AN EVALUATION OF EXISTING AND PROPOSED INJURY CRITERIA WITH VARIOUS DUMMIES TO DETERMINE THEIR ABILITY TO PREDICT THE LEVELS OF SOFT TISSUE NECK INJURY SEEN IN REAL WORLD ACCIDENTS

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ABSTRACT

This study investigated the ability of a number of different dummies and candidate injury criteria to correlate to retrospective real world data. Vehicles were chosen for their rate of soft tissue neck injury (STNI) claims in the field. The front seats of the selected vehicles were mounted on a HyGe sled and loaded with instrumented dummies. They were then subjected to a simulated rear impact using a bi-modal pulse of 16km/h delta V.

Correlation of the measured and calculated test results, especially the NIC, Lower Neck Extension Moment ($-My_{lower}$) and the newly proposed LNL-index (Lower Neck Load index), to claims frequency was investigated. In conclusion, good correlation with claims data was found using $-My_{lower}$ with the Hybrid III and RID2 dummies. Similarly good correlation was found with the LNL-index when applied with the RID2 dummy. It should be noted that $-My_{lower}$ and the LNL-index could not be investigated on the BioRID II, since no lower neck load-cell was available for this dummy at the time of testing.

INTRODUCTION

The three kinematic phases of low speed rear impact

Low speed rear impacts are common occurrences, especially in areas of high traffic density. Injuries in these impacts are not life-threatening, and are generally classified as AIS 1 neck injuries, and are also referred to as Soft Tissue Neck Injuries. The majority of these injuries resolve in less than 10 days, but a smaller portion can persist longer than six weeks. Due to the large number of rear impacts, the total number of these STNIs, based on insurance claim frequencies, is large. Various injury mechanisms have been hypothesized, but none proven due to a lack of diagnostic tools. However, a general consensus is building in the scientific community that the complex kinematics of the head and neck before, during and after head-to-head restraint contact in rear impacts result in distortions of the neck. These distortions may be different in the three phases of impact, therefore, each phase may have different injury mechanisms.

The first phase is generally described by the S-shape deformation of the neck [Penning 1994, Gauner 1997]. In this phase, there is translation of the head relative to the thorax with very little rotation of the head forcing the upper part of the neck into flexion and the lower part into extension. The second phase is the phase of maximum extension. In this phase, the occupant's lower neck is in extension while the upper neck may be in flexion or extension depending on the seat geometry and the size/posture of the occupant. The third phase involves rebound of the torso from the seat back. The rebounding torso can interact with the seat belts and produce deformations in the neck due to inertial loading [Muser et al. 2000].

It is important to find appropriate criteria for each of the phases of impact since the injury mechanisms are quite different.

It is difficult to attribute real world STNI to a particular phase in the impact event. Retrospective studies can generally answer the question of whether an injury has occurred or not but cannot give any indication about the kinematic phase in which the injury occured. This makes it difficult to assign an appropriate weighting factor to each of the individual phases for an overall assessment.

To date, several criteria have been proposed to assess the risk of STNI in low speed rear impacts: NIC [Boström et al. 1996] for use in the first phase, Nkm [Schmitt et al. 2000] and -My_{lower} [Prasad et al. 2001] for the second phase**,** and rebound velocity [Muser et al. 2000] for the third phase.

Currently, NIC is the most widely used criterion for assessing low-intensity neck loading. It assumes that hydro-dynamic pressure changes in the spinal canal are produced as a result of the S-shape motion in the early stage of a rear-end impact and is the injury mechanism in this stage [Boström et al. 1996]. It is assumed that there is a correlation between the pressure, and the relative acceleration and velocity of the occipital condyles relative to the first thoracic vertebra. However, limitations exist [cf. e.g. Boström et al. 2000, Muser et al. 2000] suggesting that only values obtained within approximately the first 150 ms of the crash are reasonable. Furthermore, the NIC peaks before or at initial head contact with the head restraint**;** much before maximum extension of the neck. Therefore, NIC can only be used to assess injury potential during the first phase.

