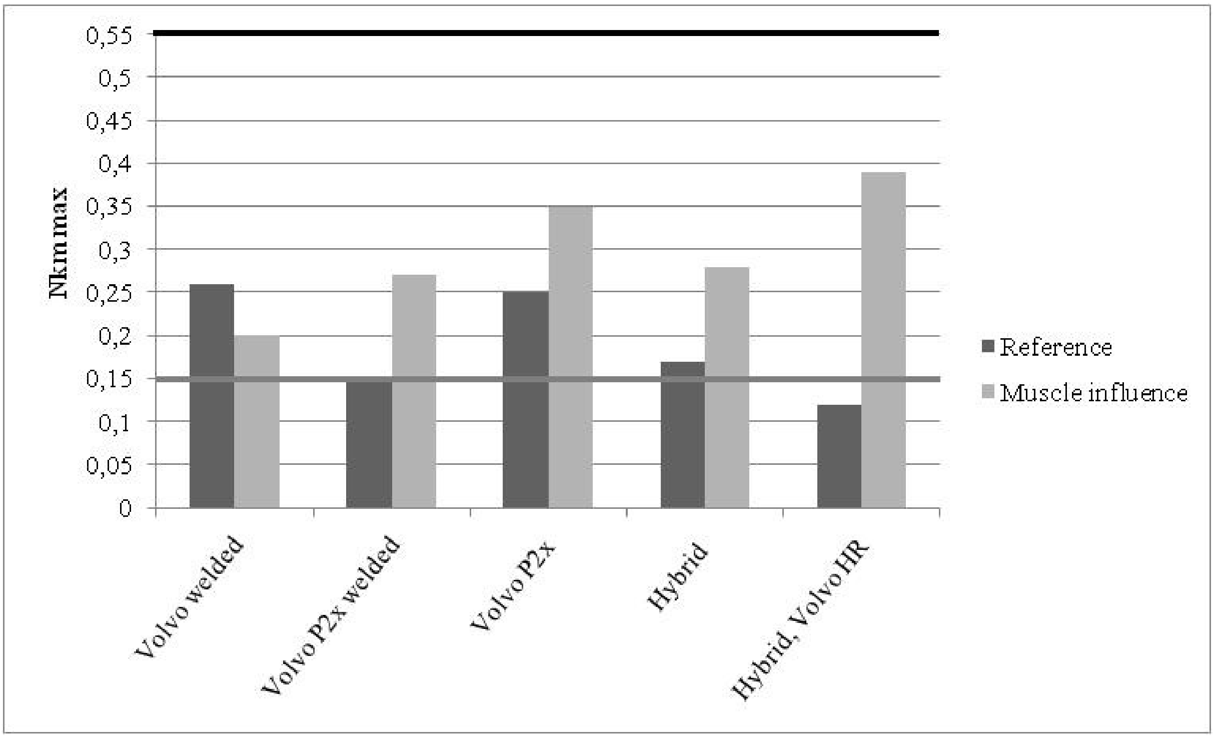

Figure 18 Nkm_{max} comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

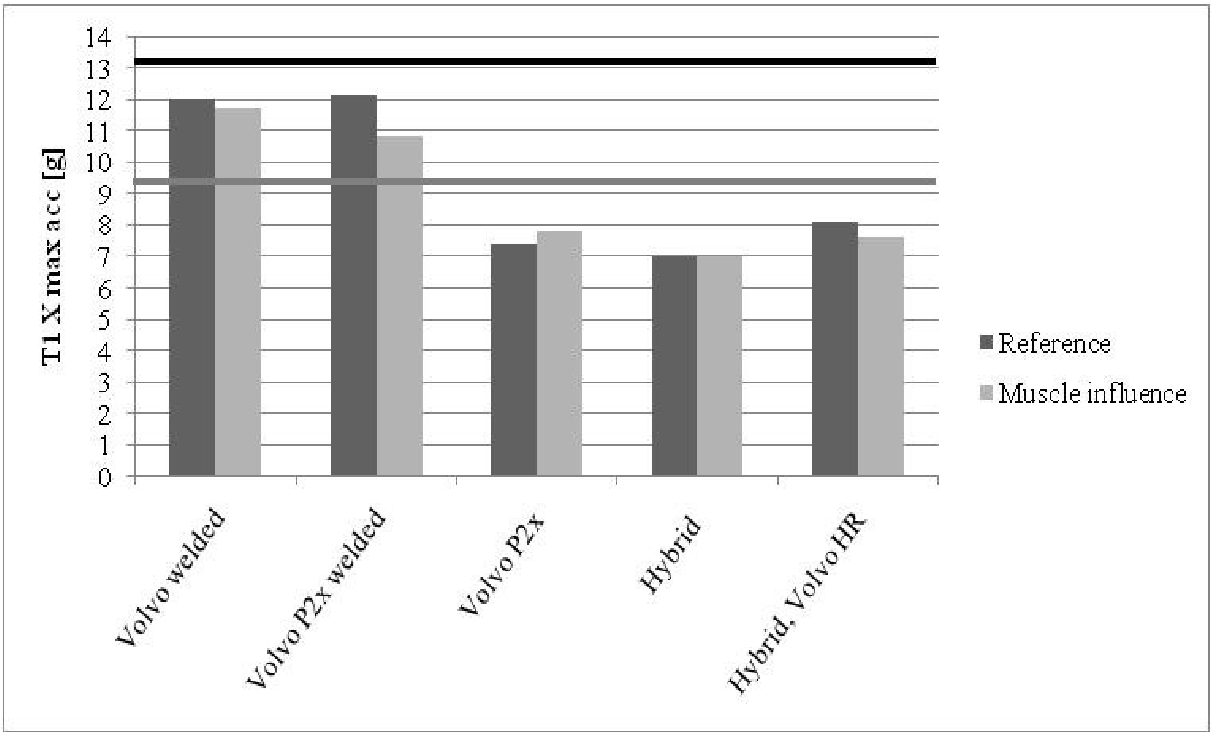

Figure 19 $T1_{max}$ x acceleration comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

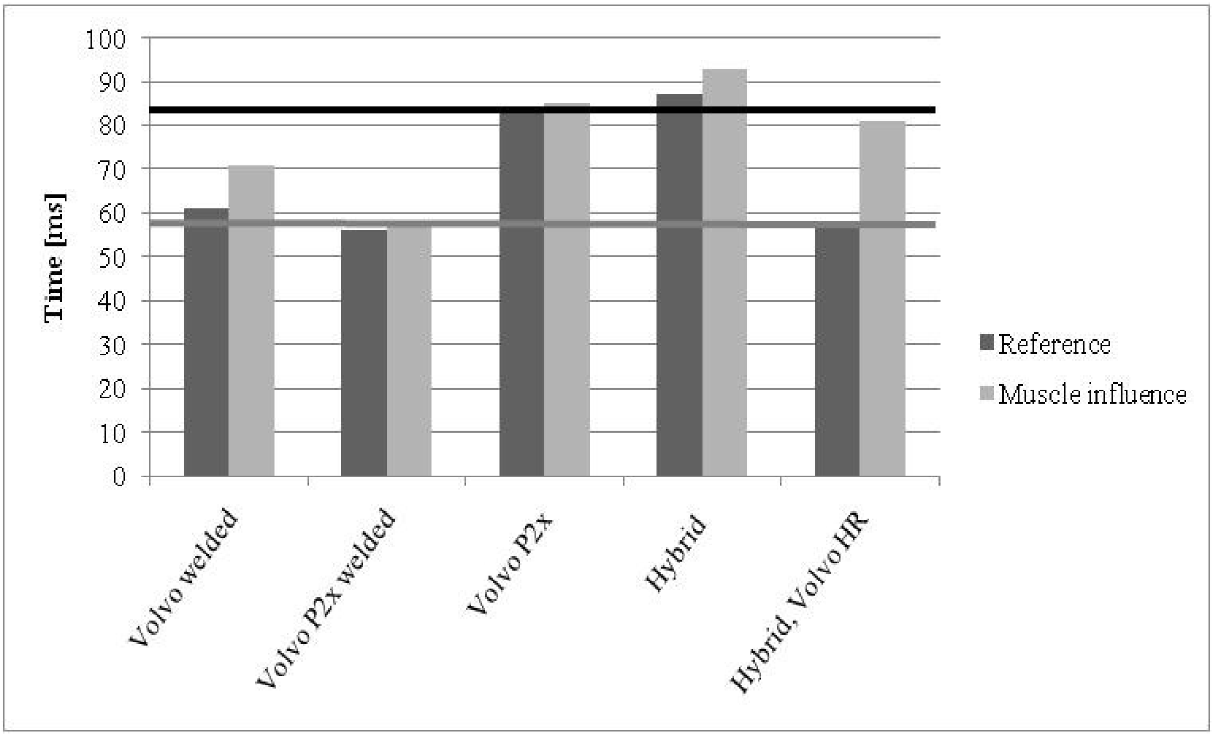

Figure 20 Time to head rest contact comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

