Neck Injury Criterion Validation Using Human Subjects and Dummies

Mario Darok, Eduard P. Leinzinger, Arno Eichberger and Hermann Steffen

Abstract. In the study described in this chapter, we report the neck injury criterion validation with human volunteers, post-mortem human subjects, and dummies. A total of 70 sled tests with volunteers and 28 sled tests with post-mortem human subjects were analyzed. In addition, computer simulations with a Hybrid III dummy equipped with a modified neck (TRID) were performed using the occupant simulation software MADYMO®. Results showed a correlation between the neck injury criterion and velocity change, crash pulse, head restraint position, neck extension angle, head angular acceleration, and neck. Furthermore, the pressure in the spinal canal of post-mortem human subjects during rear-end impact was measured. In these tests, a correlation of the neck injury criterion with the magnitude of pressure amplitudes of the cerebrospinal fluid during whiplash motion was found. For a correctly positioned head restraint, computer simulations revealed a correlation of the neck injury criterion with neck loads of the Hybrid III dummy.The neck injury criterion explains dangerous impact conditions with respect to soft tissue injuries of the neck with acceptable accuracy. Therefore, the results suggest the use of the neck injury criterion as an indicator for the risk of soft tissue neck injuries.

1. Introduction

Over the past four decades, the so-called whiplash motion following rear-end impact has been held responsible for soft tissue injuries of the neck. This motion consists of translational movement of the spine leading to maximal retraction (s-shape) and extension and, is thus, the reason for severe injuries such as ruptured ligaments or bone fractures caused by rear-end accidents (Figure 1). Ruptured ligaments or bone fractures can be easily diagnosed; in most of the crashes, however, a minor neck injury occurs and even a complete examination fails to reveal any characteristic or evidential findings.

![Initial position, spine straightening, max. retraction, extension](image)

Figure 1: Occupant kinematics during rear impact.

Up to now, the injury mechanism has not been identified nor has an injury criterion been established. Several studies suggest the relation of the injury to neck hyperextension or to maximum neck moments and head excursion angles, respectively [14]. Other studies indicate a possible responsibility of the neck flexion during rebound in a later phase of the
motion and explain the fact by increased seatbelt usage [10]. Also, shear forces in a rather early phase of the typical whiplash motion were taken into account as a possible injury inducing mechanism [23]. Another theory proposed a relation of soft tissue neck injuries to facet joint lesions of the cervical vertebrae [16,25-26]. A hypothesis that relates whiplash injuries to pressure effects in the spinal canal of the cervical spine has been suggested by Aldman et al. [1]. This theory has been investigated in several experiments with pigs (Figure 3) [17,20]. A correlation between pressure amplitudes in the spinal canal and histopathologic findings in the nerve cells of the spinal ganglia was found. As the objects tested were not human, the different anatomy should be considered in the pig results. Furthermore, the simulated swift head motion induced by a pulling force on the head is not exactly comparable to real head-neck motions during a rear-end collision. However, based on these experiments, a mathematical model was built which resulted in the proposal of the neck injury criterion (NIC) [2-3].

\[ \text{NIC} = \alpha_{\text{ad}} x 0.2 + \left( \alpha_{\text{a}}^{2} \right) \]

\( \alpha_{\text{ad}} \) Relative acceleration between first spinal vertebra (C1) and first thoracic vertebra (T1)

\( \alpha_{\text{a}} \) Relative velocity between C1 and T1, i.e., the time integral of \( \alpha_{\text{ad}} \)

Further investigations were taken to apply the NIC on data measured in human subjects and dummies. Would the pressure effects observed in the pig model be replicable in human volunteers or post-mortem human subjects (PMHS), respectively? Is there a correlation of the peak pressure within the spinal column with the NIC value?

![Figure 3: Schematic setup for simulated whiplash extension experiments on pigs by Svensson et al. [20].](image)

2. Methodology

2.1 Human Volunteer Tests

Seventy sled tests with volunteers were analyzed. The tests were performed using different types of car seats (standard and prototype). Impact conditions considered to be the most realistic for rear-end collisions were chosen. Therefore, crash pulses measured in real car crashes by a crash data recorder (UDSM™ by Mannesmann-Kienzle) were simulated on the sled. The first series (Series A) included 36 sled tests with 12 different human subjects that were divided into two groups. Referring to the spine region, all subjects tested were examined by a manual therapist and claimed to be healthy. The first group consisted of five females and one male (body size approximately comparable to the 50th percentile female Hybrid III dummy). The second group consisted of six male subjects (body size approximately comparable to the 50th percentile male Hybrid III dummy). A modified standard car seat was used: The back was fixed in order to prevent bending and the head restraint was replaced by a larger prototype that was in initial contact to the head of the volunteer. The head accelerometer was fixed to the side of the head at approximately at the height of the head center of gravity. The location of the torso accelerometer at the front of the chest was comparable to that of the Hybrid III dummy. Tests were performed at a moderate sled impact speed of 5 to 5.5 kph. The average crash pulse was between 2.5 and 3g, and the crash duration was between 60 to 70 ms. This test series represents a rather hard pulse at low velocity change level (collision without vehicle damage).

For the second series (Series B) of 34 tests performed in 1995 by Elsberger et al., the NIC was also calculated [6]. In this series, tests were performed using different standard car seats and under realistic impact conditions. A moderate crash pulse (crash duration approximately 120 ms) was chosen. The velocity change of the sled was 11 kph for the first 17 tests and 9 kph for the last 17 tests.

Boström et al. proposed the calculation of the NIC at 50 mm head retraction [3]. As this method was not applicable in the present study, a 3 ms peak (using the same calculation routine as for the 3 ms head acceleration criterion) was detected (instead) and defined as the NIC. This method proved to be more accurate and reliable as it demonstrates that the maximum NIC is in accordance with the maximum retraction phase.

2.2 Tests With Post-Mortem Human Subjects (PMHS)

Tests with human subjects were performed according to federal regulations. Ethical guidelines for usage of human cadavers in scientific research were strictly followed. These guidelines included obtaining, treatment, anonymity, and disposal of post-mortem human subjects. Since pressure transducers are not applicable in volunteer tests, a third series (Series C) of tests with post-mortem human subjects (PMHS) was performed [5]. Twenty-eight experiments were carried out with five different PMHS. Pressure transducers were placed in the spinal canal of four PMHS. These experiments led to 21 results with valid pressure measurements, one test failed. The test setup is shown in Figure 3. The sled, accelerated by a Bungee rope to the desired impact speed and decelerated by a pneumatic braking device is shown in Figure 4.