As an assessment criterion for the second phase Nkm has been proposed. Nkm is a linear addition of shear forces and moments acting on the upper cervical spine. However, Prasad and Kim [1997] have reported that upper neck moments are relatively insensitive to seat design and impact severity with seats equipped with head restraints. Recent reports have suggested that a different mechanism of injury may be the major contributing part in STNI. [Yoganandan 2001, Grauner et. al. 1997, Ono et al 2001] In short, it has been suggested that a characteristic kinematic of the lower cervical spinal joints may result in a risk of local injury in the facet joints leading to pain. Based on these findings and earlier analysis of tests conducted with the Hybrid III dummy, it was proposed that the extension moments acting at the lower cervical or upper thoracic spine (-My_{lower}) should be used as a criterion for evaluating injury potential in the second phase [Prasad 2001 IRCOBI]. However, this criterion takes account of only one of, potentially, several forces and moments generated in the neck during impact. Therefore, a new criterion has been proposed as part of this study for further evaluation (see below).

As an assessment criterion for the third phase of impact, the rebound velocity has been proposed. This criterion measures the velocity of the head relative to the sled at the moment when the head first returns to or passes through the position it was in at time zero (prior to impact). This criterion can only be correctly measured by film analysis**.** This is considered unsuitable for routine test purposes. A simple integration of the head accelerometer signal is flawed due to the rotation of the sensor during the movement of the seatback and the extension of the head and neck that generally occurs in the second phase. Furthermore, the rebound velocity, in the biomechanical sense, is not an injury criterion, as an injury cannot occur due to a body part travelling at a given speed but only by rapidly stopping it from doing so. Therefore, the forces acting on the body during the deceleration caused by the belt system should be measured to assess this phase.

Proposed criterion LNL-index

Based on fundamental mechanical knowledge, the risk of damage to a joint in an impact scenario is largest if moments and forces are acting simultaneously. During the extension phase of a low speed rear impact with no oblique component there will generally be a negative moment of the lower neck, combined with positive shear and tension of the lower neck (using SAE J211/2 conventions). Therefore, it seems reasonable to estimate the effect of combined loading on stresses developed in the cervical vertebrae. This was originally proposed by Prasad and Daniel [1984], further refined by Mertz and Prasad [2000] and forms the basis for Nij currently used in the FMVSS208 for estimating the risk of serious neck injuries in frontal and rear impacts. However, only axial forces and bending moments were used. In the current study, the proposed Lower Neck Load Index (LNL), shown in Equation 1, combines the

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Formula to calculate LNL

Equation 1

$$LNL - index(t) = \left| \frac{\sqrt{My_{lower}(t)^2 + Mx_{lower}(t)^2}}{C_{moment}} \right| + \left| \frac{\sqrt{Fx_{lower}(t)^2 + Fy_{lower}(t)^2}}{C_{shear}} \right| + \left| \frac{Fz_{lower}(t)}{C_{tension}} \right|$$

three force components and two of the moment components measured at the base of the neck.

It is worth making a few comments on the construction of this formula. First, current research activities focus mainly on impacts that do not have any oblique element. However, future research might call for dummies and criteria that are also capable of addressing other kind of impact scenarios. Therefore, the equation presented also takes into account the lateral forces, Fy, and the moments along the anterior-posterior axis, Mx, even if these components are small in the tests reported in this paper.

The criterion presented here is independent of polarity, thereby avoiding any user error of sign convention when measuring the neck loads.

When defining the intercept values for a combined loading assessment, the ratio between the individual components is important. Whereas the values used for normalisation could, if needed, be corrected indirectly by adjusting the threshold value, the ratio implies more fundamental aspects and should therefore be chosen very carefully. Cadaveric research has shown that the inter-vertebral disk is more resistant to axial loading but comparatively less so in shear [Lin et. al. 1978]. Therefore the shear component should be rated higher. Based on the findings of previous research [Prasad et al. 2001] the extension moments are of high importance and therefore also have a high weighting (see Table 1).

Generally the quantitative results of LNL are dependent on the dummy used. Particularly when using the Hybrid III dummy the measured bending moments are higher than the ones measured with the RID2. This is related to the much softer neck design of the RID2. Therefore, when defining performance thresholds, the dummy used is an important consideration.

