EVALUATION OF THE APPLICABILITY OF THE NECK INJURY CRITERION (NIC) IN REAR END IMPACTS ON THE BASIS OF HUMAN SUBJECT TESTS

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ABSTRACT

Over the last decades a large amount of studies have dealt with the phenomena of soft tissue injuries following rear end impacts. Up to now neither the injury mechanism has been identified nor an Injury Criterion has been established. Therefore it is very difficult to develop neck protecting seat systems and to evaluate their effectiveness.

Recently, Boström et al. have proposed a so called Neck Injury Criterion (NIC) which shall predict neck injuries following rear end impacts. This Criterion is based upon a hypothesis of Aldman. In principle, the theory claims pressure effects in the spinal canal to be responsible for soft tissue neck injuries. Svensson et al. have investigated the theory by performing tests with anaesthetized pigs. It was demonstrated that pressure peaks in the spinal canal occur at a certain phase of the movement of the cervical spine ("S-shape", "maximum retraction"). Also damage to the spinal ganglia was found in these tests and related to the pressure effect theory.

The purpose of this investigation is to apply the NIC Criterion on data measured in human subject tests. 70 volunteer tests and 28 cadaver tests were analyzed in order to investigate the movement of the human spine. The NIC was calculated for all these tests and compared to other data that could be of interest for neck injuries (e.g. angular acceleration of the head).

Out of this analysis it is concluded that the NIC in conjunction with other parameters is a useful indicator for the prediction of neck injuries following rear end impacts.

THE BASIS of this study is the Neck Injury Criterion (NIC) which is defined as following (Boström, 1996, 1997):

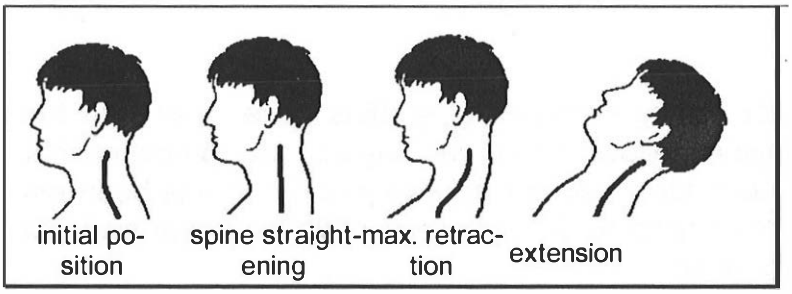
$$\boxed{NIC = a_{rel} \cdot 0.2 + v_{rel}^2} \text{ (tolerance level: 15 m}^2\text{/s}^2\text{)}$$

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

NIC is a Injury Criterion that is not validated on human beings but a result of a hydrodynamic model of the spinal canal during rear impact (Boström, 1996) using data of pig experiments (Svensson, 1993). Also the proposed tolerance level of 15 m^2/s^2 is only related to pigs. NIC shall be calculated at a certain phase of the neck movement, the so-called 's-shape' or 'maximum retraction'. Several studies (Ono, 1997; Svensson, 1993; Boström, 1996) indicate that this phase occurs in a rather early stage. Fig. 1 shows typical occupant kinematics. The chest of the occupant is accelerated by the seat back but the head stands still because almost always the head has no direct initial contact to the headrestraint and is therefore unrestrained. Even if the initial gap is very small, the headrestraint is moved away from the head because of the bending of the seat back that is loaded by the occupant. This results in a rearward translational movement of the head relative to the torso, causing flexion of the upper neck and extension of the lower neck. This 'retracting' movement is limited by the neck in a s-shape, followed by extension of the whole neck.

Fig. 1: Occupant kinematics during rear impact

the spine straightens from its initial position (lordosis); the translational movement continues until max. retraction is reached; finally the whole spine extends



METHODOLOGY

The Neck Injury Criterion was applied to human subject tests performed by the Institute for Mechanics. Boström et al. (1997) proposed to calculate NIC at 50mm of head retraction. In terms of accuracy it was very difficult to apply this method to the test series described in this study. Instead a 3ms-peak (using the same calculation routine as for the 3-ms head acceleration criterion) was detected and defined as NIC. This method proved to be more accurate and reliable. It is demonstrated that the maximum NIC occurs approx. at the same moment of time as the maximum retraction phase.

A total number of 70 sled tests with volunteers and 28 tests with post mortem human subjects (PMHS) were analyzed. The tests were performed using different types of car seats (standard car seats and prototype seats). Impact conditions were chosen that were considered to be most realistic for rear-end collisions. For this reason, crash pulses measured in real car crashes by accident data recorder (UDS™ by Mannesmann-Kienzle) were simulated on the sled.

SLED TESTS

VOLUNTEER SERIES A

The first test series included 36 sled tests with 12 different human subjects, divided into two groups. All tested subjects claimed to be healthy especially at spine region. A manual therapist checked the spine of the subjects by some simple tests. The first group consisted of 5 females and 1 male (body size approx. comparable to the 5th percentile female H-III Dummy), the second group of 6 male subjects (body size approx. comparable to the 50th percentile male H-III Dummy). A standard car seat was used, but some modifications were done: The seat back was fixed in order to prevent bending of the seat back and the headrestraint was replaced by a rather large prototype headrestraint, that was in initial contact to the head of the volunteer.