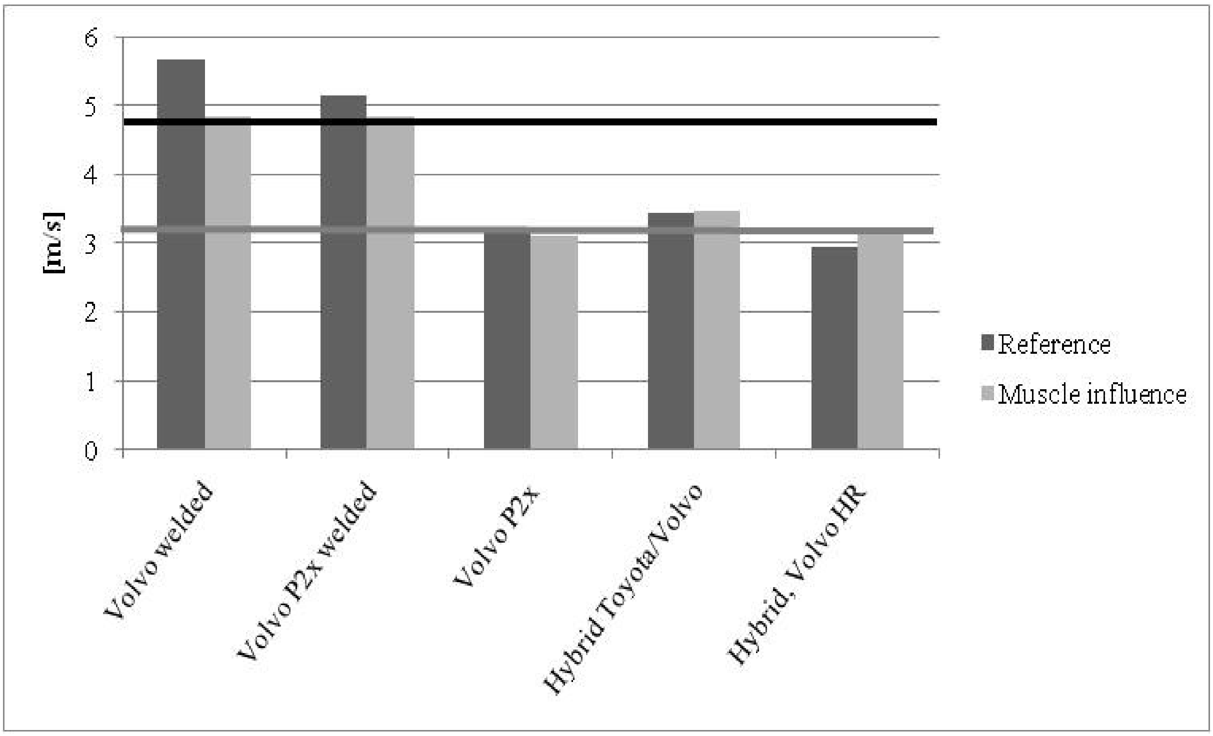

Figure 21 Head rebound velocity comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

Comparison of NIC curves between tests with and without muscle substitute can be seen in Figure 22-26.

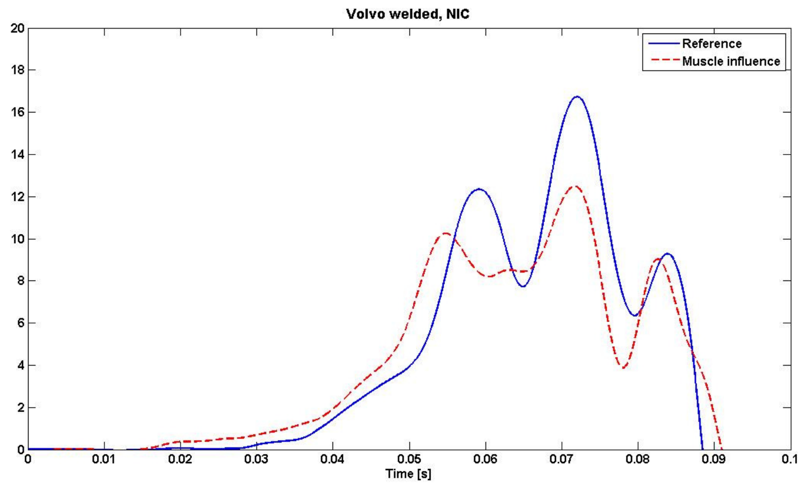

Figure 22 Volvo welded seat. Comparison of NIC during the first 100ms after impact for reference test and muscle substitute test.

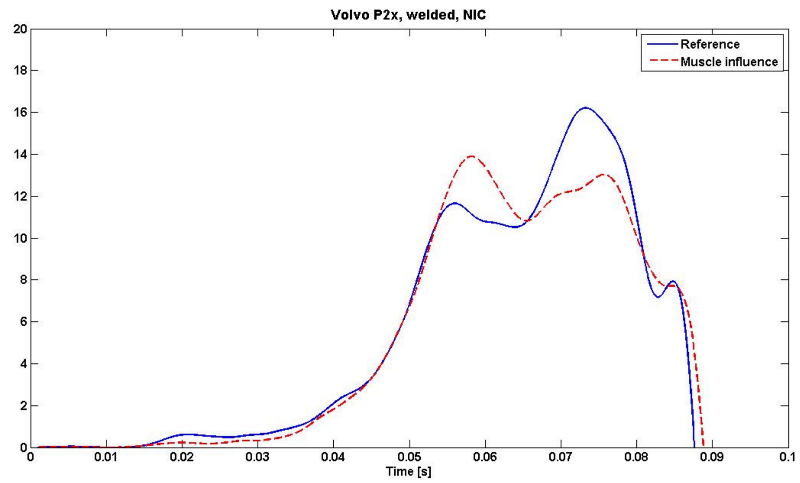

Figure 23 Volvo P2x welded seat. Comparison of NIC during the first 100ms after impact for reference test and muscle substitute test.

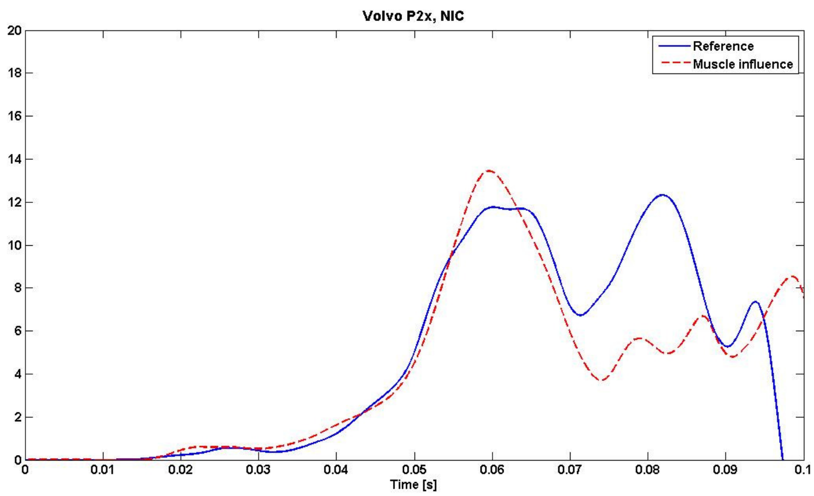

Figure 24 Volvo P2x seat with WHIPS. Comparison of NIC during the first 100ms after impact for reference test and muscle substitute test.

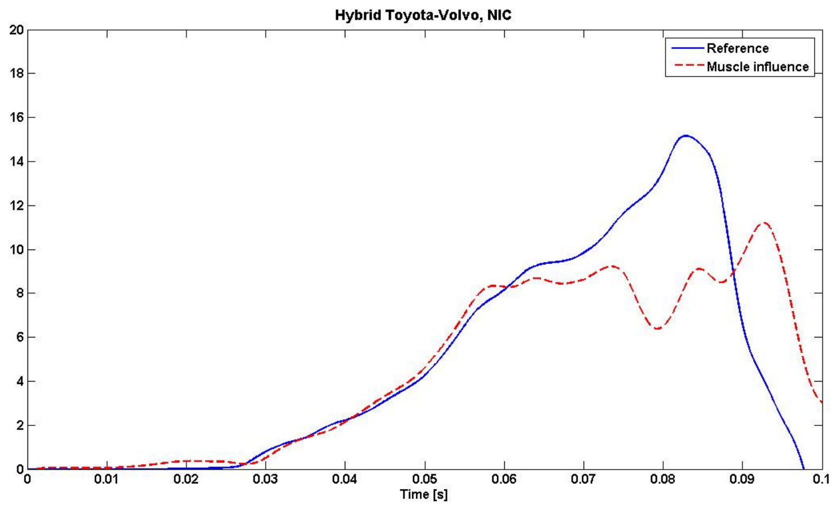

Figure 25 Hybrid Toyota-Volvo seat. Comparison of NIC during the first 100ms after impact for reference test and muscle substitute test.

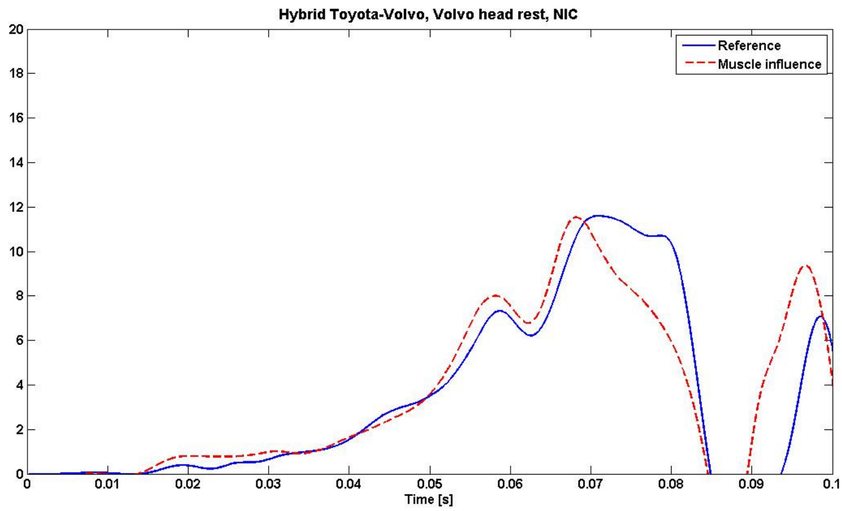

Figure 26 Hybrid Toyota-Volvo seat with Volvo head rest. Comparison of NIC during the first 100ms after impact for reference test and muscle substitute test.