Table 1

Proposed LNL-index intercept values

C-moment	C-shear	C-tension
15	250	900

A correction for the moments measured at the RID2 load-cell accounting for the offset of the centre of the cell to the centre of the T1 vertebra was not used as this effect is considered small. For the Hybrid III the offset between actual T1 location and the load cell measurement point is large. Therefore, the moments measured on Hybrid III have been corrected using the standard formula (see Equation 2).

METHOD

Review of injury criteria by means of retrospective study

In order to evaluate the currently proposed evaluation criteria, a sled test series was conducted (in co-operation with Johnson Controls, Inc. Germany) using a retrospective approach. We believe that any evaluation testing should reflect the reported risk in the field. Therefore, front seats that are from vehicles that have known poor field performance should also have a poor score when tested and vice versa. For this purpose three seats have been chosen. One seat was from a vehicle identified by make and model as a poor performer (seat A (n=79)) based on GDV insurance claim data from 2000. A second seat with acceptable performance (seat B (n=152)) and a third seat from a vehicle with good performance (seat C (n=96)) were also tested. All vehicles are of contemporary design levels, of the same class and are expected to appeal to comparable demographics. In this statistic (n) refers to all insurance claims as a result of rear impacts for the specific model in the GDV database.

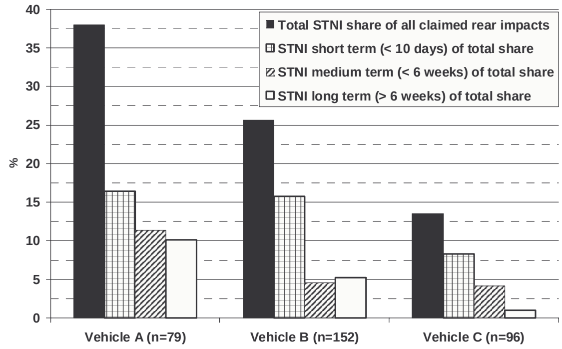
Formula to correct lower neck moment for Hybrid III

Equation 2

$$My_{lower\ corrected} = My + (0.0282 \cdot Fx) + (0.0502 \cdot Fz)$$

Figure 1

Share of STNI reported for Vehicle A-C 100% = all recorded rear impacts for that vehicle. Reports from 2000

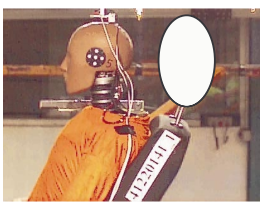


Additionally, a version of seat B was tested that is equipped with an active anti whiplash protection system. Systems of this design type have recently been reported to reduce injury by 43% on a statistically relevant sample size. [IIHS 2002]. Therefore, it could be expected that seat B-active should have a significantly improved performance over the standard seat B.

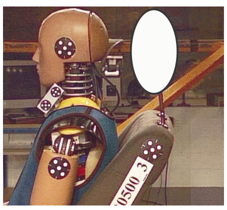
The seats were mounted on a HyGe sled. All seats were adjusted according to the FMVSS208 procedure (mid track, backrest angle of 25deg and headrest fully up) and tested with the Hybrid III, BioRID II and the RID2 dummies (see Figure 2).

Figure 2

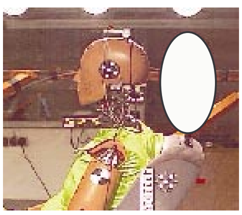
Images of the test setup



Hybrid III seated in seat A



RID2 seated in seat B

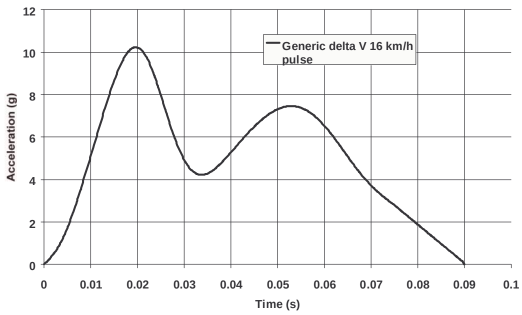


BioRID II seated in seat C

The most up to date seating procedure known at the time was used for the BioRID II and RID2 dummy. The initial horizontal distance between the head of each dummy and the front of the head-restraint was documented (see Appendix Table 2). Subsequently, the seats were subjected to simulated rear impacts using the previously reported bi-modal pulse [Heitplatz et al. 2002] shown in Figure 3.