The condition chosen for the test setups was close to the setups chosen for the real-world rear impact crashes. Crash investigations show that most rear impacts occur at low velocity change level. Hell and Terning report that the most common velocity change for the struck car is about 10 kph and the most vulnerable at 16 kph [9,21]. These results give evidence of comparable conditions in Europe. For this reason, sled impact velocities of approximately 9 and 15 kph were chosen. According to the reband velocity of the sled, this led to velocity changes (of the sled) of 10 and 16 kph. As in the volunteer tests, the parameters of the sled pulses were based upon crash recordings at these velocity changes. A contact switch was used to determine the first contact of the sled with the brake and thus define time zero (\( T_{0} \)). This moment corresponded with the first contact of two colliding
motion and explain the fact by increased seatbelt usage [10]. Also, shear forces in a rather early phase of the typical whiplash motion were taken into account as a possible injury inducing mechanism [23]. Another theory proposed a relation of soft tissue neck injuries to facet joint lesions of the cervical vertebrae [16,25-26]. A hypothesis that relates whiplash injuries to pressure effects in the spinal canal of the cervical spine has been suggested by Aldman et al [1]. This theory has been investigated in several experiments with pigs (Figure 3) [17,20]. A correlation between pressure amplitudes in the spinal canal and histopathologic findings in the nerve cells of the spinal ganglia was found. As the objects tested were not human, the different anatomy should be considered in the pig results. Furthermore, the simulated swift head motion induced by a pulling force on the head is not exactly comparable to real head-neck motions during a rear-end collision. However, based on these experiments, a mathematical model was built which resulted in the proposal of the neck injury criterion (NIC) [2-3].

\[
\text{NIC} = a_{\text{rel}} \times 0.2 + (v_{\text{rel}})^2 \quad \text{(tolerance level: 15 m/s}^2\text{)}
\]

Where:
- \(a_{\text{rel}}\) is the relative acceleration between first spinal vertebra (C1) and first thoracic vertebra (T1)
- \(v_{\text{rel}}\) is the relative velocity between C1 and T1, i.e., the time integral of \(a_{\text{rel}}\)

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Figure 3: Schematic setup for simulated whiplash extension experiments on pigs by Svensson et al [20].

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vehicles. The test conditions are listed in Table 1. Velocity change, acceleration level, and head restraint position were varied in order to study their effects.

![PMHS test setup](image)

Figure 3: PMHS test setup.

The PMHS were placed on a Volkswagen PKO seat. Longitudinal and height adjustments were locked in the middle position. The seat was checked after each test, and in case of damage, it was replaced. An upright head position was ensured by a foam block (applied at the chest) that did not influence the head-neck kinematics during impact. Twenty-eight tests were performed with five PMHS (4 males, 1 female). Only four were equipped with pressure transducers. Their age ranged from 30 and 87 years with an average of 62 years. The post-mortem time varied between 42 and 110 hours. According to the case history in all subjects, a relevant trauma prior to death could be excluded (Table 2).

![Pneumatic brake](image)

Figure 4: Pneumatic brake.

### Table 1: Test conditions of pressure experiments.

<table>
<thead>
<tr>
<th>Test ID</th>
<th>Sled Impact Velocity (mph)</th>
<th>Sled Average Acceleration (g)</th>
<th>Sled Max Acceleration (g)</th>
<th>Horiz. Dist. (head to head restraint) (mm)</th>
<th>Vert. Dist. (top of head to top of head restraint) (mm)</th>
<th>Seatback Inclination (deg)</th>
<th>Subject</th>
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### Table 2: Post-mortem human subject data.

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<th>Body Weight (kg)</th>
<th>Body Height (cm)</th>
<th>Time Since Death (h)</th>
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</table>

Between death and the experiments, the PMHS were stored at 4 degrees Celsius. Rigor mortis (shoulders, elbows, pelvis, knees) was removed prior to test. The neck region was not moved before the tests. Pressure measurements in the cerebrospinal fluid (CSF) were performed using catheter-tip pressure transducers placed subdurally in the spinal canal, similar to the tests performed with pigs by Svensson et al [20]. Due to the anatomical differences between pigs and humans, a new technique for positioning the transducers in the spinal canal had to be developed. Therefore, a small opening (approximately 2 cm) was carefully made into the top of the cranium, slightly right to the center line, without causing any damage to the dura or influencing the pressure effects (Figure 5). The distance for fixing the transducers to a metal wire was estimated from the neck length of the PMHS (Figure 6). The position of the upper transducer was at C1-C2 and the lower transducer at C6-C7. The distance between the transducers was about 8 cm. The stiffness of the wire was not considered to have any influence on the head-neck kinematics. The metal wire was then passed through the small opening in the cranium, through the dura and the brain, and placed in the spinal canal. This special technique for positioning the pressure transducers was used to avoid any damage of the soft tissues of the neck, which probably would have caused an unwanted alteration of neck mobility. Following the series of tests, in all cases, an autopsy was performed to locate the position of
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<td>11.7</td>
<td>75</td>
<td>34</td>
<td>25</td>
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</tr>
<tr>
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<td>11.7</td>
<td>90</td>
<td>248</td>
<td>20</td>
<td>PMHS98#5</td>
</tr>
</tbody>
</table>

### Table 2: Post-mortem human subject data.

<table>
<thead>
<tr>
<th>PMHS I.D.</th>
<th>Sex</th>
<th>Age</th>
<th>Body Weight (kg)</th>
<th>Height (cm)</th>
<th>Time Since Death (h)</th>
<th>Cause of Death</th>
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</tr>
<tr>
<td>PMHS98#3</td>
<td>F</td>
<td>86</td>
<td>50</td>
<td>149</td>
<td>85</td>
<td>heart failure</td>
</tr>
<tr>
<td>PMHS98#4</td>
<td>M</td>
<td>57</td>
<td>95</td>
<td>173</td>
<td>110</td>
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<tr>
<td>PMHS98#5</td>
<td>M</td>
<td>87</td>
<td>75</td>
<td>159</td>
<td>70</td>
<td>coronary failure</td>
</tr>
</tbody>
</table>

Between death and the experiments, the PMHS were stored at 4 degrees Celsius. Rigor mortis (shoulders, elbows, pelvis, knees) was removed prior to test. The neck region was not moved before the tests. Pressure measurements in the cerebrospinal fluid (CSF) were performed using catheter-tip pressure transducers placed subdually in the spinal canal, similar to the tests performed with pigs by Svensson et al [20]. Due to the anatomical differences between pigs and humans, a new technique for placing the transducers in the spinal canal had to be developed. Therefore, a small opening (approximately 2x2 cm) was carefully made into the top of the cranium, slightly right to the center line, without causing any damage to the dura or influencing the pressure effects (Figure 5). The distance for fixing the transducers to a metal wire was estimated from the neck length of the PMHS (Figure 6). The position of the upper transducer was at C1-C2 and the lower transducer at C6-C7. The distance between the transducers was about 8 cm. The stiffness of the wire was not considered to have any influence on the head-neck kinematics. The metal wire was then passed through the small opening in the cranium, through the dura and the brain, and placed in the spinal canal. This special technique for positioning the pressure transducers was used to avoid any damage of the soft tissues of the neck, which probably would have caused an unwanted alteration of neck mobility. Following the series of tests, in all cases, an autopsy was performed to locate the position of
the transducers (Figure 7). Due to the equipment available for these experiments, the vascular system of the PMHS could not be pressurized and it was not possible to take X-rays.