Comparison between the head acceleration in x-direction for tests with and without muscle substitute can be seen in Figure 27-31.

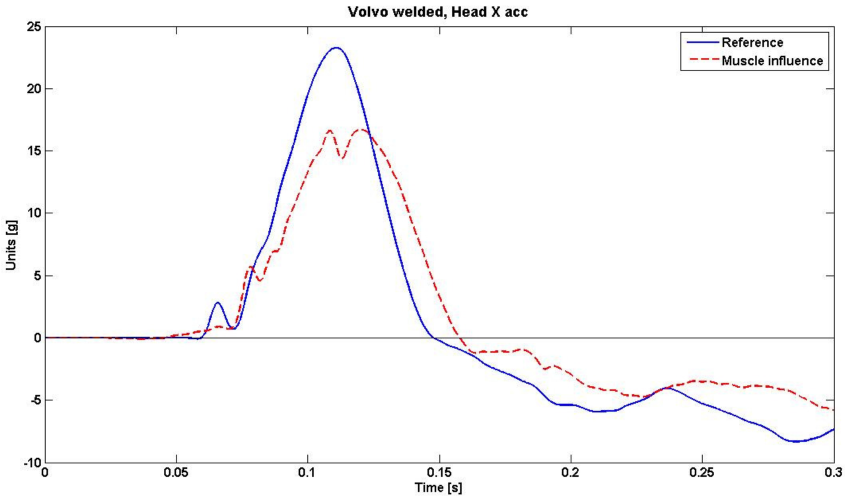

Figure 27 Volvo welded seat. Comparison of head acceleration in x-direction for reference test and muscle substitute test.

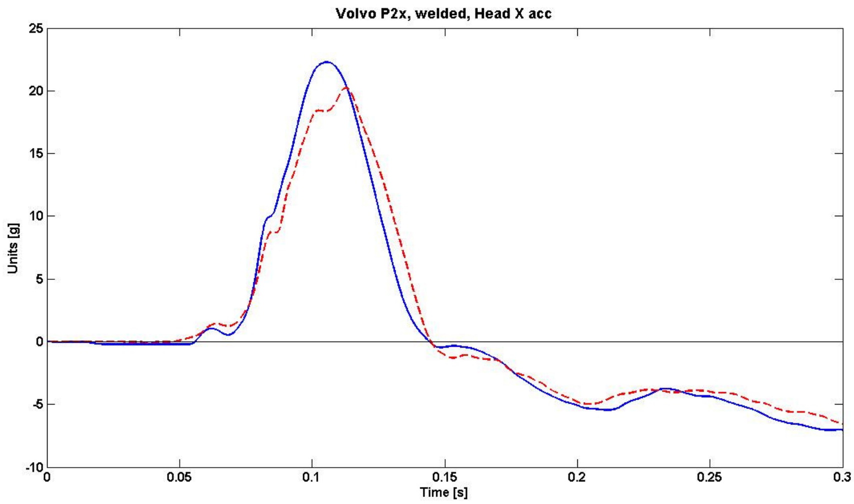

Figure 28 Volvo P2x welded seat. Comparison of head acceleration in x-direction for reference test and muscle substitute test.

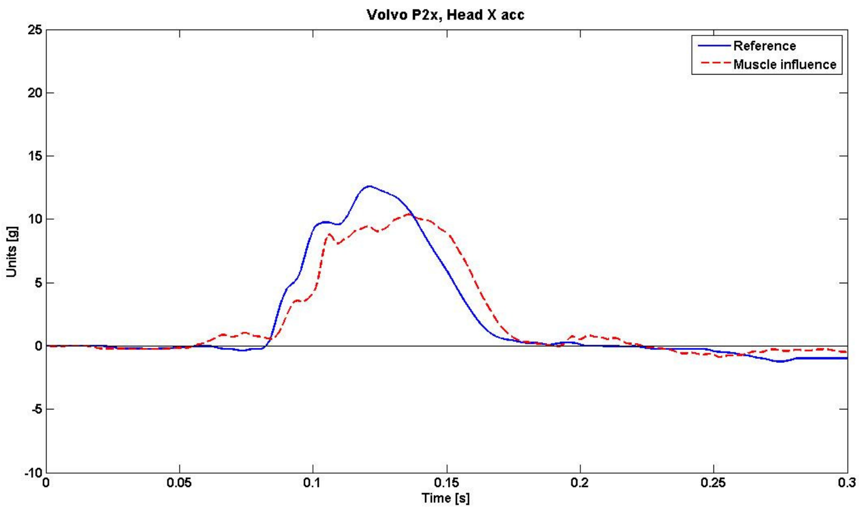

Figure 29 Volvo P2x seat with WHIPS. Comparison of head acceleration in x-direction for reference test and muscle substitute test.

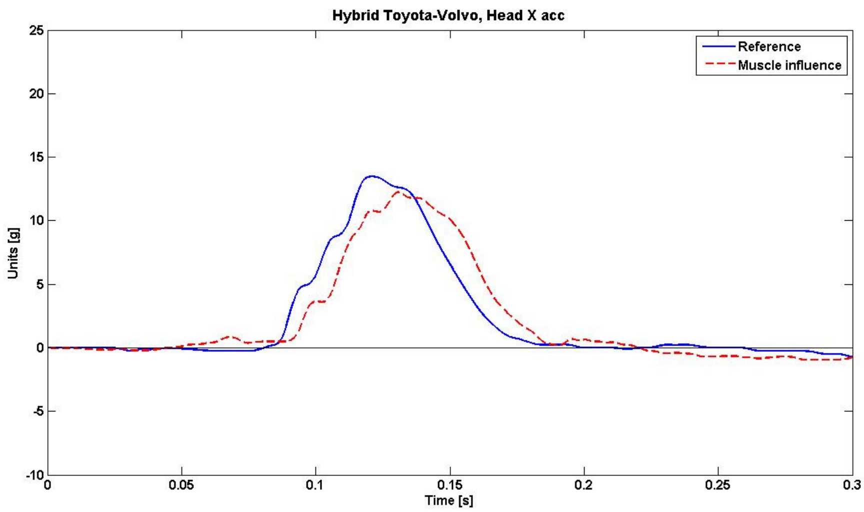

Figure 30 Hybrid Toyota-Volvo seat. Comparison of head acceleration in x-direction for reference test and muscle substitute test.

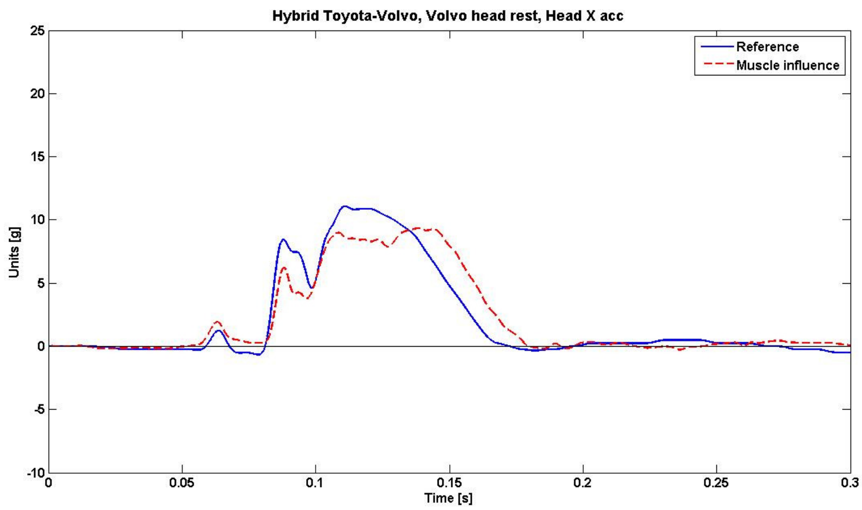

Figure 31 Hybrid Toyota-Volvo seat with Volvo head rest. Comparison of head acceleration in x-direction for reference test and muscle substitute test.

The difference in EuroNCAP score for NIC, T1x-acceleration, head-to-head rest contact time and head rebound velocity between reference tests and tests with muscle influence are shown in Figure 32-35. The total EuroNCAP score for the various seat designs (reference) and medium pulse are summarized in Figure 36. The highest score from T1 acceleration and head-to-head rest contact time results have been chosen in the summary of the score. N_{km} curves for the different seat types, comparison of Upper neck force in x-direction and Upper neck force in z-direction for reference tests and tests with muscle influence for the five seat designs, and EuroNCAP score for N_{km}, Upper neck force in x-direction and Upper neck force in z-direction are shown in Appendix.