Fig. 2: Test setup series A

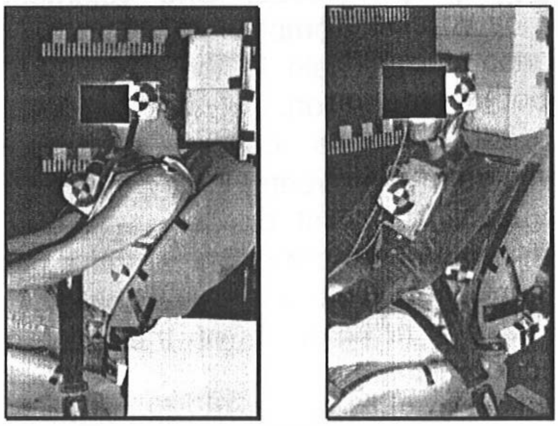
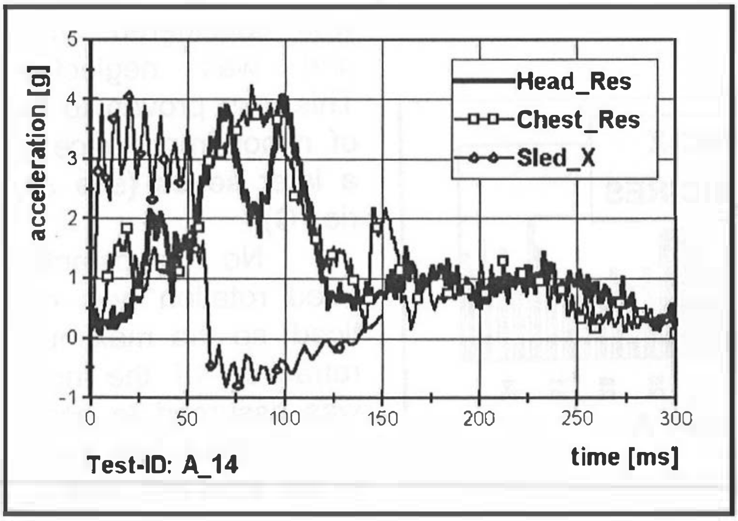


Fig. 3: Typical resultant accelerations in series A

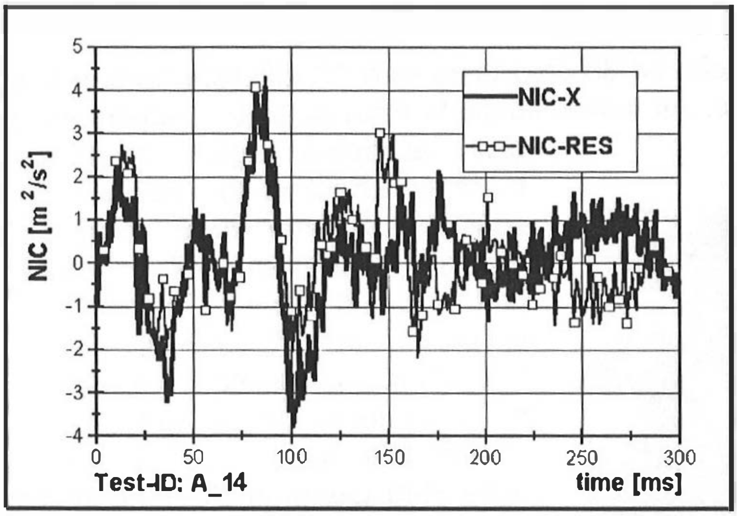


The head accelerometer was placed at the side of the head approx. at the height of the center of gravity. The torso accelerometer was located at the front of the chest approx. at the same height as the torso accelerometer of the Hybrid-III-Dummy.

Tests were performed at a rather moderate sled impact speed of approx. 5 to 5.5 km/h. The average crash pulse was between 2.5 and 3g, crash duration approx. 60 to 70 ms. This test series represents a rather hard pulse at low velocity change level (collision without vehicle damage).

Due to the low impact speed and the optimal adjusted headrestraint, injuries to the cervical spine should be very unlikely. For this reason this test series should result in NIC values far below the injury threshold. Fig. 3 shows results of one of these tests.

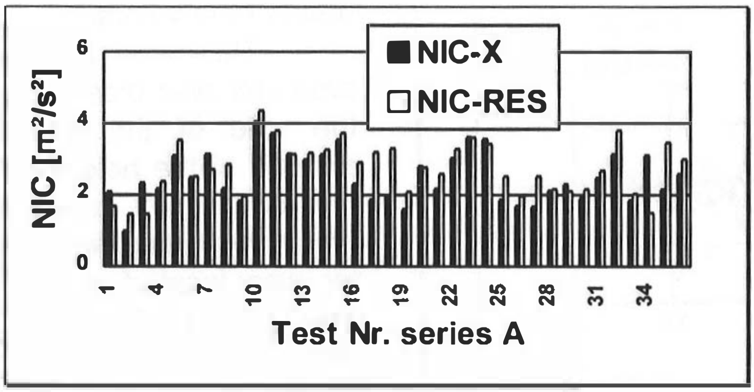
Fig. 4: Typical time-NIC history in series A



None of the volunteers complained about injuries in the region of the cervical spine after the tests. Also days after the test no neck pain was reported. One volunteer complained about minor pain in the lumbar region. Though this could be caused by other events, it was assumed that due to the rather hard sled pulse and the fixed seat back, the loading of the lumbar and thoracic spine was higher than in real life accidents.