Figure 3

Generic delta V 16 km/h acceleration pulse



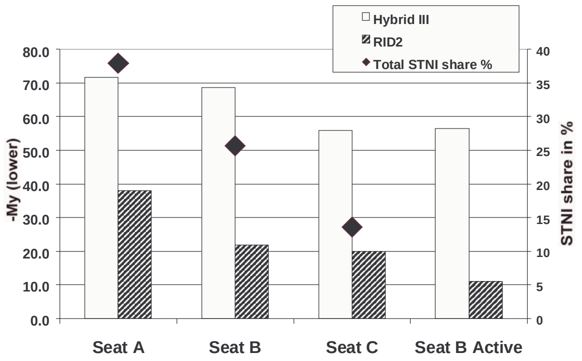
RESULTS AND DISCUSSION

Lower Neck Extension Moment (-My)

The peak lower neck extension moments (-My) measured in the tests are shown in Figure 4. Note that the moments for the Hybrid III have been corrected for the offset of the load cell to T1 using the standard equation (see Equation 2). The moments for the RID2 have not been corrected. The results of lower -My for this study correlate well to the number of real world claims regardless of the dummy used. It should be noted that the results with the Hybrid III neck show substantially higher values than those from the RID2. This is easily explained as the Hybrid III neck is stiffer and therefore, the resultant moments acting on the lower neck load cell are higher for a similar amount of extension of the neck. Comparing the lower neck moments of the different dummy designs, it can be shown that the qualitative correlation of measured results and claims frequency is good for both dummies. However, the quantitative differences between claims frequency and neck moment appears to correlate better with the RID2 dummy. The active seat design (B-Active) reduced -My by 18% with the Hybrid III and 49% with the RID2 compared to a similar seat without this system (B). However, it should be noted that the initial distance between the head and head restraint was not the same in tests with seat B and seat B-Active when tested with the RID2 dummy (Table 2 in the Appendix). The initial distance is known to affect head/neck responses.

Figure 4

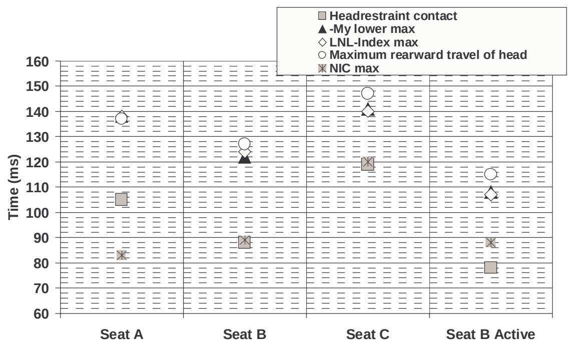
Results for –My lower



Looking at the timing of individual events during the crash sequence (see Figure 5), the lower neck extension moments peak around the same time as the head and neck reach maximum extension. Therefore, lower neck extension moments can be regarded as assessment criteria for phase 2.

Figure 5

Event timing

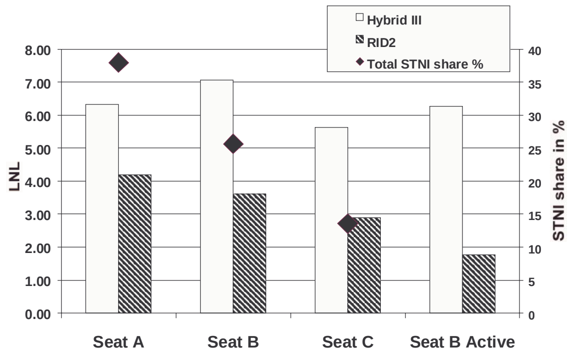


Lower Neck Load Index (LNL-index)

When using the RID2 dummy and the LNL index there is good qualitative correlation to the insurance claims frequency data (see Figure 6). Furthermore, there is a significant 26% reduction in the LNL index for seat C compared to seat A. It is notable that the active head restraint reduced the LNL index by 51%. This correlates well to the recently published insurance data that has shown a reduction of 43% in insurance claims for vehicles where this type of system was installed [IIHS 2002].