Figure 5: Location of opening for transducer positioning.

Figure 6: Pressure transducers (arrows) fixed to a metal wire.

Two triaxial accelerometers (Endevco 7267A) placed on the head and chest of the PMHS were used. The dynamic measurement of the spinal curvature was performed utilizing one biaxial accelerometer at the height of T1 (Entrap EGAS), one angular accelerometer at the head (Endevco 7302B), two catheter tip pressure transducers in the spinal canal, and a so-called spine band (Figure 8). The pressure transducer was the same model as used in the pig experiments [20]. Sled acceleration in longitudinal and lateral directions was measured by a biaxial accelerometer (Mannesmann UDS™ crash recorder). Transducer signals were transferred via cable to a DC-amplifier (Dewetron DAQ). The signals were sampled at 10 kHz by a 64 channel data acquisition board for personal computers (Microstar DAP 3000u, DAP MSXB018). PMHS transducer data were filtered with a CFC 1000 filter. The sled acceleration was filtered with a CFC 60 filter. Data were sampled and analyzed by DIADEM™ data acquisition software for personal computers. Calculation methods for CFC filter and injury criteria (e.g., 3 ms maximum) were implemented according to specifications by the automotive industry. For statistical analyses, correlation ($r$ and $r^2$) and significance values ($p$) were calculated.

Figure 7: Location of pressure transducer in situ (PMHS9842).

Figure 8: Instrumentation of the PMHS.

3. NIC Calculation and Pressure Amplitude

The NIC calculation method of Bostrom et al. was applied and is shown in figure 9 [2]. This method also has been used introducing this criterion based upon the pig experiments [20]. Modifications of the original formula were applied according to Lichtenberg et al. Resultant accelerations were used for NIC calculations instead of the x-components [7]. Because of the natural curvature of the human spine, the alignment of the T1-accelerometer to the longitudinal (x-) direction was difficult. Considering the modifications, no significant deviations of our results from the original formula could be observed [7]. Therefore, the NIC was defined as the 3 ms maximum of the first 150 ms ($\text{NIC}_{3\text{ms}}$) of the time history of the modified NIC$_{\text{3ms}}$ (Figure 10). According to the NIC theory, transducer pressure closest to the C4 level was calculated and compared with the
the transducers (Figure 7). Due to the equipment available for these experiments, the vascular system of the PMHS could not be pressurized and it was not possible to take x-rays.

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calculated NIC as described above. The definition of the pressure amplitude as investigated by Bosström et al. is shown in figure 11. The same definition was used for the PMHS tests. The NIC represents the amplitude, which is positive or negative depending on the location of the pressure transducer and the mode of the cervical spine. In case of a retraction movement during a rear impact, the NIC is positive (Figure 11).

\[
NIC = NIC_{\text{max}}(t = t_{\text{max}}) = \frac{\sigma_{\text{max}}(t) + \sigma_{\text{max}}(t-0.2) + \sigma_{\text{max}}(t-0.4)}{3}
\]

where:
- \( \sigma_{\text{max}}(t) \) longitudinal acceleration of first thoracic vertebra (T1)
- \( \sigma_{\text{max}}(t) \) longitudinal acceleration of first cervical vertebra (C1)
- \( \sigma_{\text{max}}(t) \) relative acceleration between T1 and C1
- \( v_{\text{max}}(t) \) relative velocity between T1 and C1
- \( NIC_{\text{max}}(t) \) time history of NIC
- \( NIC_{\text{max}}(t) \) NIC at 5-hp (maximum reaction) of the cervical spine
- \( t_{\text{max}} \) instant of time at 5-hp (maximum reaction) of the cervical spine

Figure 9: NIC formula proposed by Bosström et al. [2].

\[
\text{NIC}_{\text{max}}(t) = \text{Maximum of } \left( \sigma_{\text{max}}(t) + \sigma_{\text{max}}(t-0.2) + \sigma_{\text{max}}(t-0.4) \right)
\]

where:
- \( \sigma_{\text{max}}(t) \) resultant acceleration of first thoracic vertebra (T1), lateral component was neglected
- \( \sigma_{\text{max}}(t) \) resultant acceleration of head (COG)
- \( \sigma_{\text{max}}(t) \) relative resultant acceleration between T1 and head COG
- \( v_{\text{max}}(t) \) relative velocity between T1 and head COG
- \( NIC_{\text{max}}(t) \) time history of modified NIC

Figure 10: Modified NIC calculation by Eichberger et al. [5].

3.1 Dummy Test Simulations

In order to improve neck protection and to develop suitable seat systems, the automotive industry needs tools to evaluate their effectiveness. Therefore, it proved to be useful to apply the NIC in tests where an anthropomorphic test device (ATD) replaces the human. Hybrid III dummies, which have been developed for high-speed frontal impacts, have shown unsatisfying biofidelity in low-speed rear-end impacts [8,18]. Currently, a lot of effort is being done to improve the biofidelity by development of a special rear impact dummy [4,20,22]. In our study, the investigation of the NIC was performed only with one kind of dummy because the objective was not to investigate the biofidelity of new dummy designs, but to study the principal behavior of NIC in dummy tests. Therefore, the Hybrid III dummy, equipped with a modified neck (TRID) was used [22]. The performance of the NIC in ATD tests was evaluated by investigations of parameters that proved to be of importance. Such parameters are velocity change, acceleration level, head restraint position, seatback properties, etc. In addition, the relationship between the NIC and neck loads was investigated. The investigation of all relevant parameters would have required a large amount of sled tests. Therefore, computer simulations were performed (occupant simulation software Madymo®). A positive side effect of these simulations is that stochastic influences of laboratory conditions (such as repetitability of sled tests) are avoided. The seat of a middle class car was modeled by using multi-body simulation techniques. Therefore, properties of the seat cushion, recliner joint, and the head restraint were measured in approximately static loading tests and implemented to the model. The test object, a database of the 50th percentile Hybrid III dummy equipped with a more biofidelic neck (TRID), was positioned according to standard seating procedures. Seat and head restraints were adjusted to standardized positions as used in frontal crash testing. The seat and whole model are illustrated in Figure 12. Validation of the whole model was performed in a series of sled tests using the same conditions as in the numerical model. Tests were performed at three different velocities and repeated several times to reproduce the obtained results. Results of the simulation and sled experiments showed acceptable agreement. A parametric study was performed by variation of the parameters listed above. According to the formula setup at a NIC standardization meeting (Gothenburg, 1998), the NIC was calculated for these simulations. The results were analyzed and compared to upper and lower neck loads. However, it has to be emphasized that the head movement was not investigated in the computer simulation.