The NIC time history was calculated for all 36 tests, using the formula described above. Though the formula for the NIC uses the accelerations of C1 and T1, the results of the chest accelerometer were used as T1 accelerations and the head acceleration as C1 acceleration. The influence of exact location was considered small and was neglected. This was proven to be of minor importance in a later series (see series C).

Fig. 5: NIC results of series A



No substantial head rotation was noticed, so the maximum retraction of the neck was assumed to occur at the maximum value of the time-NIC history. At all tests NIC starts to decrease at a certain moment of time (typically 80 to 120 ms, depending on impact conditions). Fig. 4 illustrates a typical result of NIC calculations in series A. A 3ms-peak was calculated and defined as NIC. The NIC was calculated using only the X-components of the head and chest accelerations and defined as NIC-X. In order to apply the NIC to other test series where it is not possible to extract the X-component from the resultant accelerations (see series B), the NIC was also calculated using resultant head and torso accelerations (NIC-RES). The results are quite similar, because the X-component is the dominant direction and the NIC is calculated at maximum retraction where influences from head rotation are negligible. As illustrated in Fig. 5, NIC values of approx. 2 m^2/s^2 to 4 m^2/s^2, far below the proposed injury threshold, were observed. The comparison of NIC-X and NIC-RES is satisfying (r=0.71). Larger deviations can be explained by attachment of the accelerometers (see series C).

VOLUNTEER SERIES B

For a test series (34 tests) performed in 1995 (Eichberger, 1996), the NIC was recalculated. In this series, tests were performed on different standard car seats in realistic impact conditions. A moderate crash pulse (Crash duration approx. 120ms) was chosen. The velocity change of the sled was 11 km/h for the first 17 tests and 9 km/h for the last 17 tests.

Fig. 6: Typical NIC-time history (series B)

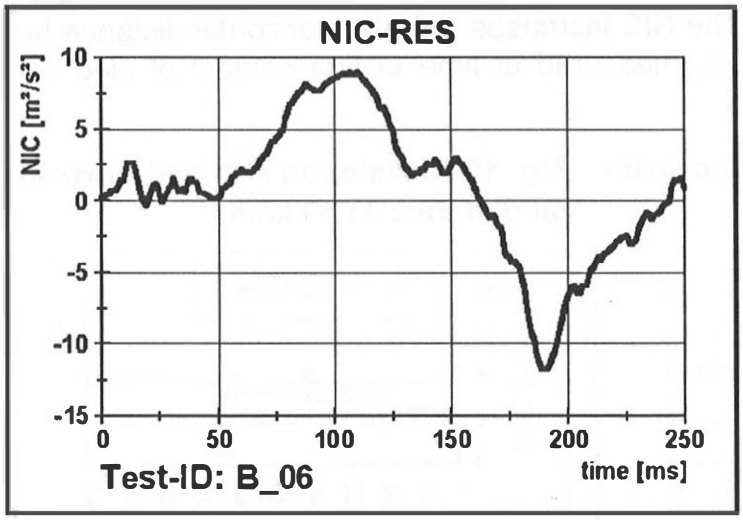
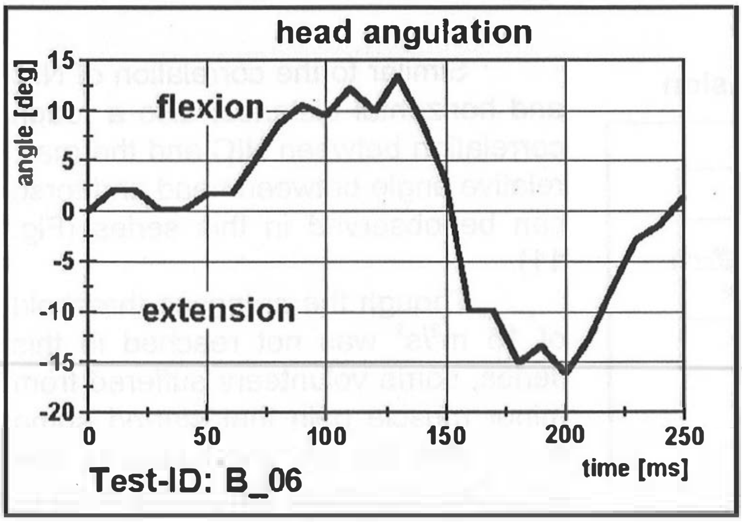