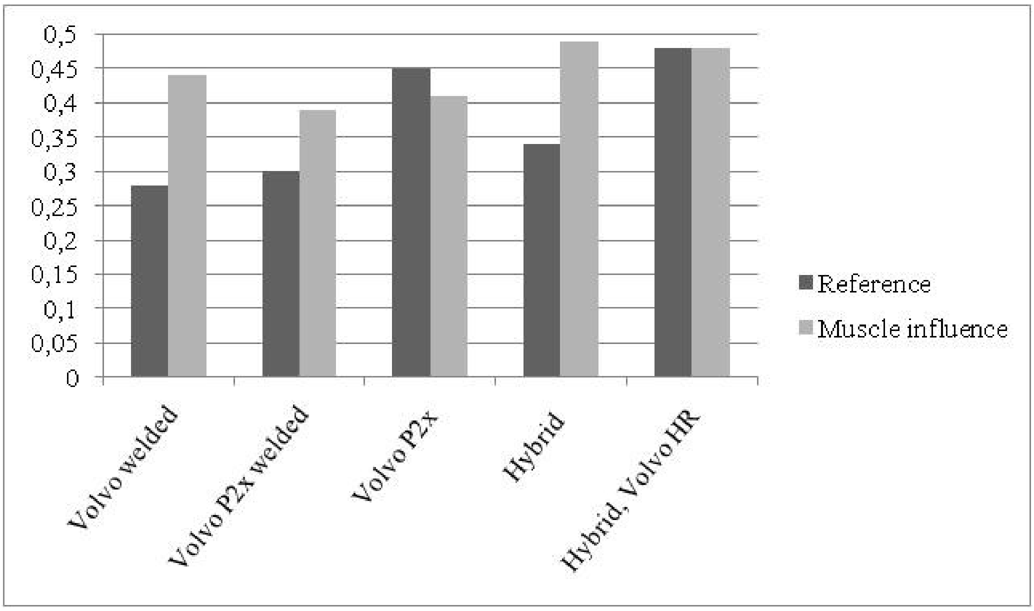

Figure 32 EuroNCAP score for NIC.

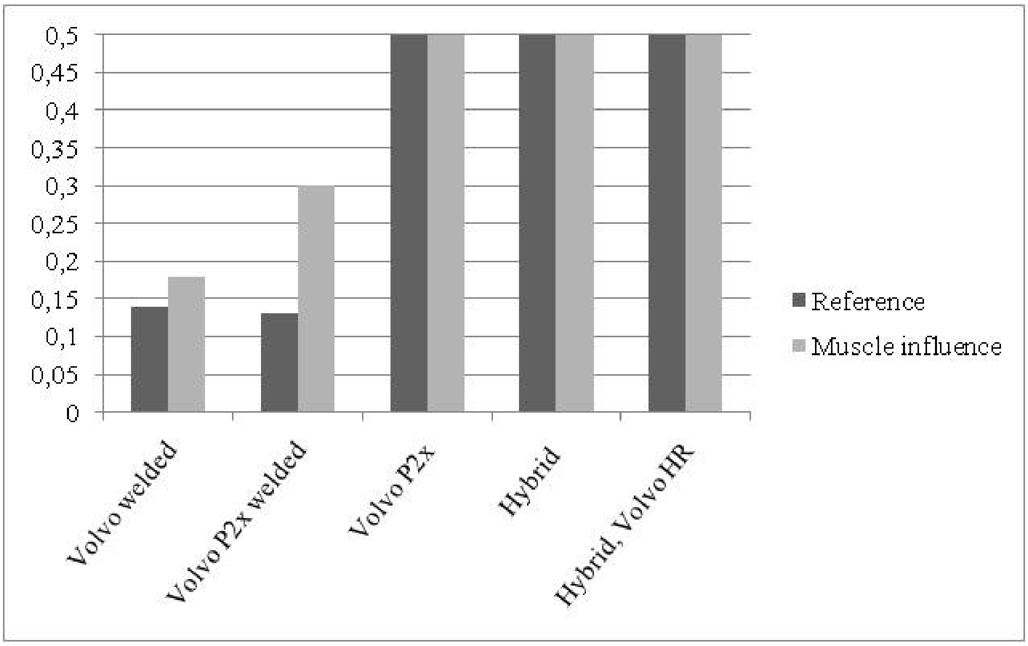

Figure 33 EuroNCAP score for T1 x acceleration.

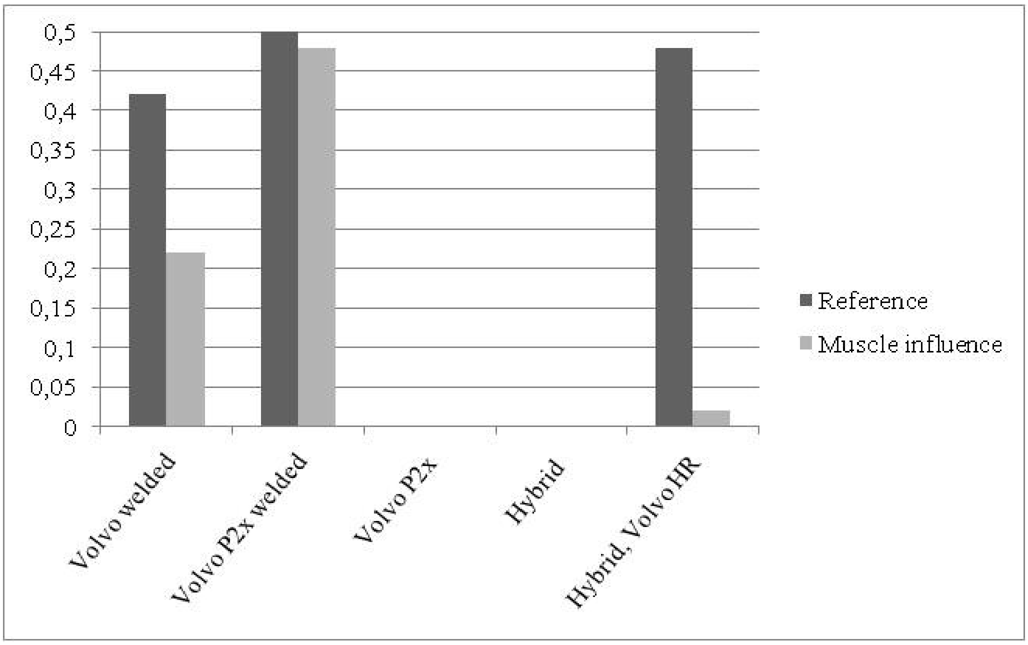

Figure 34 EuroNCAP score for head-to-head rest contact time.

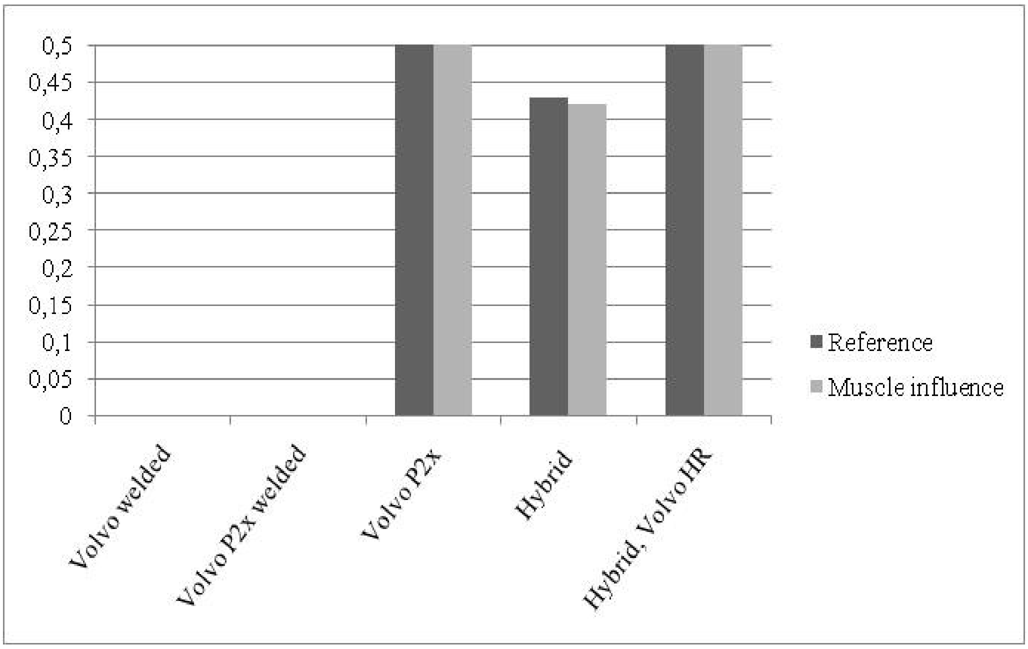

Figure 35 EuroNCAP score for head rebound velocity.

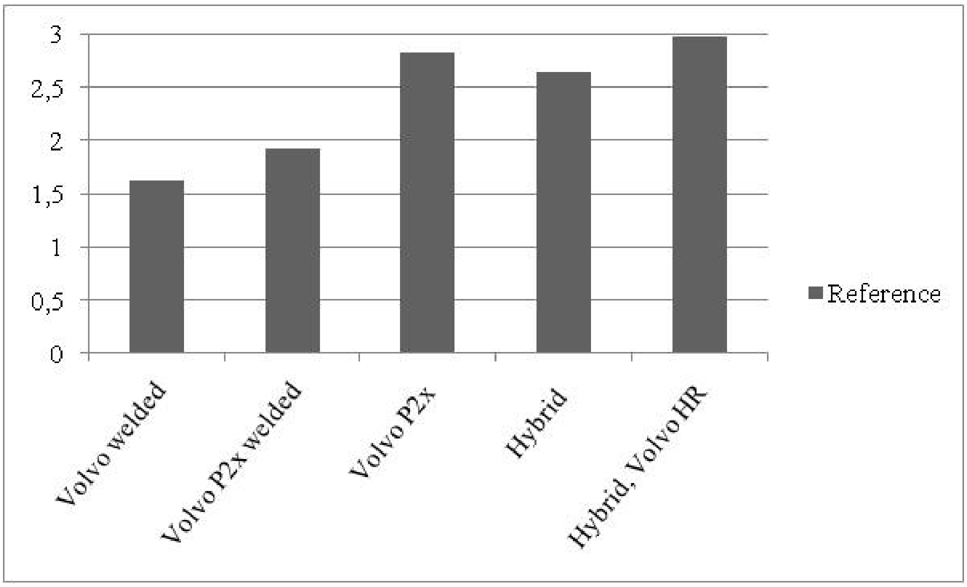

Figure 36 Total EuroNCAP score the different seats in tests without muscle influence.

