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# A comparison between NIC-values and upper neck moment during the early phase of neck motion in low speed rear impacts

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## Abstract

Whiplash associated disorders to car occupants still constitute a major problem in rear end collisions. Determining neck injury mechanisms and designing methods to measure neck injury related parameters are current research concerns. Recently, a new neck injury criterion (NIC) has been proposed and shown to be able to predict real-life neck injuries by comparison with insurance data. In this study, NIC has been used to compare the rear impact dummy (RID) neck with the human neck on the basis of dummy sled tests and recently published findings from neck experiments with volunteers and cervical specimens. For the RID-neck, a linear correlation ($r^2$=0.93) was found between the NIC and the upper neck moment. On the basis of this finding we suggest the existence of a correlation between NIC and the mechanical loading parameters of the neck during the early flexion-extension phase of neck motion. Our analysis also showed that the RID-neck qualitatively but not quantitatively behaves like the human neck during this early phase of motion.

## Introduction

Whiplash associated disorders as a consequence of low speed rear collisions have been given increased attention over the last ten years. In Sweden epidemiological studies of neck injuries, development of a suitable dummy spine, and research into underlying neck-injury mechanisms are currently ongoing studies. In the middle of the 80s, Aldman [1986] suggested that rapid volume changes in the spinal canal could be a cause of neck nerve injury. Svensson et al. [1993a] tried this hypothesis successfully in their experiments and found damages on the central nervous system. Also, Svensson [1993] developed the rear impact dummy (RID)-neck capable to retract, that is; able of rearward (relative the torso) head translation without rotation, phase 1 in Figure 1.

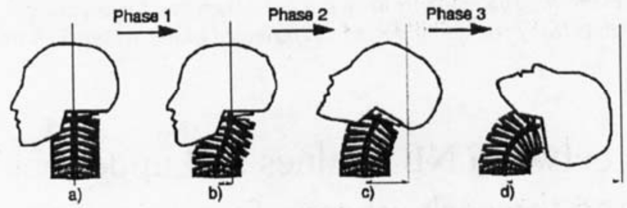


Figure 1 - Schematic view of four parts of the head-neck motion during a rear-end collision: a) initial posture, b) full or maximum retraction, c) maximum rearward angular velocity of the head is reached, d) hyper extension. (from Svensson [1993]).

Based on the findings of Svensson et al. [1993a], Boström et al. proposed a neck injury criterion NIC [1996]. The scientific basis for the NIC-criterion has been further substantiated in the recent work where NIC-values in simulated real-life rear collisions have been compared with the actual injury outcome [Boström et al., 1997].

Ono et al. [1997] and Kaneoka et al. [1997] analyzed the motion of the cervical vertebrae of human volunteers during simulated rear end crashes. Although the level of violence was non-injurious, they found that the neck, in general, retracted, and that the vertebrae, in particular moved in a rather unusual way beyond their normal range of motion. Grauer et al. [1997] used neck specimens to show the occurrence of the same phenomena at higher levels of violence.

Although the injury mechanisms have not yet been firmly established, existing research data may indirectly *explain* the underlying causes of injury (an explanation could be sufficient in order to design neck protection devices in the car).

The aim of this study is to compare the RID-neck with a human neck regarding retraction properties, and to compare the possible transient pressure effects, estimated using NIC, with the inertia loading on the neck quantified by the measured upper neck moment and shear force.

## Material and methods

This study is based on a reanalysis of 20 mechanical simulations of occupant loading during a rear end impact, performed by Svensson et al. [1993b]. In these tests, the RID-neck was used. Five different front seats were used, and the tests were performed for both a square wave 5g sled pulse, $\Delta v$=5 km/h and a square wave 7g pulse, $\Delta v$=12.5 km/h. The dummy was instrumented with x-directional accelerometers in the head, chest and pelvis, and force and moment transducers at the lower and upper part of the neck. NIC was calculated, for these tests, with the formulas:

$$NIC = a_{rel} \cdot 0.2 + v_{rel}^2 \qquad 1)$$
$$a_{rel} = a_{T1} - a_{head} \qquad \text{(local x-direction)} \qquad 2)$$
$$v_{rel} = \text{integral}(a_{rel}) \qquad 3)$$
$$a_{T1} = 1.4 \cdot a_{chest} - 0.4 \cdot a_{pelvis} \qquad \text{(local x-direction)} \qquad 4)$$