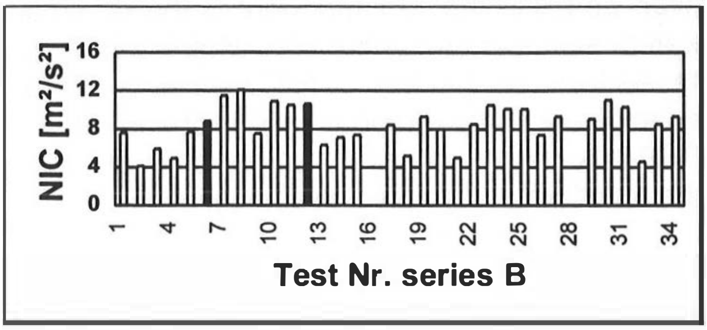
Fig. 7: Typical head angle-time history (series B)
relative angle between head and chest target



NIC calculations were performed similar to series A, using 3ms maximum of the time-NIC histories. It was only possible to calculate NIC based on resultant accelerations (NIC-RES) and not on the X-components (NIC-X).

It can bee seen from Fig. 6 and Fig. 7 that maximum NIC occurs approx. at the moment of time when the initial flexion of the head is complete and

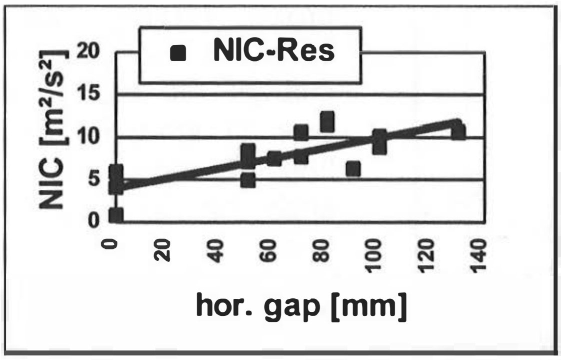
Fig. 8: NIC results of series B
Black bars indicate cervical complaints



extension movement starts.

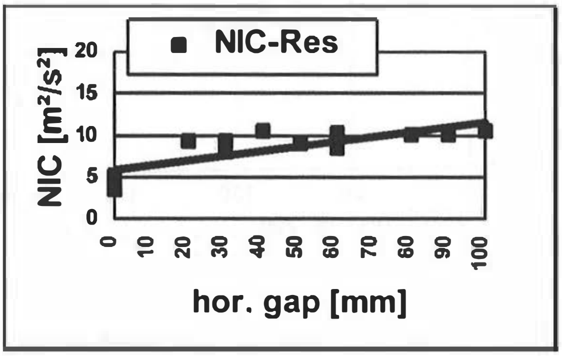
In Fig. 8 the results of NIC calculation (same method as series A) can be observed. Compared to series A, the NIC values are increased significantly, especially when the initial gap between head and headrestraint is larger. The correlation between NIC and the horizontal distance between head and the headrestraint is illustrated in Fig. 9 and Fig. 10. The NIC increases with the horizontal distance between head and headrestraint, measured at time of first contact of sled and deceleration element.

Fig. 9: Correlation NIC and horizontal distance ΔV=11 km/h



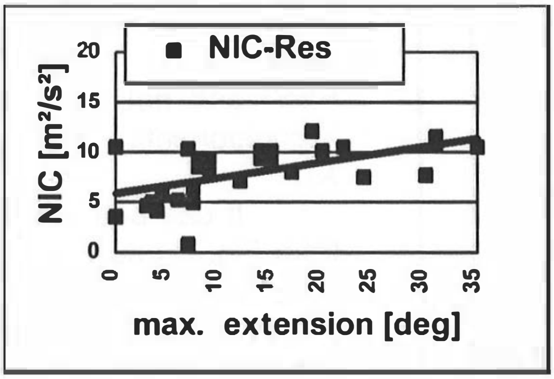
n=15; r=0.75 (r^2=0.57; p=0.000641)

Fig. 10 Correlation NIC and horizontal distance ΔV=9 km/h



n=15; r=0.81 (r^2=0.66; p=0.000080)

Fig. 11: NIC and head extension



n=30; r=0.56; (r^2=0.32; p=0.000908)

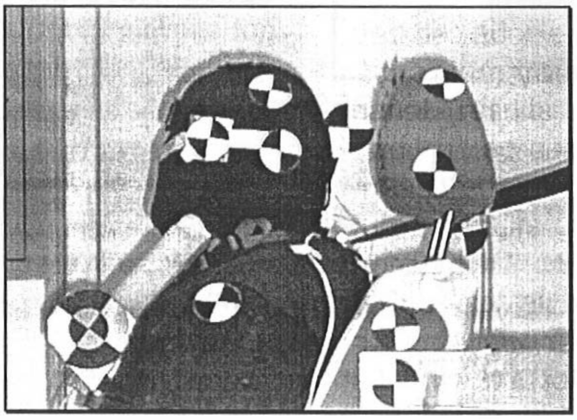
Similar to the correlation of NIC and horizontal distance, also a rough correlation between NIC and the max. relative angle between head and torso can be observed in this series (Fig. 11).

Though the proposed threshold of 15 m^2/s^2 was not reached in this series, some volunteers suffered from 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 WAD for 3 weeks (muscle pain, restricted range of motion of the spine, headache). The injury could not be diagnosed objectively (MRI, X-ray, blood test). A closer look on the volunteers anamnese showed that he could have had minor pretraumatic changes of the spine. For this reason the WAD-threshold of this subject could have been lower compared to an average human being.