5 Discussion

5.1 Retraction resistance measurements

In the retraction resistance study I found that men are about two times stronger than women in maintaining a normal head position when a backward force was applied. This result corresponds well with previous work (Cagnie et al., 2007, Vasavada et al. 2008, Larochelle et al. 2009) although I have had a different approach in measuring the strength. In the previous studies, presented in my literature review, somewhat more advanced equipment was used to measure neck strength in extension and flexion where the test subjects pushed against a load cell in different directions. In my studies some of the test subjects reported that they felt discomfort when the backward force increased and said stop even though they probably could have resisted a greater force. Another parameter that affects my results is that it was difficult to ensure that the test subjects did not move their head slightly forward to help resist the force. It is complicated to define neck flexion strength and measurement of maximum strength assumes that the test subjects are applying maximum effort. This is difficult or almost not possible to verify, although my recorded neck strengths are comparable to or exceed previously reported values. The findings support the hypothesis and demonstrate that neck strength in women is different than in men.

5.2 Surface EMG measurements

EMG activity in the right SCM was recorded with the MyoTrac EMG biofeedback system and muscle onset was defined as the time interval from a sound trigger to a distinct increase of the EMG output signal. A group of 12 test subjects and two recordings per subjects resulted in 24 values. Mean value and one standard deviation for muscle onset was 76±28ms. Since four values differed a lot from the other these values were removed and the corrected muscle onset was 75±11ms. A normal probability plot was done which indicated that the 20 values were normally distributed. The recorded muscle reaction times are in the same interval as in previous work (Magnusson et al., 1999, Brault et al., 2000 Hernández et al., 2006) and indicate that the muscle contract was done reflexively rather than voluntarily. The definition of onset varies in the studies. I only focused on the onset and not on other parameters, e.g. amplitude and peak value, since these parameters varied a lot with test subjects. Some test subjects had a more well defined and superficial SCM muscle than others and it was easier to get a strong signal. In all the previous work I have studied the trigger, or applied stimuli, is the sudden velocity change of the sled that causes an acceleration of the body when it is hit by the backrest. In this study the trigger was auditory and people respond very different to a sound compared to a sudden acceleration of the body where the reaction pattern is more similar between individuals. Also, Scott et al. (2008) showed that the response time is shorter with tactile stimuli compared to auditory or visual.

5.3 Sled tests with and without muscle influence

In the series of sled test the results from trials with muscle substitute was compared to a reference where no muscle substitute was used. Crash test was done where the construction was on but the spring removed. Extra weight was added to compensate for the spring. In this trial the construction moved a lot and affected the motion of the dummy´s head in a non-representative way. Therefore a dummy with no extra equipment was used as reference. The added weight when the muscle substitute was used made the dummy a little heavy on the nose but the initial distance between the back of the dummy´s head to the head rest was always the same for the same seat design. The design of the muscle substitute is fairly simple. No further extension of the spring could be seen in the film analysis indicating that the relative motion between head and torso was small. A next step could be to fix the head with the lower neck part of the dummy to make the retraction impossible. This design would be less heavy and the center of mass of the head would be less affected. The muscle substitute did not affect the T1 accelerometer that was the accelerometer closest to the construction.

The N_{km} and F_z values are not representative since the muscle substitute changes the condition too much. N_{km} is based on shear forces and bending moment. With the muscle substitute the load cell measuring N_{km} cannot work as intended to. F_z is affected by the extra weight above the center of mass of the dummy´s head.

In the seats with no WHIPS and in the Toyota-Volvo hybrid with Toyota head rest the injury risk was reduced by 25-50% with the muscle substitute compared to reference. In the seats with active WHIPS the injury risk was low both with and without muscle influence. Head acceleration in x-direction was slightly decreased with muscle substitute compared to reference. This together with an increased head-to-head rest contact time with muscle substitute is according to Carlsson (2010) who found that compared to men; women had an earlier head-to-head rest contact and a higher peak head x-acceleration. She did also find that women had an equal or higher peak T1 x-acceleration which corresponds with my findings.

6 Conclusions

In this Master´s Thesis work neck strength measurements, i.e. retraction resistance, and recordings of EMG activity in the right sternocleidomastoid muscle were performed. The results were compared with previous studies and used to design a muscle substitute for the BioRID dummy. A series of rear impact sled test with and without muscle influence and with seats with and without active WHIPS was performed according to EuroNCAP whiplash test protocol. Based on the findings of the current study, the following can be concluded: In the volunteer test the retraction resistance measurement showed that men are about two times stronger than women. This result is according to literature studies presented in the literature review. The test procedure and test device was simple and the results may not be exact but still it is clearly apparent that neck muscle strength is not the same in men and women. The reaction time interval in the EMG activity measurements corresponds with results from previous work and indicates that the muscle contract was done reflexively rather than voluntarily when a sound stimuli was applied. Although the MyoTrac EMG biofeedback device is less advanced than system used in previous studies it can successfully be used when determining muscle onset time. The overall result from the sled test indicated differences between reference and dummy with muscle substitute foremost in seats with welded WHIPS. With muscle substitute:

- Head acceleration in x-direction decreased.
- Head to head rest contact time increased.
- Neck injury risk in seats without WHIPS was substantially reduced.
- Neck injury risk in seats with WHIPS was low with and without muscle influence.

The aim of this Master´s Thesis was to evaluate the influence of neck muscles in rear impacts. Combining the conclusions of the volunteer and sled tests, female occupants seem likely to have an earlier time of contact with the head rest, higher head acceleration and a higher risk of injury. As this is confirmed by previous findings (Carlsson, 2010 and references showing higher female risk) the role of muscles may be one of the most important in understanding the differences in risk in gender. Furthermore, according to the results of this thesis females may not be at higher risk in cars equipped with WHIPS. This does also correspond with previous findings (Jakobsson, 2005a; 2005b). Further studies needs to be performed where a more advanced design of muscle substitute is used in order to minimize the influence of the extra weight and change of the center of mass of the dummy´s head. But still the results of this study can be used as guidelines for future work and the development of new studies.

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Appendix

A N_{km} curves

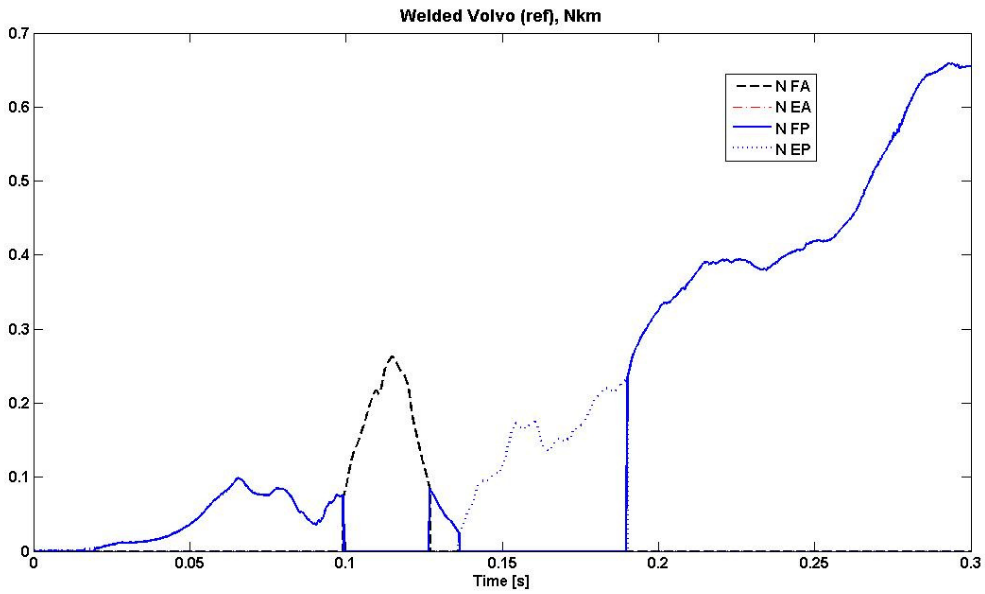

Figure A 1 Nkm for reference (i.e. without muscle substitute) in a welded Volvo seat.

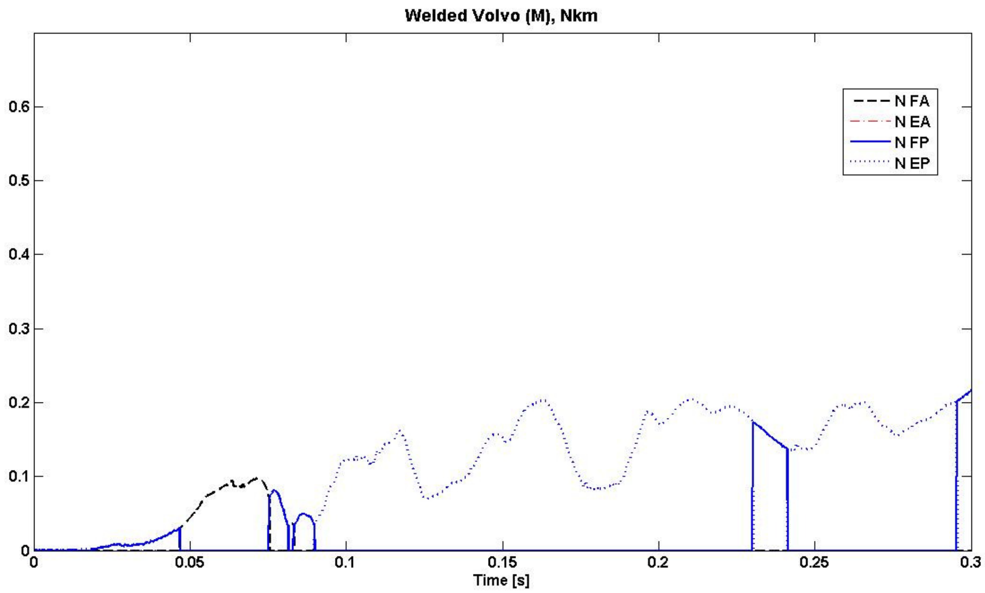

Figure A 2 Nkm for test with muscle substitute in a welded Volvo seat.