Since the acceleration was not measured at T1 in these experiments equation 4 (a correlation found between the acceleration in chest, pelvis and T1 in other experiments with the HIII dummy) was used. According to Boström et al. [1996], the NIC value corresponding to the measurement of the transient pressure effect found in the experiments of Svensson et al. [1993a], is calculated when the *major part* of the upper neck starts to extend (in contrast to the initial flexion motion). In reference [Boström et al., 1997] NIC was calculated after 50 mm of relative T1-C1 displacement. The distance 50 mm was the assumed human length scale for start of pressure transient effects. For the tests in this study the intial maximum NIC value, referred to as the *NIC-max*, was chosen to be compared to the initial upper neck moment, referred to as the *moment-max*. See also Figure 2.

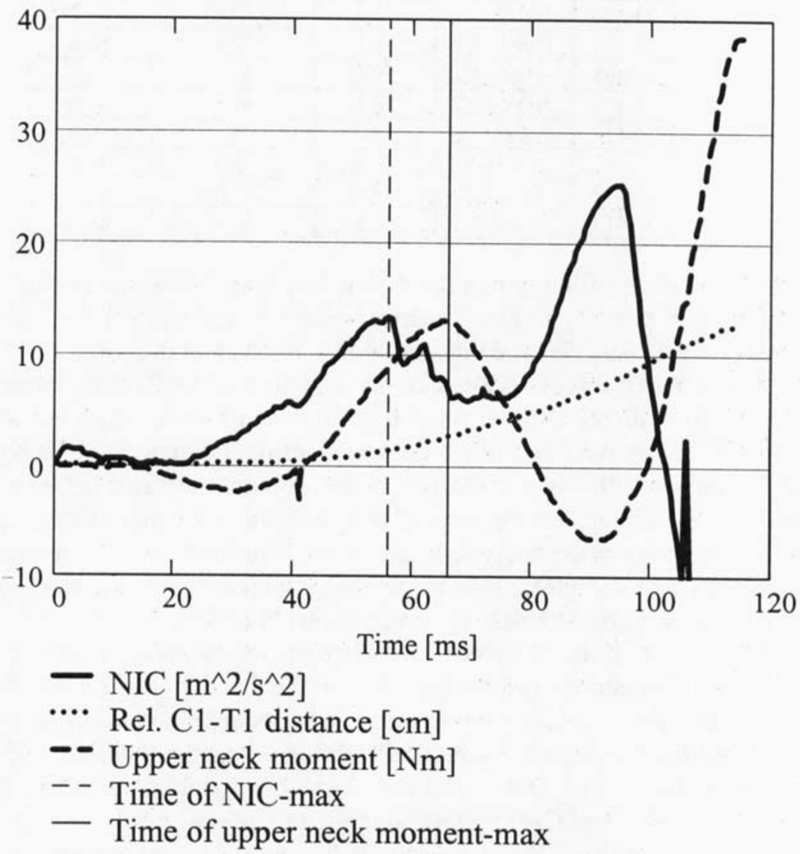


Figure 2 - NIC, upper neck moment and relative C1-T1 distance for the first 120 ms of a 12.5 km/h sled rear impact test (A HIII dummy with a RID-neck was used). Also, the events (the times) of the initial NIC and upper neck moment-max are indicated in the diagram (at 56 and 66 ms respectively).

Sensor Data - Table 4 provides a summary of the maximum head, pelvic, and torso (chest c.g. for the dummies, T1 vertebra for the cadavers) accelerations. It is evident that the dummy tests all recorded low head and chest accelerations and all measures were repeatable for a given test environment. The cadavers in tests C-411 and C-413 experienced considerably higher head accelerations than in test C-412. Furthermore, the cadaver in test C-413 experienced much higher chest accelerations than either of the other two cadaver tests. In an effort to understand the differences among cadaver tests and between the dummy and cadavers, the belt force-time histories were examined.