PMHS SERIES C

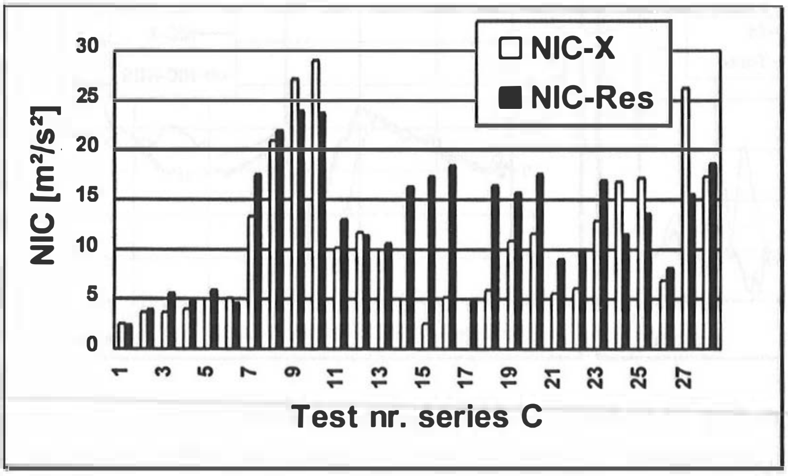
A total number of 28 PMHS experiments were performed. 5 subjects (4 males, 1 female) were seated in a seat of a very common car model. Tests were performed at impact speeds of approx. 9 and 15 km/h. The subjects were instrumented with triaxial accelerometers on the head (approx. center of gravity) and on the chest (comparable to series A and B); 1 two-dimensional accelerometer at the height of T1, 1 angular accelerometer at the head, 2 pressure transducers in the spinal canal (results not presented here) and a so-called spine-band for measurement of the spinal curvature (results also not presented here). NIC calculations were performed for all tests. Also the High-Speed Video was analyzed.

Fig. 12: Setup of test series C



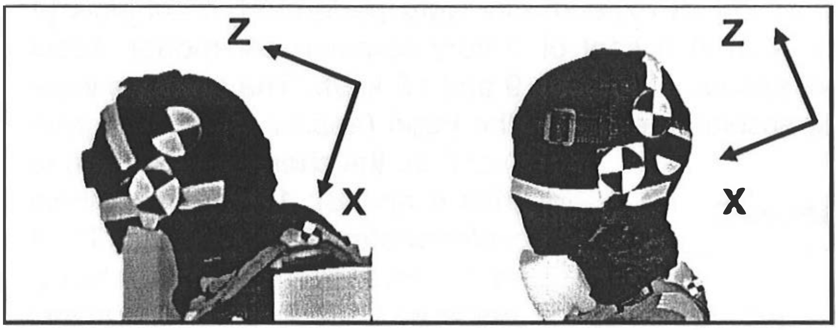
The following parameters were varied: impact speed, crash pulse, seat back inclination, headrestraint position. The autopsy showed that four subjects remained uninjured, one subject was injured in a test where the headrestraint was removed and a hyperextension injury occurred (minor ligamentous damage to the cervical spine). NIC was calculated for all tests. In this series, the acceleration of T1 was measured and therefore used for calculation of NIC. Only the X-component of T1 was used. For comparison reasons, NIC-RES was also calculated similar to the volunteer tests as described above (using the resultant chest and resultant head acceleration).

Fig. 13: Results of NIC calculation in PMHS tests



A correlation between NIC-X and NIC-RES (r= 0.69) exists, but in some tests (especially tests C_14 to C_17) greater differences occur (Fig. 13). This fact can be explained by the attachment of the sensor on the subject: The cervical spine of the subject (elderly female) was bent forward, resulting in a wrong X direction of the sensor (Fig. 14, left). For comparison, another subject (young male) is illustrated in Fig. 14 (right) also. This angular error could not be recalculated because the quality of the High-Speed Video did not allow a Video Analysis of the T1 target. For this reason it is clear that only the NIC-RES values are defined correctly and are thus discussed below.

Fig. 14: Mounting of T1 transducer



The influence of the accelerometer location can be seen in Fig. 15 and Fig. 16. The acceleration of T1 and chest are quite the same. It is strongly recommended to use low capacity (50g range) accelerometers in order to improve the accuracy of the sensors. In tests where the X-direction of the T1 accelerometer could be adjusted approx. horizontally, the NIC-X and NIC-RES correspond well. Therefore it can be concluded that the exact location of the T1 transducer does not influence the results extensively.