Figure 11: Definition of \( p \) and \( NIC_{\text{max}} \).

Figure 12: Computer simulation model.
calculated NIC as described above. The definition of the pressure amplitude as investigated by Bostrom et al. is shown in Figure 11. The same definition was used for the PMHS tests. The NIC represents the amplitude, which is positive or negative depending on the location of the pressure transducer and the mode of the cervical spine. In case of a retraction movement during a rear impact, the NIC is positive (Figure 11).

\[
NIC = NIC_{max}(t = t_{max}) - \sigma_{max}(0)\sigma_{y}(0) + \tau_{min}(0)
\]

- \( \sigma_{max}(0) \): maximum amplitude of the cervical spine
- \( \sigma_{y}(0) \): yield strength of the cervical spine
- \( \tau_{min}(0) \): minimum time before neck retraction

Figure 10: Modified NIC calculation by Eichberger et al [5].

3.1 Dummy Test Simulations

In order to improve neck protection and to develop suitable seat systems, the automotive industry needs tools to evaluate their effectiveness. Therefore, it proved to be useful to apply the NIC in tests, where a human dummy is replaced by a Hybrid III dummy, which has been developed for high-speed frontal impacts. The Hybrid III dummy was equipped with a modified neck (TRID) to reduce the performance of the NIC in ADAMS. In tests that were performed by a modified neck (TRID), the NIC was calculated for the Hybrid III dummy. In Figure 12, the seat and the seat restraint were adjusted to standardized positions as used in frontal crash testing. The seat and whole model are illustrated in Figure 12. Validation of the whole model was performed in a series of sled tests using the same conditions as in the numerical model. The test object, a database of the 50th percentile Hybrid III dummy equipped with a more biofidelic neck (TRID), was positioned according to standard seating procedures. Seat and head restraints were adjusted to standardized positions as used in frontal crash testing. The seat and whole model are illustrated in Figure 12. Validation of the whole model was performed in a series of sled tests using the same conditions as in the numerical model. Tests were performed at three different velocities and repeated several times to reproduce the results. Of all simulations, a parameter study was performed by variation of the parameters listed above. According to the formula setup at a NIC standardization meeting (Gothenburg, 1998), the NIC was calculated for these simulations. The results were analyzed and compared to upper and lower neck loads. However, it has to be emphasized that the reboard motion was not investigated in the computer simulation.
4. Results and Discussion

4.1 Human Volunteer Tests

Due to the low-impact speed and the optimal adjusted head restraint, injuries of the cervical spine are considered to be very unlikely for series A. Therefore, this test series is expected to result in NIC values far below the injury threshold. The results of one of these tests are shown in figure 13. None of the volunteers complained about injuries in the cervical spine region after the test. Even days after the test, no neck pain was reported. One volunteer complained about minor pain in the lumbar region. Although other reasons could be responsible, it was assumed that due to the rather hard sled pulse and the fixed seatback, loading of the lumbar and thoracic spine was higher than in cases of real crashes.

![Figure 13: Resultant acceleration in series A.](image)

Using the formula described in figure 9, the NIC time history was calculated for all 36 tests. Considering that the accelerations of C1 and T1 were used in the NIC formula, the results of the chest accelerometer are comparable to T1 accelerations and the head acceleration to C1 acceleration. The influence of the exact location was considered small and was therefore, neglected. This also proved to be of minor importance in the PMHS series. No substantial head rotation was noticed and therefore the maximum retraction of the neck was assumed to occur at the maximum value of the NIC-time history. In all tests, a decrease of the NIC at a certain time (typically 80 to 120 ms, depending on impact conditions) could be observed. Figure 14 illustrates a typical result of the NIC calculations in series A. A 3 ms peak was calculated and defined as the NIC.

![Figure 14: NIC-time history in series A.](image)

Using only the x-components of the head and chest accelerations, the NIC was calculated and thus defined as the NIC-x. In order to apply the NIC to other test series where an extraction of the x-component from the resultant accelerations was not possible (see series B), the resultant head and torso accelerations were used for the NIC calculation (NIC-RES). With regard to the x-component as the dominant direction and the calculation of the NIC at maximum retraction where influences from head rotations are negligible, the results of both calculations are similar. As illustrated in figure 15, NIC values of 2 to 4 m/s², far below the proposed injury threshold (15 m/s²), were observed. In comparison, the results of the NIC-x are in agreement with those of the NIC-RES (τ = 0.71). Larger deviations can be explained by the mode of the accelerometers (see PMHS series).

![Figure 15: NIC results of series A.](image)

In test series B, NIC calculations were performed similar to series A, using 3 ms maximum of the NIC-time histories. As described above, only the resultant accelerations (NIC-RES) were possible to use for the NIC calculation. Figures 16 and 17 show that the maximum NIC occurs approximately after completion of the initial head flexion and start of the extension movement.

Results of the NIC calculation for both impact velocities (same method as series A) are shown in figure 18. Compared to series A, the NIC values increased significantly, especially with a larger initial gap of head and head restraint. The correlation between the NIC and the horizontal distance of head and head restraint is illustrated in figures 19 and 20. The NIC increases with the horizontal distance of head and head restraint, measured at the time of the first contact of the sled and the deceleration element. In addition, similar to the correlation of the NIC and the horizontal distance, an approximate correlation between the NIC and the maximum relative angle of head and torso would be observed in this series (Figure 21). Although the proposed threshold of 15 m/s² was not reached in this series, some volunteers suffered minor muscle pain that started some hours after the test and lasted for one day. One volunteer (NIC-RES = 10.9) complained about symptoms of whiplash-associated disorders (WAD) for three weeks (muscle pain, restricted range of spine motion, headache). It was not possible to diagnose the injury objectively (MRI, x-ray, blood test). However, in this volunteer, a minor post-traumatic change of the spine could not be excluded. Therefore, the injury threshold of this subject could have been lower than that of an average human.
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Using only the x-components of the head and chest accelerations, the NIC was calculated and thus defined as the NIC-x. In order to apply the NIC to other test series where an extraction of the x-component from the resultant accelerations was not possible (see series B), the resultant head and torso accelerations were used for the NIC calculation (NIC-RES). With regard to the x-component as the dominant direction and the calculation of the NIC at maximum neck rotation, the results from head rotations are negligible, the results of both calculations are similar. As illustrated in Figure 15, NIC values of 2 to 4 m/s², far below the proposed injury threshold (15 m/s²), were observed. In comparison, the results of the NIC-x are in agreement with those of the NIC-RES (r = 0.71). Larger deviations can be explained by the mode of the accelerometer (see PMHS series).