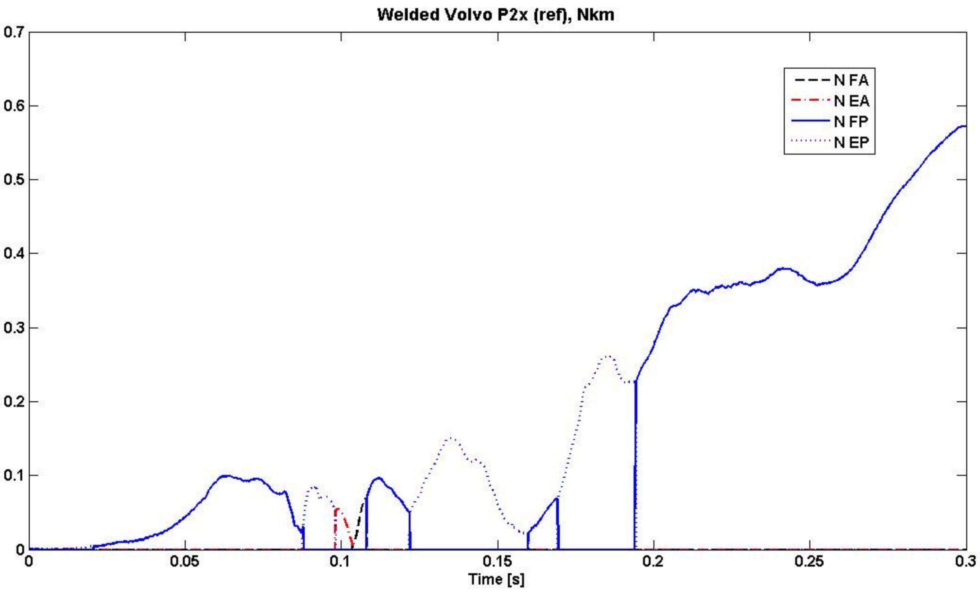

Figure A 3 Nkm for reference in a welded Volvo P2x seat.

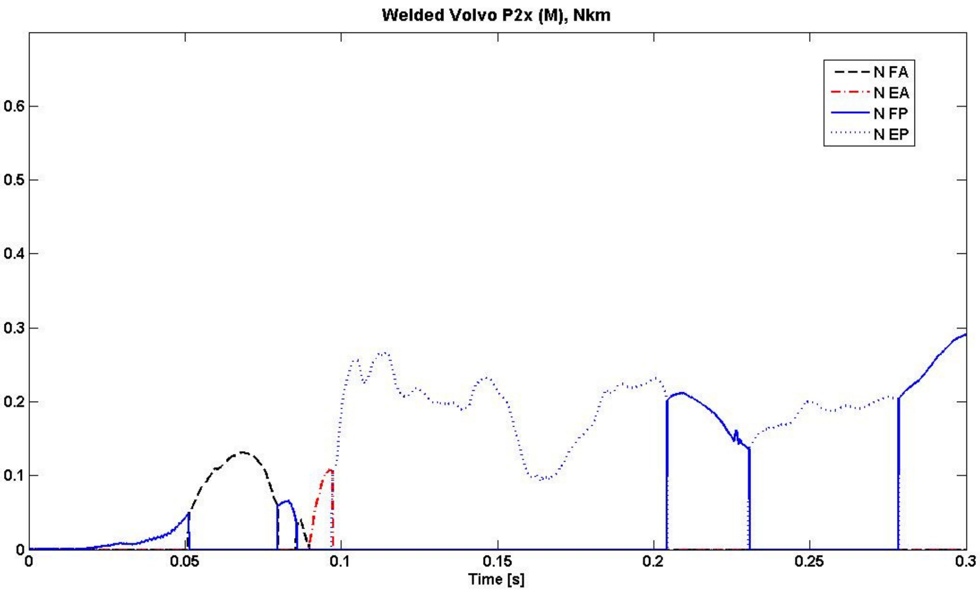

Figure A 4 Nkm for test with muscle substitute in a welded Volvo P2x seat.

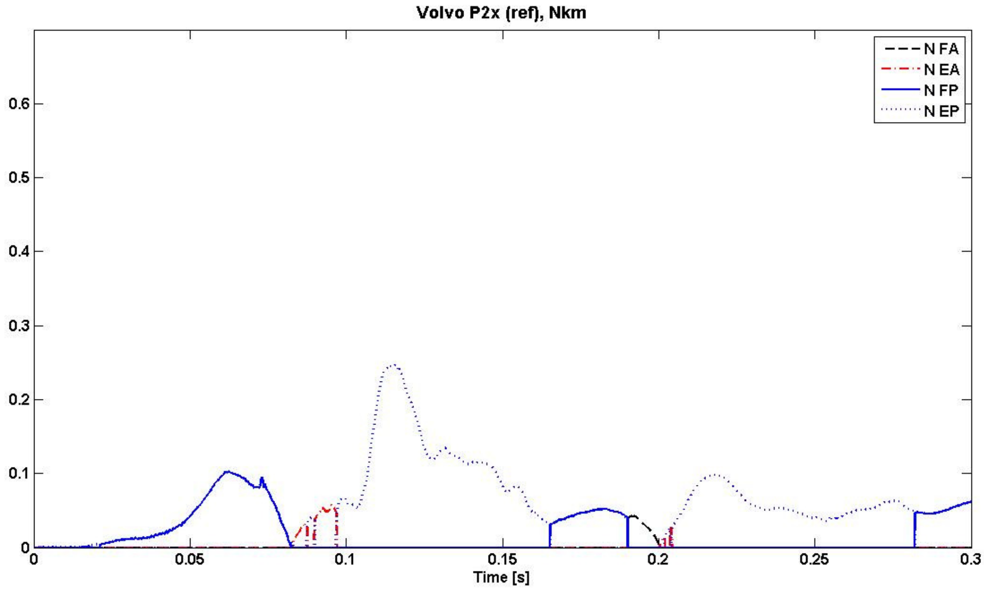

Figure A 5 Nkm for reference in a Volvo seat with active WHIPS.

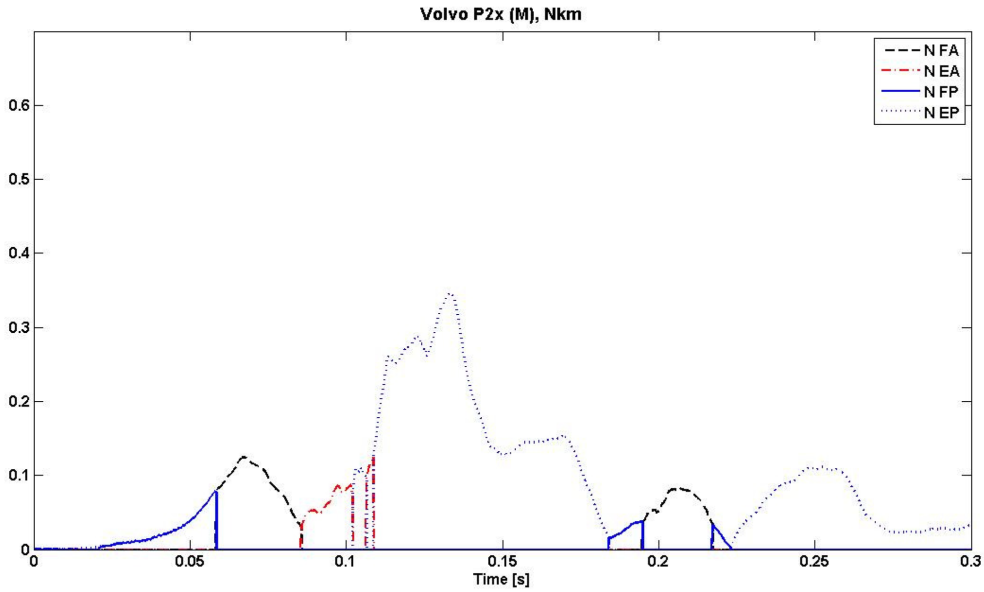

Figure A 6 Nkm for test with muscle substitute in a Volvo seat with active WHIPS.

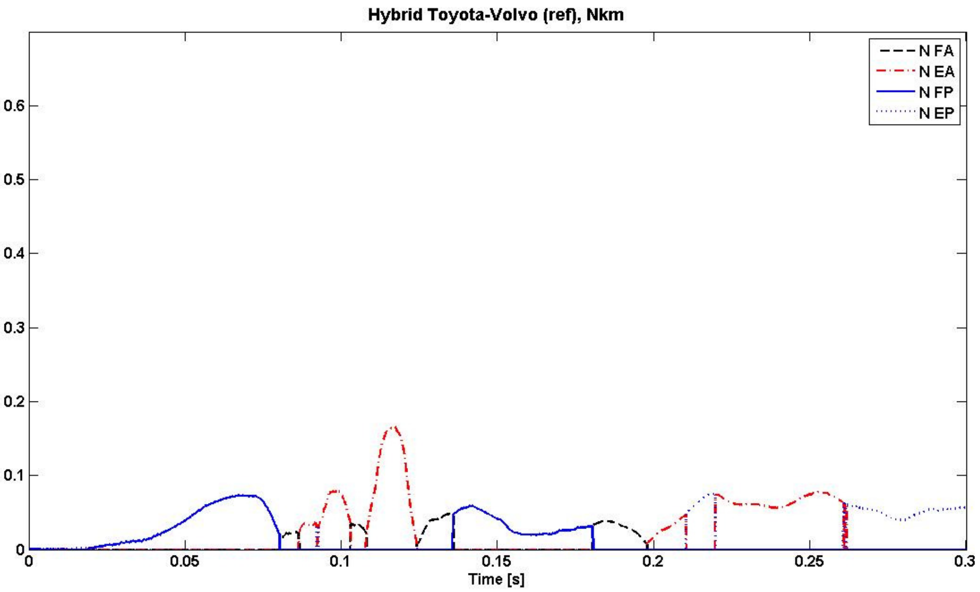

Figure A 7 Nkm for reference in a Hybrid Toyota-Volvo seat with Toyota head rest.