Figure 6

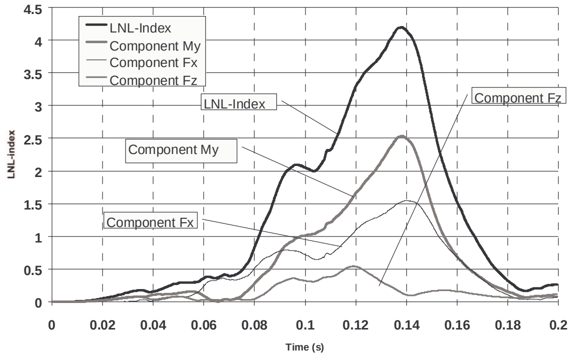
Results for LNL



An example time domain plot of the LNL index is shown in Figure 7. In this example, typical of most cases, the lower neck moments followed by the shear forces gives the largest contribution to the overall value of the LNL index. Normally the contribution of the tensile forces to the LNL is quite small for seats equipped with head restraints.

Figure 7

LNL-Index diagram for seat A and RID2 with detailed representation of the My, Fx and Fz components

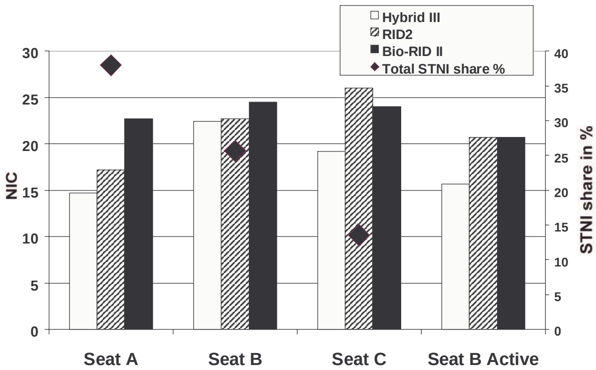


Other Criteria

In this limited study, the NIC has shown a negative correlation to the frequency of insurance STNI claims (see Figure 8). Though limited testing has been conducted in this series, indications are that the NIC, if used alone as an injury criteria for evaluating STNI's, can be misleading.

Figure 8

Results for NIC

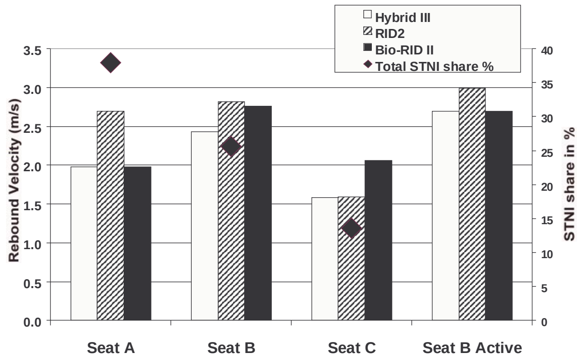


The upper neck loads have also been recorded during testing. However, there have been problems with the recorded data, in particular on the RID2 due its development status. At the time of testing, an alternative load-path was found that made the readings of the upper neck load-cell unusable. The design of the dummy has subsequently been changed. Also the plausibility of the Nkm values measured on the BioRID II dummy has been questioned. Therefore it was decided not to include the data in this paper.

The dummies' rebound velocities for each seat are shown in Figure 9. The rebound velocity is relatively independent from the dummy type used. For this limited study the rebound velocity does not correlate to the claims frequency data. However, the rebound velocities of all three dummies are significantly lower for seat C, compared to the other seats. This seat also has a low claims frequency for total STNI's. It is notable that the rebound velocity actually increases slightly if the self-aligning (active) head restraint is used.

Figure 9

Rebound velocity measured by film analysis



CONCLUSION

This initial study is too limited to make firm recommendations and further research is necessary. However some promising results have been obtained, leading to preferred routes for further research and development.