Fig. 15: Comparison between chest and T1 (resultant) acceleration

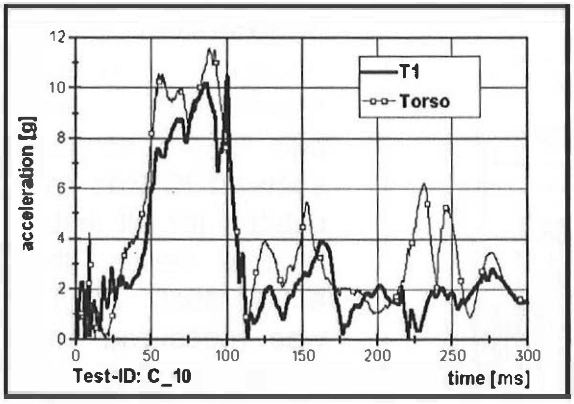
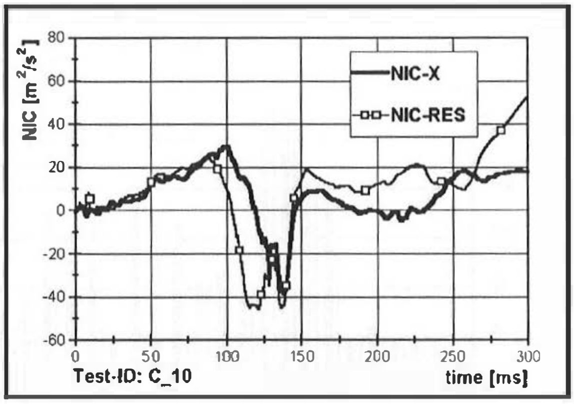
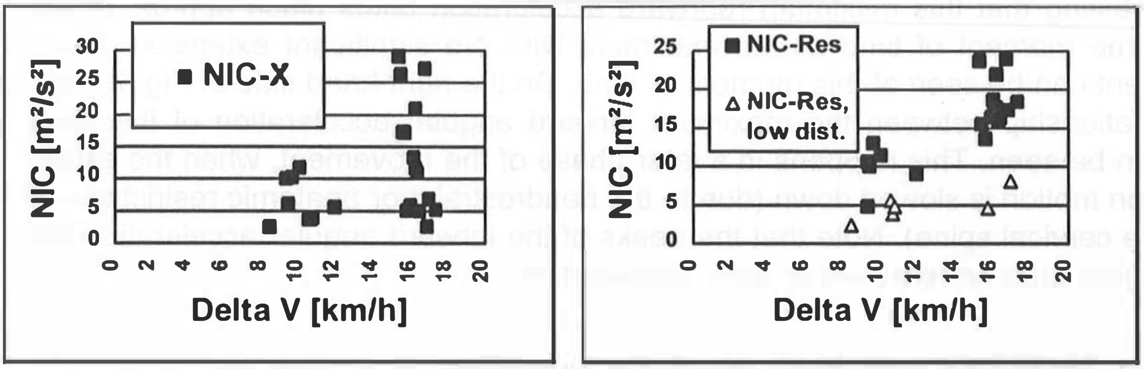


Fig. 16: Comparison between NIC-X and NIC-RES



As it can be seen in Fig. 17, NIC is roughly correlated to the velocity change of the vehicle. The correlation between Delta V and NIC-RES is better than for NIC-X. This can simply be explained by attachment of the sensor (see above). Furthermore the headrestraint position is important: In all tests were a low NIC could be observed at higher velocity changes, the head was in initial contact to the headrestraint (indicated as "NIC-Res, low distance" in Fig. 17).

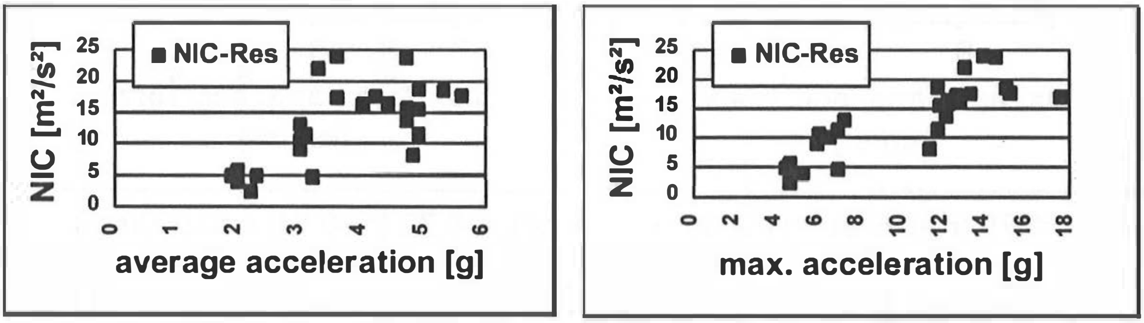
Fig. 17: NIC and velocity change



n=24; r=0.36 (r^2=0.13; p=0.086554)   n=25, r=0.70 (r^2=0.49; p=0.000055)

Fig. 18 illustrates the correlation of NIC and crash pulse. NIC depends on the maximum peak value of the acceleration as well as the average acceleration of the sled.