Table 4. Occupant response parameters.

| Test | Surrogate | Head Acc. (g's) | Chest CG/ T1 Acc. (g's) | Pelvis Acc. (g's) |
|---|---|---|---|---|
| D-406 | Hybrid III | 54.3 | 35.9 | 46.2 |
| D-407 | Hybrid III | 56.1 | 36.6 | 53.6 |
| D-408 | Hybrid III | 59.7 | 37.5 | 54.5 |
| D-409 | Hybrid III | 56.1 | 35.4 | 50.6 |
| D-410 | Hybrid III | 51.2 | 34.7 | 50.1 |
| C-411 | Cadaver | 59.8 | 43.1 | 48.4 |
| C-412 | Cadaver | 44.8 | 43.5 | 52.0 |
| C-413 | Cadaver | 82.8 | 60.4 | 61.6 |

Analysis of the dummy belt force-time histories shows repeatable behavior of the force limiting belt system. Figure 2 clearly identifies activation of the 1 kN pretensioner at approximately 17 ms, which coincided with activation of the squib for the airbag system. A rise in belt forces between 20 ms. and 40 ms. results from interaction of the deploying airbag with the anterior portion of the dummy torso which is traversed by the shoulder belt. Forward motion of the dummy does not begin to load the belt significantly until 40 ms. Increased loading of the belt by the dummy increases belt forces until detachment of the pretensioner cable. During cutting of the pretensioner cable, an overshoot in the force was observed in two of the four tests. While this increased the maximum belt force, the yielding of the torsion bar and subsequent plastic deformation resulted in a fairly constant belt load level of 3000 N to 3500 N.

Examination of the belt force-time histories for the cadaver tests in Figure 3 shows that the belt restraint system did not limit the force in tests C-411 and C-413. The force traces for dummies and cadavers exhibited similar behavior during belt pretensioning, initial occupant interaction with the airbag, and belt loading by the occupant until 60 ms. Test D-410 and C-412 exhibited yielding of the torsion bar after 60 ms. while tests C-411 and C-413 show continued increases in belt load. A sharp drop in the force at 70 ms. of test C-413 was attributed to partial severing of the pretensioning cable. The force limiting at 6.5 kN that occurred in test C-411 presumably resulted from pulling of the pretensioning cable through the retractor and across the knife edge.

## Results

As described in the previous section, ten Δv 12.5 km/h and ten Δv 5 km/h dummy-sled rear impact tests were evaluated. The maximum initial upper neck moment and NIC values are compared in Figure 3. A linear correlation ($r^2 = 0.93$) was found. Table 1 provides the mean and standard deviation for the relative neck displacement between T1 and C1 (or rather the head) at the event of NIC and moment-maxima, see Figure 2. There was also a linear relation between the upper neck moment and the upper neck shear force during phase 1 in Figure 1.

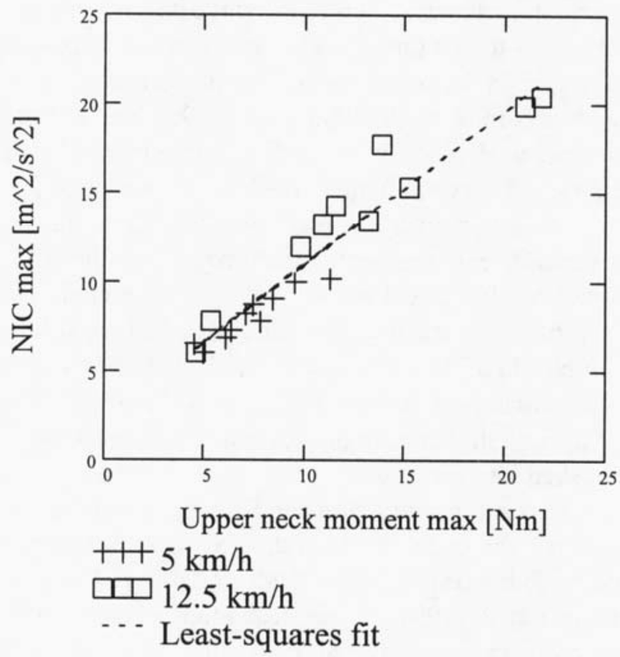


Figure 3 - NIC-max versus upper neck moment-max for ten Δv 12.5 km/h and ten Δv 5 km/h dummy-sled rear impact tests.