The Hybrid III and RID2 dummies showed correlation to the real world claims data when the lower neck extension moment is used as an assessment criterion. Also, the newly proposed LNL-index with the RID2 has shown good correlation to the real world claim data in this study. At the time of testing, the use of the BioRID II dummy with suitable instrumentation to measure lower neck moments or LNL was not possible. However, a lower neck load cell for BioRID II recently became available and this should be validated in dynamic testing. For this study, the LNL for the rebound was not assessed as realistic seatbelts were not used in the testing. However, theoretically, an application of the LNL to the rebound phase should be possible. Currently there is no quantitative force and moment data of injury thresholds available for LNL. Further research will now focus on this issue. The need for different intercept values for different dummies will also be investigated

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APPENDIX

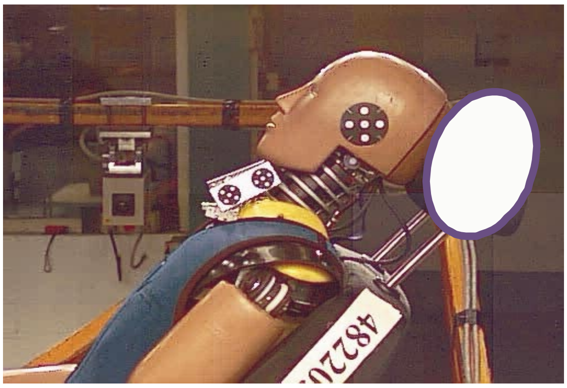
Table 2

Pre test distance Head to Headrest

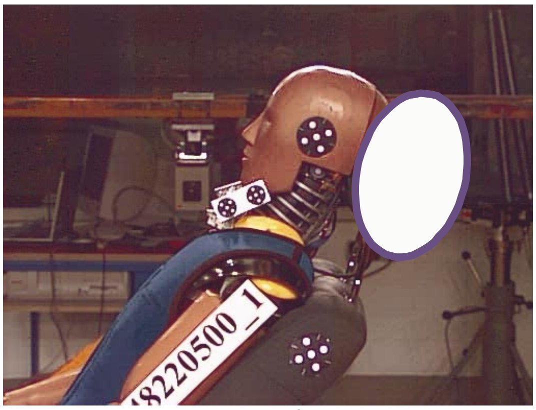
	H III	RID2	BioRID II
Seat A	95	87	82
Seat B	87	64	70
Seat C	94	75	55
Seat B - Active	83	87	68

Figure 10

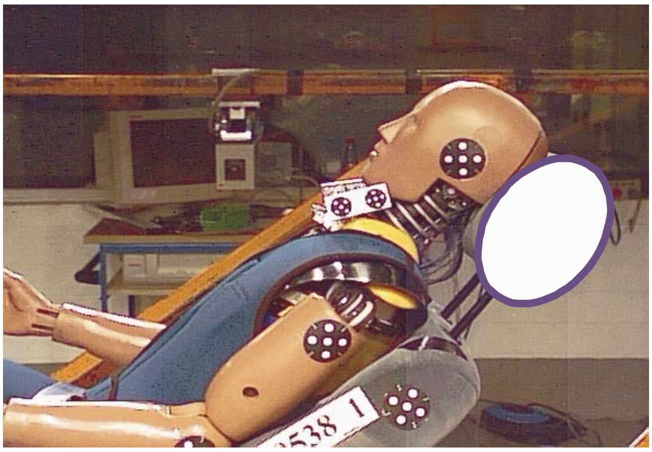
Dynamic images of RID2 at the moment of maximum rearward displacement



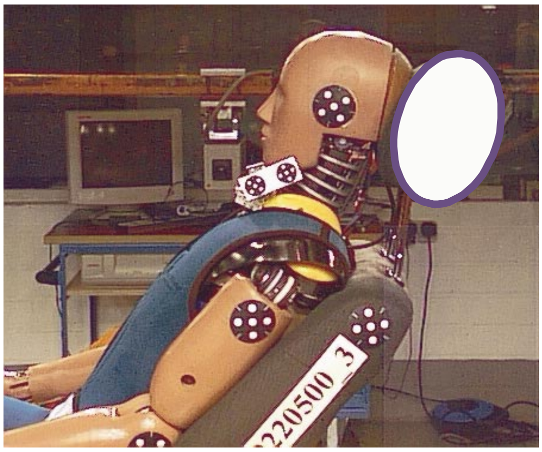
Seat A - Dynamic deformation at 137 ms



Seat B - Dynamic deformation at 127 ms



Seat B - Dynamic deformation at 147 ms



Seat B active - Dynamic deformation at 115 ms