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![Figure 16: NIC-time history in series B.](image)

![Figure 17: Head angle - time history in series B.](image)

![Figure 18: NIC results of series B.](image)

4.2 Post-Mortem Human Subject Tests

In test series C, 28 experiments with PMHS were performed [5]. Five subjects (4 male, 1 female) were placed at a Volkswagen PKO seat. Tests were performed at impact speeds of approximately 9 and 15 kph. The following parameters were varied: impact speed, crash pulse, seatback inclination, and head restraint position. Autopsy revealed that four subjects remained uninjured and a hyperextension injury (minor ligamentous damage of the cervical spine) of one subject as a result of removing the head restraint was observed. In this series, the acceleration of T1 was measured and therefore used for calculating the NIC. Only the x-component of T1 was used. A comparison of results of the NIC-RES was also calculated similar to the volunteer tests as described above (using the resultant chest and head acceleration).

Correlation between the NIC-X and the NIC-RES (r = 0.69) was observed, however, in some of the tests (especially tests C-14 to C-17) deviations occurred (Figure 22), which can be explained by the mode of the sensor attachment on the subject: The cervical spine of
(Figure 21). Although the proposed threshold of 15 m/s² was not reached in this series, some volunteers suffered minor muscle pain that started some hours after the test and lasted for one day. One volunteer (NIC-RES = 10.9) complained about symptoms of whiplash-associated disorders (WAD) for three weeks (muscle pain, restricted range of spine motion, headache). It was not possible to diagnose the injury objectively (MRI, x-ray, blood test). However, in this volunteer, a minor pre-traumatic change of the spine could not be excluded. Therefore, the injury threshold of this subject could have been lower than that of an average human.

![NIC-time history in series B.](image1)

![Head angle - time history in series B.](image2)

![NIC results of series B.](image3)

Figure 16: NIC-time history in series B.

Figure 17: Head angle - time history in series B.

Figure 18: NIC results of series B.

Figure 19: Correlation between NIC and horizontal distance (ΔV = 11 km/h).

Figure 20: Correlation between NIC and horizontal distance (ΔV = 9 km/h).

Figure 21: NIC and head extension.

4.2 Post-Mortem Human Subject Tests

In test series C, 28 experiments with PMHS were performed [5]. Five subjects (4 male, 1 female) were placed at a Volkswagen PKO seat. Tests were performed at impact speeds of approximately 9 and 15 km/h. The following parameters were varied: Impact speed, crash pulse, seatback inclination, and head restraint position. Autopsy revealed that four subjects remained uninjured and a hyperextension injury (minor ligamentous damage of the cervical spine) of one subject as a result of removing the head restraint was observed. In this series, the acceleration of T1 was measured and therefore used for calculating the NIC. Only the x-component of T1 was used. A comparison of results of the NIC-RES was also calculated similar to the volunteer tests as described above (using the resultant chest and head acceleration).

Correlation between the NIC-X and the NIC-RES (r = 0.69) was observed, however, in some of the tests (especially tests C-14 to C-17) deviations occurred (Figure 22), which can be explained by the mode of the sensor attachment on the subject: The cervical spine of
the subject (elderly female) was bent forward, resulting in a wrong x-direction of the sensor (figure 23 left); figure 23 right, shows the mode of the sensor attachment on a young male. It was not possible to recalculate this angular error because the quality of the high-speed video did not allow a video analysis of the T1 target. Therefore, it is obvious that only the NIC-RES values are defined correctly and are thus discussed below.

![Graph showing NIC and NIC-RES](image)

**Figure 22:** Results of NIC calculation in PMHS tests.

Positioning the pressure transducers was most successful in PMHS #2. The upper transducer was at the C2 level and the lower at C7, both of them subcutaneously inside the CSF space. The location of the pressure transducers is shown in Table 3. In PMHS #4, the upper transducer was located at the oval foramen and the lower transducer at the C5 level which caused a small damage of the dura at the C6 level. The caused leakage was minor and therefore not considered to notably influence the pressure readings. In PMHS #5, the pressure transducers were detected at the C1 and C6 levels inside the spinal cord. In addition, during autopsy, each subject was examined for neck lesions. A lesion occurred only in PMHS #5. It consisted of a rupture of the ligamentum longitudinale anterius between C5 and C6. This lesion resulted from the last test series which was performed without a head restraint. All other subjects showed no sign of a neck lesion. Ruptures of small vessels leading to a local hemorrhage during active blood circulation or damage to nerve fibers, as reported by Orentgen, could not be detected during autopsy [17]. The results of the pressure measurement of the cerebrospinal fluid are shown in figure 24. This proves an extensive influence of the transducer location on the results.

![Diagram showing mounting of T1 transducer](image)

**Figure 23:** Mounting of T1 transducer.

<table>
<thead>
<tr>
<th>PMHS</th>
<th>Upper Transducer</th>
<th>Lower Transducer</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMHS #2</td>
<td>C1</td>
<td>C1</td>
</tr>
<tr>
<td>PMHS #3</td>
<td>cerebellum</td>
<td>C1</td>
</tr>
<tr>
<td>PMHS #4</td>
<td>oval foramen</td>
<td>C6</td>
</tr>
<tr>
<td>PMHS #5</td>
<td>C1</td>
<td>C6</td>
</tr>
</tbody>
</table>

**Table 3:** Location of pressure transducers in the PMHS tests.

![Graph showing pressure measurements](image)

**Figure 24:** Pressure measurements of each subject.

NIC calculations were performed for all tests and compared to parameters that might indicate soft tissue neck injuries [7]. The NIC was intended for predicting the pressure amplitude (p) at the C4 level of the spinal canal [2]. Since exact positioning of the transducers was difficult, NIC results are compared to pressure measurements that were closest to the C4 level (i.e., C2 transducer for PMHS #2, C1 for PMHS #3, and C6 for PMHS #4 and PMHS #5). In comparison, the NIC and pressure amplitude (p) showed an almost linear correlation (Figure 25). However, the pressure amplitudes of PMHS #3 (very old male) differed significantly from the other subjects. Autopsy revealed nearly no cerebrospinal fluid in the spinal canal and the skull; therefore, results of the pressure measurement were not valid for this subject. Therefore, we propose positioning the transducers within the CSF space for further investigation. Due to the fact that the pressure measurement of PMHS #3 is not valid, these tests are removed in figure 25; the results of figure 24 were illustrated again in a single diagram (Figure 25) and a statistical analysis of the correlation was performed. The PMHS tests showed an absolute correlation of the NIC (NICabs) and the pressure which confirms the results of the tests with pigs (Figure 26). In addition, the time of appearance of the NIC and the pressure is of interest (Figure 27). The results show that NIC occurs often before the pressure amplitude (approximately 30 ms).
the subject (elderly female) was bent forward, resulting in a wrong x-direction of the sensor (figure 23 left); figure 23 right, shows the mode of the sensor attachment on a young male. It was not possible to recalculate this angular error because the quality of the high-speed video did not allow a video analysis of the T1 target. Therefore, it is obvious that only the NIC-RES values are defined correctly and are thus discussed below.