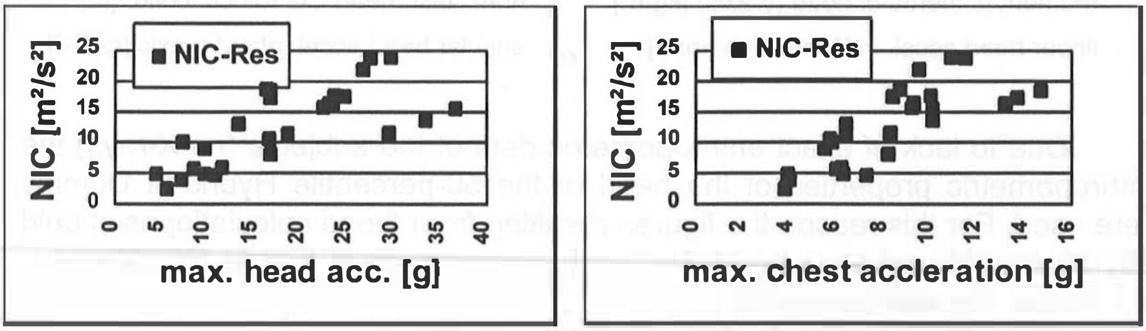
Fig. 18: NIC and sled crash pulse



n=26; r=0.66 (r^2=0.44; p=0.000138)   n=27; r=0.85 (r^2=0.73; p<<)

The correlation between NIC-RES and the max. (3ms) head and chest acceleration is also reasonable.

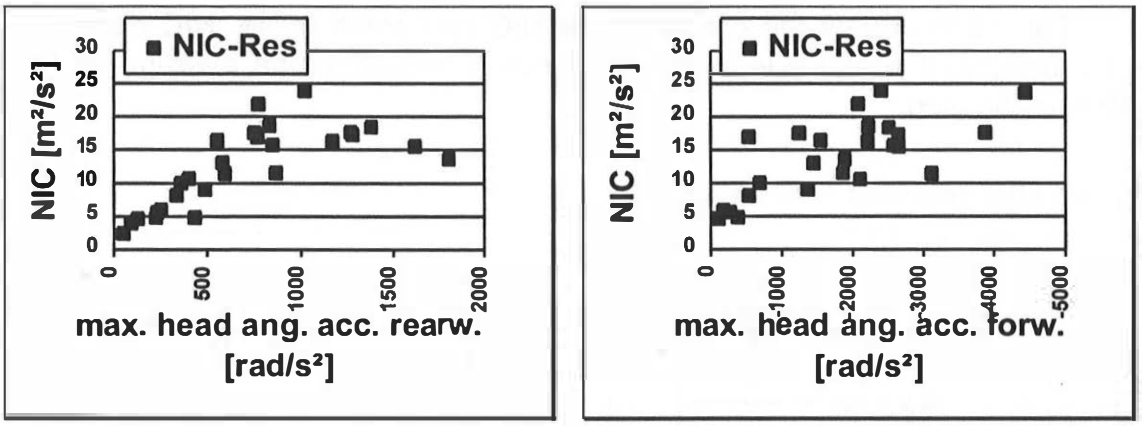
Fig. 19: NIC and max. head/chest acceleration



n=27; r=0.74 (r^2=0.55; p=0.000003)   n=27; 0.78 (r^2=0.61; p<<)

The correlation of NIC to the maximum head angular acceleration can be observed in Fig. 20. On the left hand side of the diagram the peak values for the extension (rearward angular motion) of the head is illustrated. It is very interesting that this maximum rearward acceleration takes place approx. at the same moment of time as the maximum NIC. No significant extension movement can be seen at this moment of time. On the right hand side of Fig. 20, the relationship between the maximum forward angular acceleration of the head can be seen. This happens in a later phase of the movement, when the extension motion is slowed down (due to the headrestraint or anatomic restrictions of the cervical spine). Note that the peaks of the forward angular acceleration are higher than for rearward angular acceleration.

Fig. 20: NIC vs. max. head angular acceleration
peak values occurred approx. at beginning and end of the extension motion



n=28; r=0.75 (r^2=0.57; p=0.000001) n=25; r=0,72 (r^2=0.51; p=0.000025)

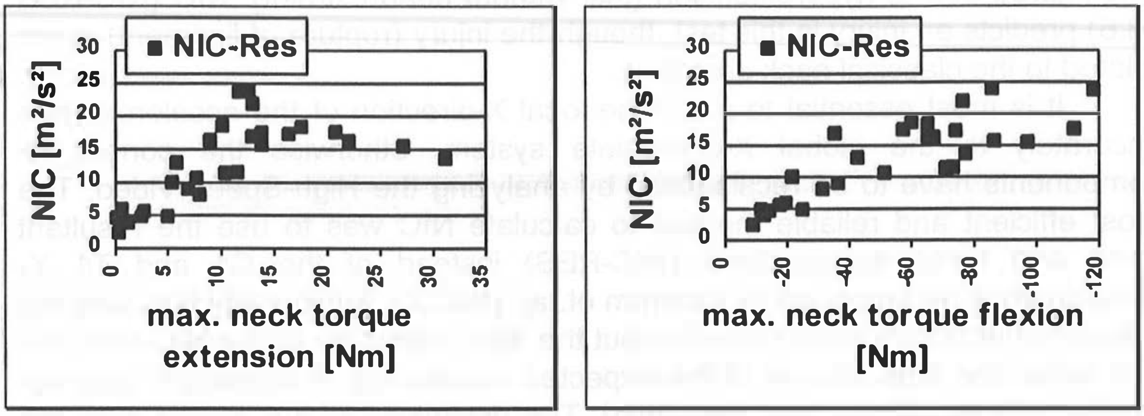
In addition, the approximated neck torque at the occipital condyles was calculated according to the following formula:

$$M_y = \dot{\omega} \cdot I - m \cdot a_x \cdot \delta_z - m \cdot a_z \cdot \delta_x$$