Table 1 - Mean value and standard deviation for the relative displacement at the event of NIC and moment-maxima, see Figure 2, for the 20 front seat sled-tests of Svensson et al. [1993b].

| | Mean value [mm] | Standard deviation [mm] |
|---|---|---|
| Relative displacement at the event of NIC-max | 16 | 5 |
| Relative displacement at the event of moment-max | 25 | 7 |

## Discussion

When inertia forces are transmitted through the human spine, as in a rear end impact, causing a retraction of the cervical spine, a (whiplash-like) *extension wave* propagates from the lower to the upper part of the neck. This is evident from the neck joint-angle time diagrams for the whiplash simulations in Kaneoka et al. [1997] and Grauer et al. [1997], where the derivative of the angles subsequently from T1 to the atlas-occiput joint changes sign "... each segmental motion starts from the lowest motion segment and gradually transfers to the upper segments" cite Kaneoka et al. [1997]. When the wave has reached the atlas-occiput joint, the head angular acceleration as well as the upper neck moment reaches its maximum value followed by a pure extension of the complete neck. Before the wave has reached the head the major part of the upper neck has been forced through a phase transition of flexion to extension causing a rapid volume change in the spinal canal. This phase transition may cause damage to the central nervous system according to Boström et al. [1996]. The two effects, the rapid volume change and the upper neck mechanical loading is manifested by the initial maximum NIC value followed by an intial maximum upper neck moment seen in the RID tests.

The RID-neck retracts in a human like way. However, quantitatively there is a significant length-scale difference. In comparing the images (contours) of the the neck during the simulated whiplash motions in Kaneoka et al. [1997], Grauer et al. [1997] and in Deng et al. [1987], with Table 1, it becomes apparent that the full retraction distance (relative C1-T1 distance from a to b in Figure 1) of the RID-neck is much shorter (about 20 mm). An estimation made from the human neck tests, yields a value of at least 50 mm relative horizontal displacement to full retraction. In the development of new biofidelic rear impact dummies this should be taken into account.

The words full (or maximum) retraction and S-shape have been used in the literature to define the neck shape when the upper levels of the neck is at flexion and the lower levels are at extension. According to the results of this study and the studies of Svensson [1993], Ono et al. [1997] and Grauer et al. [1997]: upper neck moment max, NIC-max and the event of transient pressure effects takes place close to the occurrence of this neck shape. However, when it comes to more precisely determine the event of these effects, more information is needed. In other words, phase 1 in Figure 1 must be further subdivided into new phases.

## Conclusions

In conclusion, taking into account the results of the reanalysis and comparison of the data in the studies of Svensson [1993], Ono et al. [1997], Grauer et al. [1997], and Deng et al. [1987] for low speed rear collisions:

- The RID-neck, in combination with a HIII torso, seems to *qualitatively* but not *quantitatively* resemble the human neck regarding the retraction motion.
- There is a correlation between initial maximum NIC-value and the initial maximum upper neck moment for the RID-neck.

## Recommendations

In this study, a HIII dummy with a RID-neck was used. The conclusions imply that further rear impact dummy development must take the entire spine configuration into account to achieve desired characteristics. Also, a more detailed analysis of already performed human neck experiments from a NIC point of view is recommended by other teams to further strengthen and broaden the conclusions drawn in this report.

## Acknowledgement

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## DISCUSSION

PAPER: **A Comparison Between NIC Values and Upper Neck Moment During the Early Phase of Neck Motion in Low Speed Rear Impacts**

PRESENTER: Hugo Mellander, Traffic Safety Research & Engineering AB, Sweden

QUESTION: Michael Kleinberger, NHTSA

I would like to ask Mr. Ono to make some comments because Kaneoka and Ono worked together on some of that data you were presenting and I believe they also tested the RID neck. Would you care to make a quick comment on your comparison between the RID neck and your volunteer data since you've actually seen the data, if you recall the numbers?

ANSWER: Koshiro Ono, Japan Automobile Research Institute

We have done some work using the TRID neck with the Hybrid III dummy. Yesterday, I showed a videotape and you could see a big difference between the dummy and human volunteers. The main difference comes from the different straightening of the torsos, but there is also some difference in the TRID neck. So, because of this torso effect, the TRID neck undergoes a different motion.

Q: OK. Thank you.