![Figure 23: Mounting of T1 transducer.](image)

Positioning the pressure transducers was most successful in PMHS #2. The upper transducer was at the C2 level and the lower at C7, both of them substantially inside the CSF space. The location of the pressure transducers is shown in Table 3. In PMHS #4, the upper transducer was located at the oval foramen and the lower transducer at the C5 level which caused a small damage of the dura at the C6 level. The caused leakage was minor and therefore not considered to notably influence the pressure readings. In PMHS #5, the pressure transducers were detected at the C1 and C6 levels inside the spinal cord. In addition, during autopsy, each subject was examined for neck lesions. A lesion occurred only in PMHS #5. It consisted of a rupture of the ligamentum longitudinal antecevian between C5 and C6. This lesion resulted from the last test series which was performed without a head restraint. All other subjects showed no sign of a neck lesion. Ruptures of small vessels leading to a local hemorrhage during active blood circulation or damage to nerve fibers, as reported by Ortenberg, could not be detected during autopsy [17]. The results of the pressure measurement of the cerebrospinal fluid are shown in figure 24. This proves an extensive influence of the transducer location on the results.

Table 3: Location of pressure transducers in the PMHS tests.

<table>
<thead>
<tr>
<th>PMHS I.D.</th>
<th>Upper Transducer</th>
<th>Lower Transducer</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMHS9949#3</td>
<td>C2</td>
<td>C1</td>
</tr>
<tr>
<td>PMHS9943#3</td>
<td>cerebellum</td>
<td>C1</td>
</tr>
<tr>
<td>PMHS9943#4</td>
<td>oval foramen</td>
<td>C6</td>
</tr>
<tr>
<td>PMHS9943#5</td>
<td>C1</td>
<td>C6</td>
</tr>
</tbody>
</table>

![Figure 24: Pressure measurements of each subject.](image)

NIC calculations were performed for all tests and compared to parameters that might indicate soft tissue neck injuries [7]. The NIC was intended for predicting the pressure amplitude (p) at the C4 level of the spinal canal [3]. Since exact positioning of the transducers was difficult, NIC results are compared to pressure measurements that were closest to the C4 level (i.e., C2 transducer for PMHS #2, C1 for PMHS #3, and C6 for PMHS #4 and PMHS #5). In comparison, the NIC and pressure amplitude (p) showed an almost linear correlation (Figure 25). However, the pressure curves of PMHS #3 (very old female) differed significantly from the other subjects. Autopsy revealed nearly no cerebrospinal fluid in the spinal canal and the skull; therefore, results of the pressure measurement were not valid for this subject. Therefore, we propose positioning the transducers within the CSF space for further investigation. Due to the fact that the pressure measurement of PMHS #3 is not valid, these tests are removed in figure 25; the results of figure 24 were illustrated again in a single diagram (Figure 26) and a statistical analysis of the correlation was performed. The PMHS tests showed an absolute correlation of the NIC (NICabs) and the pressure which confirms the results of the tests with pigs (Figure 26). In addition, the time of appearance of the NIC and the pressure is of interest (Figure 27). The results show that NIC occurs often before the pressure amplitude (approximately 30 ms).
A comparison of pressure-time histories in series C (PMHS tests, Figure 24) with results of the experiments with pigs (Figure 28) led to similar results. A minimum pressure in the spinal canal would be observed at the maximum retraction phase of the cervical spine after approximately 100 ms. In tests with transducers in the skull, positive pressure amplitudes could be found in PMHS tests as well as in the pig experiments [5]. The amplitude maximum in experiments with pigs was found between 20 and 90 mm Hg (Figure 29), and in the PMHS tests, the maximum was found between 1 to 220 mm Hg (Figure 26). We suggest two possible reasons for this difference: Anatomical differences of pigs and humans, and differences in the experimental setup and test conditions.

As shown in figure 26, correlation between the NIC and pressure amplitude is almost linear ($r^2 = 0.79$) and of statistical significance ($p < 0.005$). Considering the limitations of these experiments, the NIC seems to be able to calculate the pressure magnitude using head and T1 accelerations with reasonable accuracy. However, timing of the NIC and the amplitude is different (Figure 27). Using the modified NIC calculation method, where the influence of the vertical component of the acceleration might delay the NIC histories, would be a possible explanation for these deviations. Another reason could be that the NIC was defined as the 3 ms maximum of the NIC (t) of the first 150 ms and not the NIC at the maximum retraction of the neck. During autopsy, it was not possible to investigate any damage to the nerve cells as demonstrated by Öntöngen et al.; no evidence can be provided as to which pressure amplitude level will induce nerve cell damage of the spinal ganglia [17]. First indications of threshold levels for soft tissue neck injuries concerning the NIC.
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are given [2, 7, 24]. In comparison with the living pig experiments, the absence of arterial blood pressure in the PMHS of our study did not allow any conclusions about the influence of blood and intracranial pressure to pressure effects of CSF during rear-impact motions. This influence could be of interest in future research. The modified NIC calculation method could also be a limitation, although comparisons showed no significant deviation [7]. The influence of the accelerometer location is shown in figures 30 and 31. The T1 acceleration is in agreement with the chest acceleration. It is strongly recommended to use low-capacity (50g range) accelerometers in order to improve the accuracy of the sensors. In tests where it was possible to adjust the x-direction of the T1 accelerometer approximately horizontally, the NIC-X and NIC-RES correlated well. Therefore, it can be concluded that the exact location of the T1 transducer does not extensively influence the results.