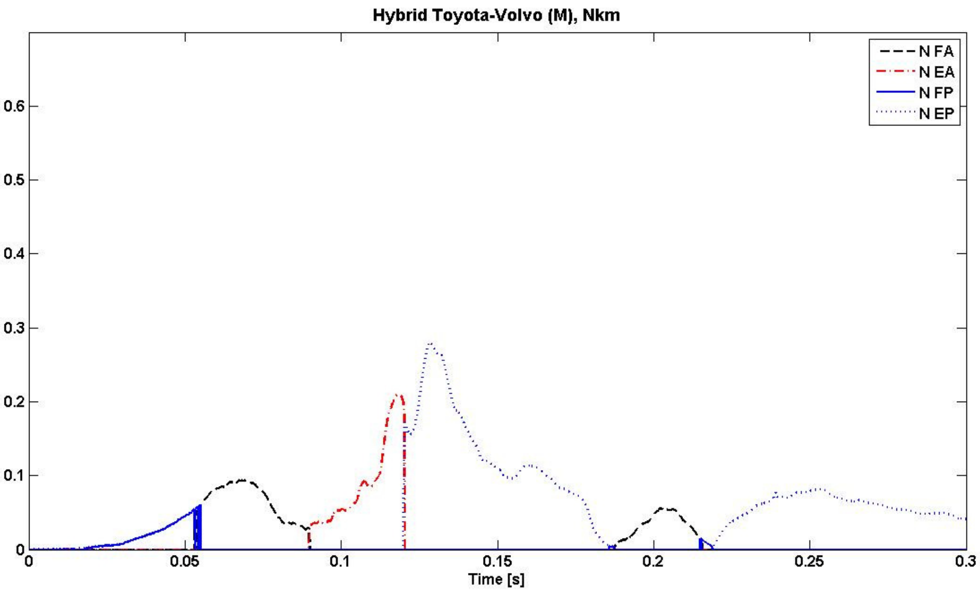

Figure A 8 Nkm for test with muscle substitute in a Hybrid Toyota-Volvo seat with Toyota head rest.

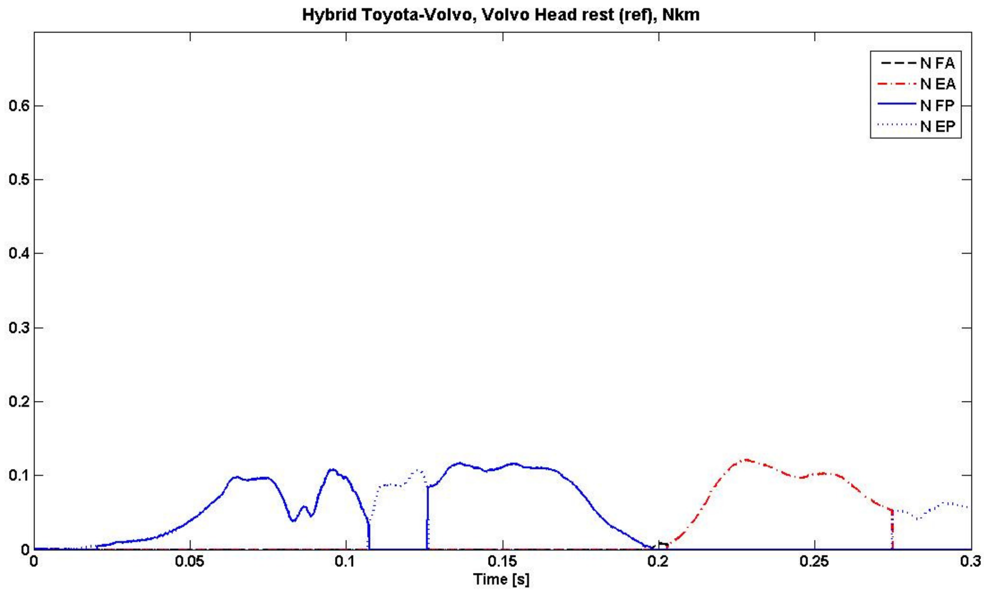

Figure A 9 Nkm for reference in a Hybrid Toyota-Volvo seat with Volvo head rest.

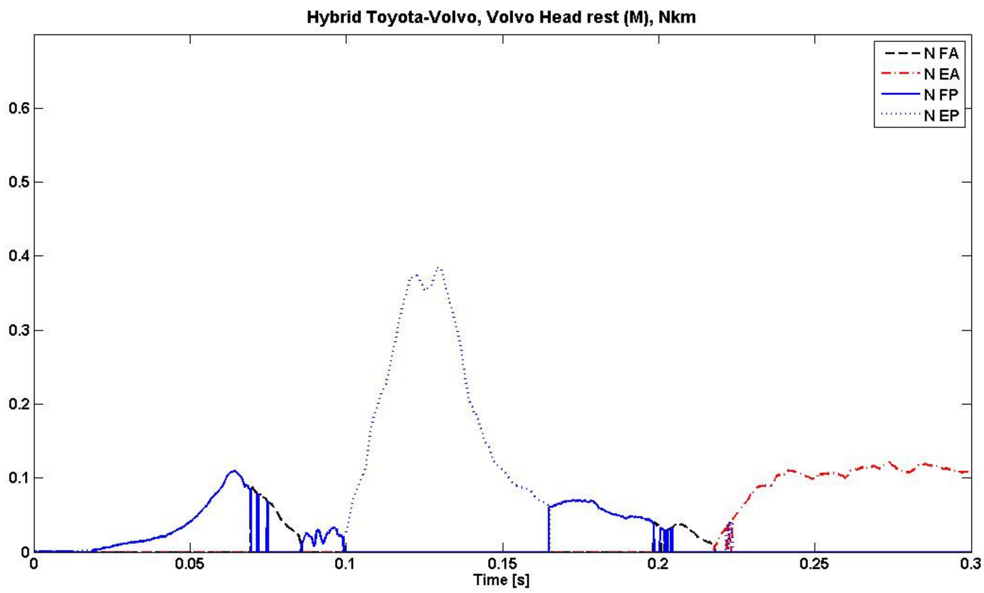

Figure A 10 Nkm for test with muscle substitute in a Hybrid Toyota-Volvo seat with Volvo head rest.

B Results from sled tests

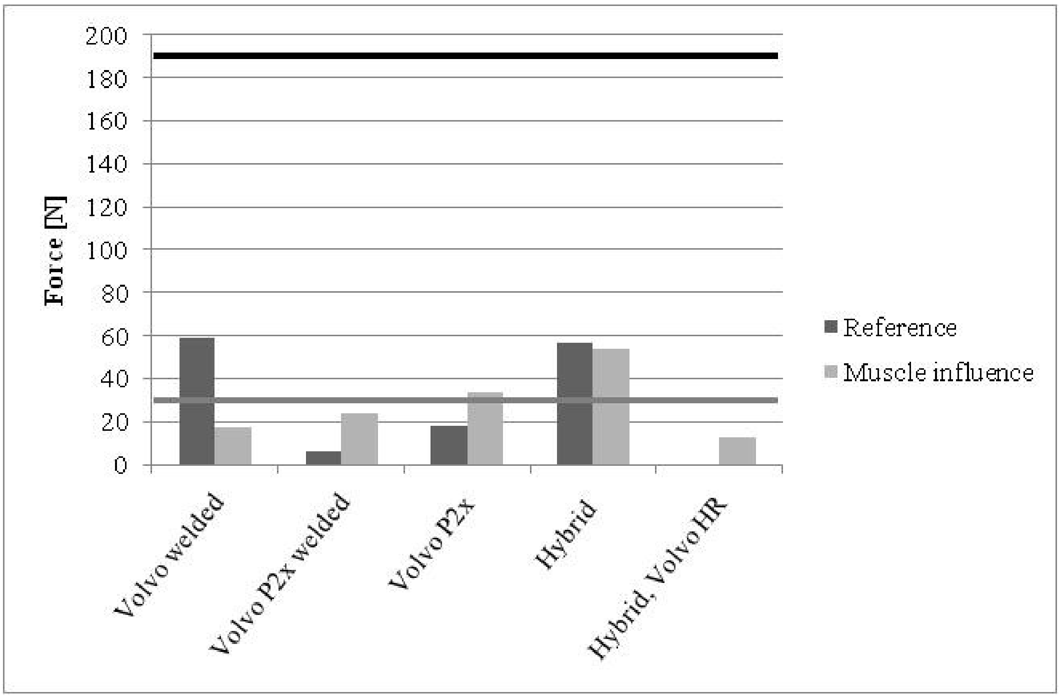

Figure B 1 Upper neck force F_x comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