Q: Guy Nusholtz, Chrysler Corporation

Is the basic purpose of this neck just for the type of test setup that you had? I noticed the way that you are doing the NIC calculation, it seems that you are ignoring any rotation in the head because you are just integrating the accelerations to get your velocities. The moments also didn't compare in time. So you've got a correlation but the peak moments are out of phase and perhaps that only works in that type of configuration. Are you planning to work towards generalizing the neck or is it just good for that type of test?

A: Certainly, there are several things going on. We are continuously refining the RID neck and we are trying to come up with a neck that could eventually be used on a dummy with associated dummy criteria. So, certainly, yes, we will continue this work.

Q: OK. Thank you.

Q: Rolf Eppinger, NHTSA

I'm continually confused by people that attempt to associate neck injury with upper neck moment. If I start with your design of the RID neck, I can understand that the moment that is applied to the head, comes completely from the first C1 vertebra or the equivalent of C1 on your RID neck. That moment is a function of what is applied from C2 and so forth on down. So, all the moment is progressively generated along each one of the vertebra and carried completely through the vertebral column. If I were to consider a physiological system that would mimic your RID neck, all muscle insertions and origins could only span one cervical motion segment, which we know is not anatomically correct. Sometimes we have origins and insertions that possibly span the entire neck. This is a very important distinction that we have to understand because even the Mertz analysis and a variety of other analyses, look at the sum of all the forces that the neck applies to the head and then calculates a resultant force and moment at a particular point. That is an imaginary moment. It is a calculated moment and not a truly applied moment through the physiological structure. So, much of the moment that you are seeing in a RID neck or in any dummy neck could be generated as a result of forces being applied at a distance from C1. And so, if you are trying to associate injuries with moments measured in a dummy neck, that moment could be substantially different than the moment that is actually in a true physiological system. I would suggest that possibly you look at other factors other than moment to make an association with injury. I would like some of your thoughts on that.

A: OK. I understand and appreciate what you are saying. Again, this is an ongoing project. Per Lovsund, do you want to explain a little bit of your thoughts about how we are going about this?

A: Per Lovsund, Chalmers University of Technology, Sweden

I agree that the moment is probably not the best injury measurement here, but we have found a correlation between the moment of the NIC criteria in pig experiments where we have found injuries to nerve cells or the nerve roots in the spine. So, we have correlation here between these NIC criterion and actual injuries seen in the animals. I will also agree that the moment we measure in the RID is affected by the vertebrae and also by the spine of the Hybrid III dummy, which we think is not good. This is not the final solution, so that is why we have a new dummy now with a completely new spine. We hope to have in the final version good characteristics of the whole spine, which of course, will influence the moment. These are preliminary findings, and we would now like to have a discussion about neck injury criteria. This is just our first step.

Q: If I would take my argument to an extreme. If all moments were generated by muscles attached to the head that are fastened to T1, then you could have an effective moment on the head, but you would have no moment in the neck at all. So, if you were making a mechanical analog, that is the most extreme situation you would have. You would have a column, but the column could have a frictionless pin at the top for the occipital condyle joint, and no moment would be transmitted into the neck at all. All the moment is translated down and is a function of the relative motion of the head with respect to T1. Now, in reality, it is probably somewhere halfway in between. In other words, some of the moment is generated by muscles as a function of the relative position of the head, and some moment is applied along the vertebral column through the occipital condyles. I'm trying to understand, if you create this mechanical analog and continue to use moment as the criterion of consequence, how you can attempt to find a correlation back to injury with that?

A: I would like to add that this was just a suggestion. We are not trying to replace the NIC with the moment at the moment.

Q: I guess that addresses neck design in general. If you look at our THOR neck, we tried to go to the opposite extreme. We actually have external cables that attach to the head and go down into the bottom and we're getting lots of complaints by people saying that we don't see any neck moments anymore. Well, maybe that is more realistic than all the current necks and how they are constructed. So if I'm creating artificial moments that are just a function of my design choices, how should that have an association back to injury?

A: OK. I get your point.

Q: Per Lovsund, Chalmers University of Technology, Sweden

This is not the final solution. We are playing with different muscle substitutes in a new model.

Q: Jeff Pike, Ford Motor Company

Is any work being done on perhaps a two-tier NIC; one for the relatively minor injuries and one for the more severe ones?

A: No, not that I am aware.