**Figure 30:** Comparison between chest and T1 (resultant) acceleration.

**Figure 31:** Comparison between NIC-X and NIC-RES.

The NIC is correlated with the velocity change of the vehicle (Figure 32). Furthermore, the head restraint position proved to be of importance. In all tests where a low NIC was found at higher velocity changes, the head was in initial contact to the head restraint (indicated as “NIC-Res, low distance” in Figure 32). Figure 33 illustrates the correlation of NIC with the crash pulse. The NIC depends on the maximum peak of acceleration as well as on the average acceleration of the sled. The correlation between NIC-RES and the maximum (3 ms) head and chest acceleration is also reasonable (Figure 34). Correlation of the NIC with the maximum head angular acceleration is shown in figure 35. The peak values for the head extension (rearward angular motion) are given on the left. It is very interesting that this maximum rearward acceleration appeared approximately at the same time as the maximum NIC. No significant extension movement could be seen. Correlation of the maximum forward angular head acceleration given on the right of figure 35, shows that it appeared in a later movement phase, when the extension motion is slowed down (due to the head restraint or anatomic restrictions of the cervical spine). The peaks of the forward angular acceleration were higher than those of the rearward angular acceleration. Peak values occurred approximately at the beginning and end of the extension motion.

The approximate neck torque at the occipital condyles was calculated according to the following formula:

\[ M_T = \hat{\varphi} \cdot \delta_T - m \cdot \overline{a}_l - m \cdot \overline{a}_y \cdot \delta_y \]

- \( M_T \): neck torque at occipital condyles [Nm]
- \( \hat{\varphi} \): vert. dist. from head CG to occ. cond. [m]
- \( m \): head mass [kg]
- \( \overline{a}_l \): linear head accel. in Z-direction [m/s^2]
- \( \delta_T \): linear head accel. in X-direction [m/s^2]
- \( \overline{a}_y \): angular head accel. about y-axis [rad/s^2]

Due to a lack of exact anthropometric data of the subjects (no x-rays), the anthropometric head properties of the 50th percentile Hybrid III dummy were used. According to this, the figures resulting from these calculations should be considered only approximate. Neck injury criteria and maximum neck torque for extension and flexion are well correlated (Figure 36). A very interesting fact is that the timing of the NIC and the maximum rearward angular acceleration of the head are similar. Comparing the timing of the NIC and the maximum neck moment (for extension) also proved to be true (Figure 37). Typically, the NIC occurs a little bit earlier (approximately 10 to 20 ms).

**Figure 32:** NIC and velocity change.

**Figure 33:** NIC and sled crash pulse.
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\[ M_T = \dot{\theta} \cdot I - \dot{m} \cdot a_x \cdot \delta_x - \dot{m} \cdot a_y \cdot \delta_y \]

- \( M_T \): neck torque at occipital condyles (Nm)
- \( \dot{\theta} \): angular head accel. about y-axis [rad/s²]
- \( I \): moment of inertia of head (y-axis) [kgm²]
- \( \dot{m} \): linear head accel. in Z-direction [m/s²]
- \( a_x \): linear head accel. in X-direction [m/s²]
- \( \delta_x \): vert. dist. from head CG to occ. cond. [m]
- \( \delta_y \): horiz. dist. head CG to occ. cond. [m]

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**Figure 32:** NIC and velocity change.

**Figure 33:** NIC and sled crash pulse.
4.3. Dummy Test Simulations

4.3.1 Velocity Change

According to literature, the velocity change (Delta-V) of the struck car is of major importance in rear impacts. Delta-V is commonly used to assess the impact severity and the risk of neck injuries in actual collisions. In this study, an idealized crash pulse (half sine shape, crash duration 100 ms) was used to investigate the influence. The amplitude of the sine pulse was increased resulting in different levels of velocity changes of the sled. These pulses were used to study the influence of Delta-V isolated from all other parameters. Results showed that the correlation between Delta-V and the NIC is almost parabolic and the NIC seems to represent the amount of kinetic energy loaded to the occupant (Figure 38).

4.3.2 Crash Duration

Given a constant Delta-V (16 kph) and the shape of the crash pulse (half sine), the crash duration characterizes the stiffness of the impact, i.e., the influence of the car structure stiffness of the colliding vehicles. The crash duration was varied between 25 and 150 ms. Results showed a decreasing NIC with increasing crash duration and an approximate 25 percent decrease of the NIC from a hard impact (50 ms crash duration) to a soft impact (125 ms) (Figure 39).
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![Figure 38: Influence of velocity change on NIC.](image1)

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![Figure 39: Influence of crash duration on NIC.](image2)
4.3.3 Horizontal Distance

The influence of the head restraint position was investigated by adjusting the head restraint to several positions ranging from direct contact to a maximum horizontal distance of 200 mm. Results of the simulations showed low NIC and neck loads whenever the head restraint was close to the head. By increasing the distance, the NIC and neck loads grew significantly until a horizontal distance of approximately 150 mm. Enlarging the gap to more than 150 mm showed no significant increase of the NIC whereas neck loads were still advancing Figure 40).

![Graph](image)

Figure 40: Influence of horizontal distance between head and head restraint.

4.3.4 Vertical Distance

With respect to the height of the head restraint position, some studies indicate an influence on the risk of cervical spine distortions. Therefore, the vertical height was adjusted to different positions and in one case the head restraint was completely removed. Results of the simulations showed an increase of the NIC when lowering the head restraint. However, the difference between NIC in a simulation with a high head restraint position (CG of head and CG of head restraint at same height) and a simulation where the head restraint was removed is only 10 percent (Figure 41). This, together with high NIC results in tests with a horizontal distance between head and head restraint of more than 150 mm, may be an explanation why head restraints did not show significant neck protection in collision statistics.

Due to the extension motion of the neck, on the other hand, neck loads especially the neck torque, increased drastically (up to 500%) when removing the head restraint (Figure 42). The maximum value of the upper neck torque during extension motion increased from approximately 20 up to 120 Nm. This result clearly showed that the NIC has been designed for assessing the retraction motion and not for predicting hyperextension injuries. This type of injury has to be assessed by other biomechanical criteria such as neck loads.

4.3.5 Seatback Stiffness

The influence of the recliner stiffness was investigated by varying the loading stiffness of the seatback, i.e., plastic deformations were not considered. Results showed a possible reduction of the NIC and neck loads by using a stiff recliner joint. The use of stiff seats is advantageous to seats with a low stiffness, because the head restraint does not move as far away from the head due to yielding of the seatback. According to this, contact of the head with the head restraint occurs early and thus limits the amount of retraction and extension motion. Energy absorption of the seatback by plastic deformation or a damping system was not investigated.

4.3.6 Correlation of the Neck Injury Criteria and Neck Loads

An almost linear correlation between the NIC and neck torque in the upper and lower neck was found when the head restraint was present (Figure 44). Due to hyperextension motion, neck torque increased drastically with a very low head restraint or without head restraints, i.e., an inefficient seat design.

![Graph](image)

Figure 41: Influence of vertical head restraint adjustment on NIC.

![Graph](image)

Figure 42: Influence of vertical head restraint adjustment on upper neck torque.