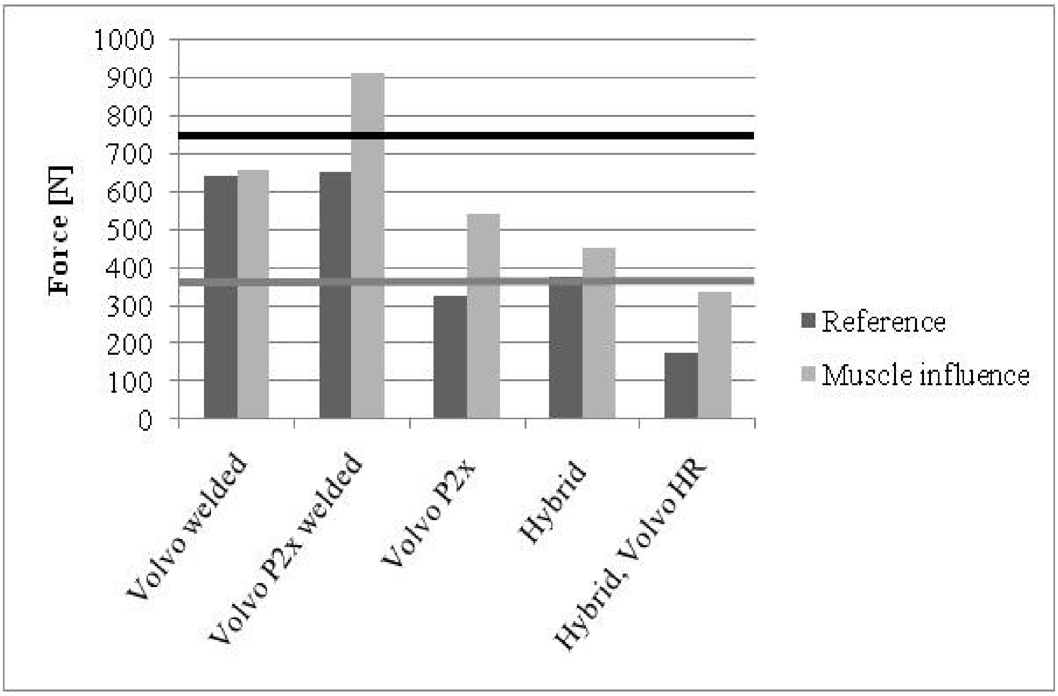

Figure B 2 Upper neck force F_z comparisons for reference and muscle substitute test results for different seat types. The vertical lines represent the EuroNCAP higher performance level (grey) and lower performance level (black).

C EuroNCAP score

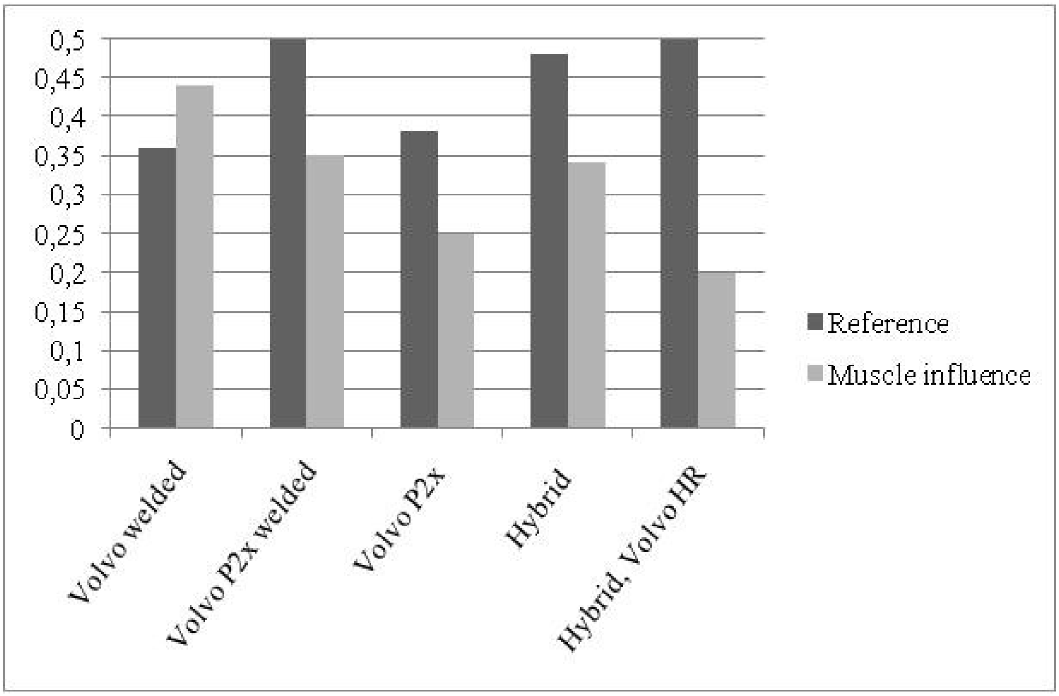

Figure C 1 EuroNCAP score for Nkm

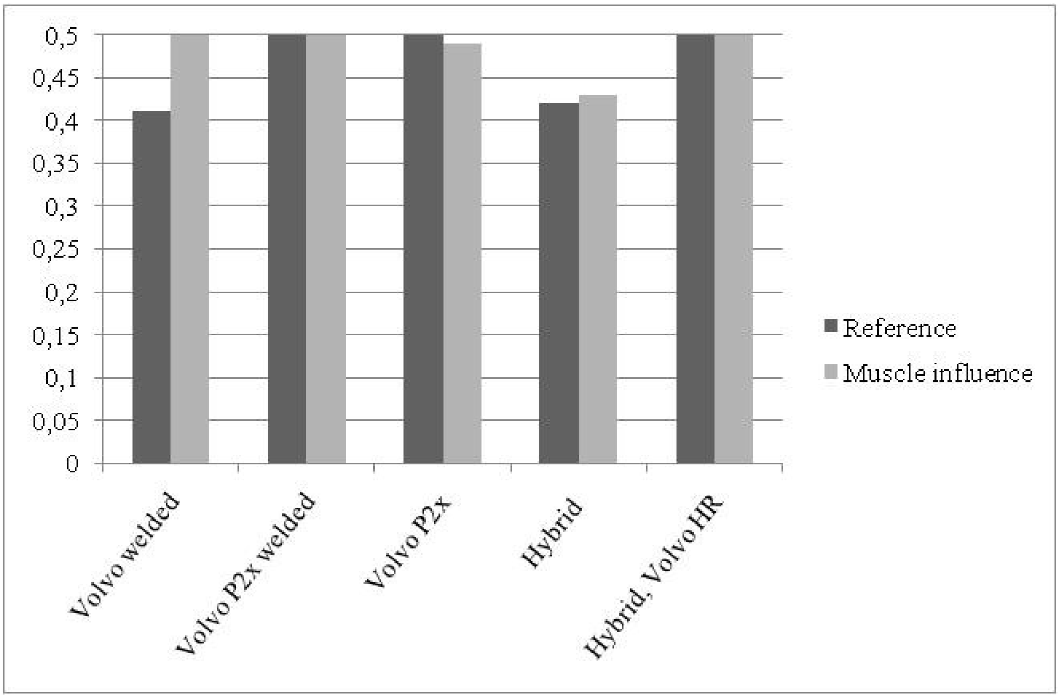

Figure C 2 EuroNCAP score for upper neck force F_x

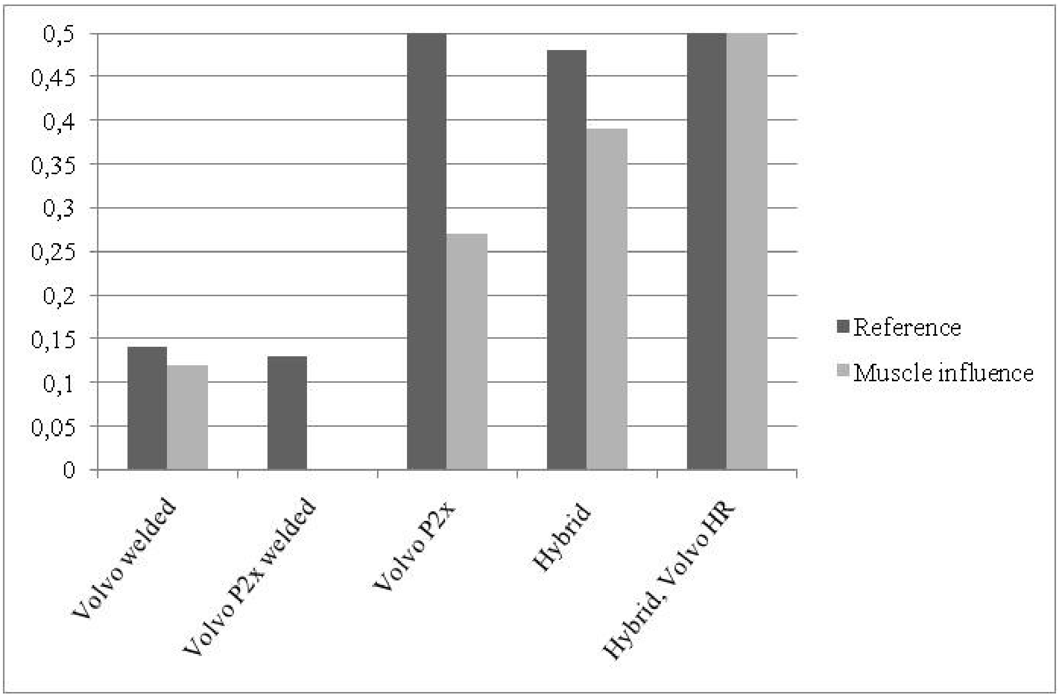

Figure C 3 EuroNCAP score for upper neck force F_z