M_y	neck torque at occipital condyles [Nm]	δ_z	vert. dist. from head CG to occ. cond. [m]
m	head mass [kg]	a_z	linear head accel.in Z-direction [m/s²]
I	moment of inertia of head (y-axis) [kgm²]	δ_x	horiz. dist. head CG to occ. cond. [m]
a_x	linear head accel. in X-direction [m/s²]	$\dot{\omega}$	angular head accel. about y-axis [rad/s²]

Due to lack of exact anthropometric data of the subjects (no X-rays) the anthropometric properties of the head of the 50-percentile Hybrid-III Dummy were used. For this reason the figures resulting from these calculatiopns should only be considered as a trend. NIC and maximum neck torque for extension and flexion are well correlated (Fig. 21).

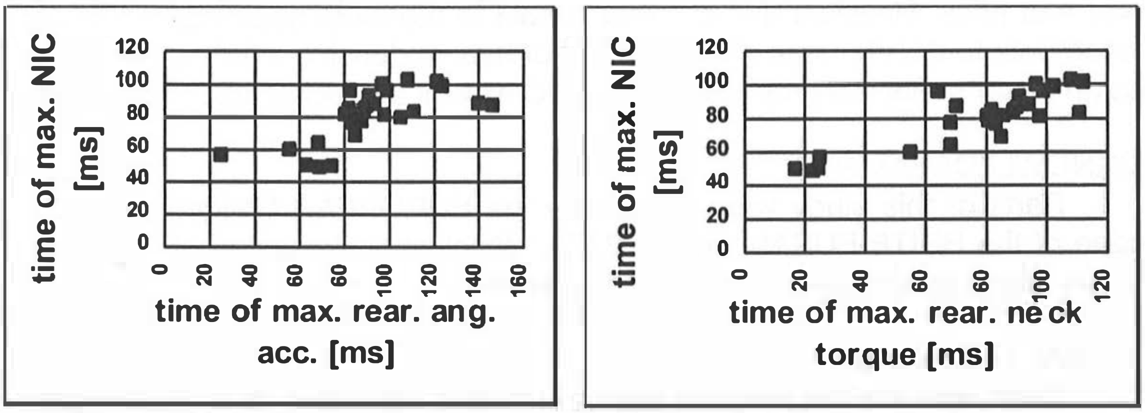
Fig. 21: NIC vs. max. neck torque (occ. cond.)



n=28; r=0.64 ($r^2=0.41$; p=0.000151) n=28; 0.84 ($r^2=0.70$; p<<)

A very interesting fact is that the timing of the NIC and the maximum rearward angular acceleration of the head is very similar. This is also true if the timing between NIC and neck moment (for extension) is compared (Fig. 22). Typically, NIC occurs a little bit earlier (approx. 10 to 20 ms).

Fig. 22: Timing of NIC, head angular acceleration and neck torque



n=28; r=0.69 ($r^2=0.47$; p=0.000023) n=28; r=0.86 ($r^2=0.75$; p<<)

CONCLUSIONS

The NIC proposed by Boström et al. (Boström, 1996) was applied to human subject tests. The results were compared to other parameters. Human subjects tests with volunteers were below the proposed injury level of 15. Minor complaints were reported in tests with a NIC of approx. 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, crash pulse and headrestraint position. Also neck extension angle, head angular acceleration and neck torque is correlated to NIC, though the pressure theory does not claim the extension motion to be injurious. In one PMHS test a subject was injured due to hyperextension (test without headrestraint). NIC (NIC-RES 18.6) predicts an injury in this test, though the injury (rupture of ligament) is not related to the classical neck distortion.

It is most essential to place the local X-direction of the accelerometers accurately to the global X-coordinate system, otherwise the correct X-components have to be recalculated by analyzing the High-Speed-Video. 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 Boström et. al. (NIC-X). Additionally NIC was not calculated at 50mm head retraction but the 3ms maximum of the NIC-time history within the time interval of the expected occurrence of maximum head retraction (80 to 120ms) was calculated. The deviations of the results from the original NIC proposal are minor. NIC calculated from resultant acceleration is also more reasonable when applying it to other impact directions (frontal and side impact).

Out of this study it is concluded that the NIC predicts dangerous impact conditions with respect to soft tissue neck injuries with acceptable accuracy. NIC is correlated to impact parameters that were considered injurious like velocity change, crash pulse, headrestraint position, head and torso accelerations, head angulation, head angular acceleration and neck torque. In all investigated tests a high NIC is always related to extensive relative motion between head and neck. However it is still reasonable to measure neck loads and displacements to decide upon the possibility of hyperextension injuries. For future design of car seats it is proposed to use NIC as well as neck loads.

ACKNOWLEDGEMENTS

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ETHICAL REMARKS

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 obtainment, treatment, anonymity and disposal of postmortem human subjects.

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