![Graph](image)

Figure 43: Influence of recliner joint stiffness on NIC.
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5. Summary

The NIC proposed by Bostrom et al. was applied to human subject and dummy tests [2]. The results were compared with other parameters. Human subject tests with volunteers gave results below the proposed injury level of NIC = 15. Minor complaints were reported in tests with a NIC of approximately 10. No complaints were reported in tests with a NIC below 8 and no long-term effects in volunteer tests occurred. A correlation was found between the NIC and velocity change, crush pulse, and head restraint position. In addition, the neck extension angle, head angular acceleration, and neck torque proved to be correlated with the NIC, although according to the pressure theory, the extension motion is not expected to be injurious. One PMHS test resulted in an injury of a subject due to hyperextension after removing the head restraint. Although this injury (rupture of ligament) is accompanied by a high NIC value (NIC-RES = 18.6) in this test, it is not related to classical neck distortion.

In order to avoid a recalculation of the correct x-components by analysis of the high-speed video, it proved to be necessary to adjust the local x-direction of the accelerometers accurately to the global x-coordinate system. The most efficient and reliable method to calculate NIC was to use the resultant head and torso accelerations (NIC-RES) instead of the C1 and T1 x-accelerations as proposed by Bostrom et al. (NIC-X). Additionally, we used the 3 ms maximum of the NIC-time history within the time interval of the expected occurrence of maximum head retraction (less than 150 ms) instead of the 50 mm head retraction for the NIC calculation. The deviations from the original NIC proposal were minor. In addition, the NIC calculated from resultant acceleration would be reasonable when applying it to other impact directions (frontal and side impact).

Four PMHS were equipped with accelerometers and pressure transducers in the spinal canal. Opposite to the experiments with pigs, a test setup reproducing realistic conditions for a rear impact collision was chosen. For this purpose, results of crash investigations and statistics were used. A linear correlation of the NIC (calculated from accelerometer readings) with the pressure amplitude (calculated similar to the experiments with pigs) was observed. However, the timing of appearance was slightly different. Furthermore, our results indicate a correlation of the NIC with the magnitude of pressure amplitudes of the CSF during whiplash motions of the tested PMHS. Due to the experimental setup it was not possible to detect any soft tissue neck injuries resulting from these pressure effects. We suggest the investigation of the influence of these effects on soft tissue neck injuries for the future.

According to the computer simulations in this study, in rear impact crashes where the head-neck extension motion is limited by a head restraint, the NIC is correlated with neck loads of the Hybrid III dummy (with the TRID neck). However, the NIC is not able to predict hyperextension injuries whenever the head restraint is very low or not present. The NIC is sensitive to Delta-V, acceleration pulse level, shape of crush pulse, head restraint position, and seatback properties. The Delta-V and the horizontal distance between head and head restraint influence the NIC and neck loads remarkably. However, the computer simulations in this study are limited. First, the biodiety of the Hybrid III (even when using the TRID neck) is limited; second, only one seat was investigated. Third, in spite of validation in sled tests, the accuracy of the mathematical model under extreme conditions, especially at high velocities where seatback collapsing is to be expected, is not proven.

6. Conclusion

The NIC explains dangerous impact conditions with respect to soft tissue neck injuries with acceptable accuracy. The NIC is correlated to impact parameters that were considered to be injurious like velocity change, crush pulse, head restraint position, or head and torso accelerations; furthermore, the head angular acceleration, head angular acceleration, and neck torque. In all experiments, a high NIC was always correlated to extensive relative motion of the head and neck. Results of the simulations suggest the use of the NIC as an indicator for the risk of soft tissue neck injuries in rear impact dummy testing, although the biodiety of the dummy is limited. With respect to possible neck injuries resulting from hyperextension, we recommend considering neck loads, forces, and moments in the upper and lower neck. However, it is still reasonable to measure neck loads and displacements to decide the possibility of hyperextension injuries. For future design of car seats, it is proposed to consider the effect of the NIC as well as neck loads and maximum spinal extension angles.

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References

Hybrid III and Hybrid III with a TRID Neck in Low-Speed Rear Impacts

Agnès Kim and Priya Prasadi

Abstract. This study compares the performance and biofidelity of a Hybrid III (Part 5728) and a Hybrid III with a TNO rear impact dummy neck in rear impacts. First, the anthropomorphic test devices were subjected to side-to-side rigid bench tests. Using human volunteer and cadaver data obtained as a baseline, the biofidelity of the anthropomorphic test devices is determined at Delta-Vs of 16 and 24 kph. The responses of both anthropomorphic test devices agreed fairly well with the available data. Therefore, the biofidelity of the dummies is adequate to predict headneck kinematics in rear impacts of these severities. The behaviors of the Hybrid III and TNO rear impact dummies are similar and no major differences were seen between their responses. Additionally, although the generated neck injury criterion curves were similar in shape, there were some differences between the two dummies. Next, the Hybrid III and TNO rear impact dummy were tested on production seats at Delta-Vs of 8 and 16 kph. The seats were chosen because they were low-speed frequency scores represented seats with an average, less than, and better than average frequency of neck injury claims. In these tests, the responses of the two anthropomorphic test devices were similar with no observed significant differences. For the Hybrid III and TNO rear impact dummy, the corrected lower neck extension moment predicted the trend in neck injury frequency scores at both test speeds. For the Hybrid III, the uncorrected lower neck extension moment and the NICD values also correlated well with the neck injury frequency ratings at both speeds. Considering only these three seats, the corrected lower neck extension moments and the NICD values correlated for the Hybrid III. When test results of other different production seats were added to the sample, the correlation between the C7-T1 extension moment and the NICD was no longer seen.

1. Introduction

In 1997, there were approximately 1.92 million rear-end motor vehicle collisions in the US with 32.8 percent resulting in injuries [9]. Most of the reported injuries were low-severity neck injuries, commonly known as whiplash injuries, soft-tissue injuries, cervical sprains, etc. This problem has not only heightened research efforts into possible injury mechanisms, but has also spurred the development of various anthropomorphic test devices (ATD) or their components specifically for use in low-speed rear impacts. Chalmers University of Technology introduced the rear impact dummy (RID) neck in the early 1990s and since then, has continued developing its prototype ATD with the current version the BioRID P3 [3,12]. TNO developed the TRID (TNO Rear Impact Dummy) neck and is also developing an ATD specifically for rear-end collisions [13]. In 1997, Prasad et al. presented data on the performance of three ATD configurations (Hybrid III, Hybrid II1 with RID neck, and TAD-50) in low-speed rear-impact collisions. The purpose of this paper is to present similar data again on the Hybrid III and on the Hybrid III with the TRID neck, evaluating their biofidelity in low-speed rear-end collisions. In addition, results from low-speed rear impact production seats with the Hybrid III and the TRID are also presented [10-11].