A STUDY OF CURRENT NECK INJURY CRITERIA USED FOR WHIPLASH ANALYSIS. PROPOSAL OF A NEW CRITERION INVOLVING UPPER AND LOWER NECK LOAD CELLS.

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ABSTRACT

Nowadays several injury criteria are being used in the analysis and evaluation of whiplash risk in automotive rear impacts (NIC, Nkm, LNL, etc.). This study presents a review of the most accepted injury mechanisms and evaluates the advantages and inconveniences of the commonest criteria at present. Taking into account the conclusions arrived at during this comparison, a new criterion is proposed using the signals registered in the upper and lower neck load cells of a crash test dummy, trying to minimize the disadvantages previously found in the other criteria. In order to validate this study a series of sled tests with a BioRID-II dummy have been performed and its results analyzed, confirming the assumptions made during the review of the present criteria and showing a very promising response to the new one. In conclusion, the use of injury criteria involving the load cells situated in both ends of the neck at the same time is recommended as the best way to deal with the dynamics produced during the whiplash movement in a rear impact.

INTRODUCTION

In a rear-end car crash, even at low speed, the head of the occupants of the struck vehicle normally suffers a motion related to the torso that produces sudden distortions of the neck. Although in the most severe cases this movement can produce the fracture of cervical vertebrae, the commonest related lesions are only classified as minor injuries (AIS 1) [1]. Nevertheless, these lesions, known as whiplash-associated disorders (WAD) or simply whiplash, produce painful and often long-term or even chronic symptoms, causing huge economic costs to the society at the same time.

During the last few years a certain number of experimental procedures have appeared trying to evaluate the capacity of the automotive seats to protect the occupants in a rear-end crash. Currently the most accepted of these procedures (IIWPG [2][3], Folksam [4], ADAC, etc.) are using dynamic sled tests and the crash test dummy BioRID-II [5][6][7]. One of the main problems in the development of this kind of procedures has been related to the lack of a full understanding of whiplash injury mechanisms, even though several theories have been proposed trying to give an explanation to the observed symptoms. At the same time, a certain number of injury criteria have been developed looking for a correlation with the different proposed mechanisms. At present there is still a debate about which of these criteria should be taken into account to describe the ability of a seat to protect the neck of the occupants in a rear-end impact properly. In this situation, the groups that are developing new test procedures are adopting either several criteria simultaneously ([4]) or none of them, basing their assessment on the direct comparison of loads and accelerations ([3]). At this point, the lack of a criterion unifying the different injury mechanisms that can be used easily on a test protocol is clear.

The main objective of the presented work was to make a critical review of the commonest injury criteria used at present, trying to analyze the advantages and disadvantages of each one of them. The results would provide a better understanding about the different criteria themselves and, if possible, give guidelines for the definition of a new criterion solving the possible problems found.

METHODS

Keeping this objective in mind, the first question is: how do we evaluate a whiplash injury criterion? or even better, what do we expect from it?. The points found by the authors to answer this question are the following:

1. The criterion must be representative of one or more injury mechanisms, indicating and quantifying the probability of injury. It must be sensitive to the factors related to these injuries and able to give an assessment about different impact conditions. It must be able, for instance, to determine which seat is safer for an occupant with regard to the considered mechanisms when using a particular acceleration pulse.

2. At the same time it should be repeatable and stable. Values measured in similar situations should not be too different.

3. It should not be sensitive to other processes

different to the mechanism analyzed. Variables not related with the injury mechanism should not have a great influence on its value.

4. When possible, for practical reasons, the criterion should be easily and quickly calculated. It should use values directly measured during the test and avoid non automatic operations.

These points evidence that in order to proceed to the evaluation of the different criteria it is convenient to get the best possible understanding about what happens where and when in a typical rear-end impact. The dynamics of the neck and head have been studied both in the literature and with results of tests using the BioRID-II dummy. In addition, a review of the most accepted injury mechanisms has been done.

After these reviews, the most common injury criteria have been analyzed trying to understand their weak and strong points. A series of four sled tests with seat, dummy and seat belt have been done in order to validate the obtained conclusions. All the tests have been carried out at CIDAUT, using a MTS inverse catapult and a BioRID-II Rev.f fully instrumented dummy. The forces at the seat belt were measured using a Messring belt load cell, in order to get extra information about the rebound phase. The seating procedure was based on [2]. The position of several characteristic points of the dummy was registered with a FaroArm portable 3D measurement system, in order to guarantee its reproduction when using similar seats. The sled was accelerated using the IIWPG 16 Km/h pulse [2] (Figure 1 shows the acceleration measured in the different tests). Four Redlake high-speed digital cameras were used during the tests in both on-board and off-board positions, taking images at 1000 fps. When necessary, image analysis was done using the software Falcon eXtra. All the signs and axis mentioned on the present paper are according SAE J1733 standard ([8]).

Two models of seats have been chosen for the tests. As none of them has been specifically designed to prevent whiplash, we will refer to them as Seat "A" and Seat "B". Seat "A" is a common car driver seat, while Seat "B" is a minibus rear seat with an integrated 3-point seat belt. This forces its structure to be very rigid and, therefore, is expected to give worst results with regard to whiplash protection. Three tests were done with "A" type new seats (numbers 001, 002 and 003), and a fourth one was done with a seat "B", also new (number 004). In this way we could analyze the repeatability and sensitivity of the different criteria. Figure 2 shows the rotation of the backrest in the tests, measured from the high speed images. The difference of stiffness between both models of seats appears clearly here (the rotation on the fourth test has been quite lower than on the other ones). The variability of the behaviour of the "A" seats can also be observed, even when using similar acceleration pulses. This can be used as a reference when studying the repeatability of the criteria.

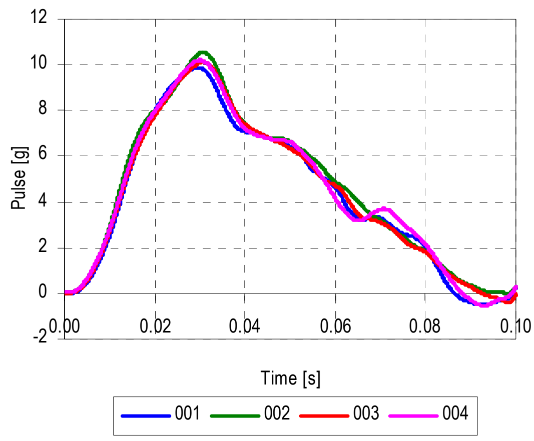

Figure 1. Acceleration pulses of the tests.

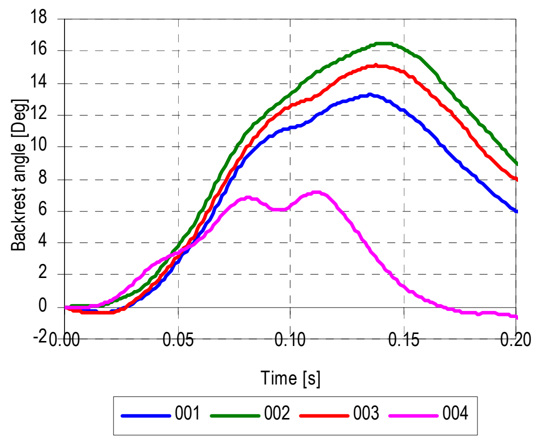

Figure 2. Rotation of the backrest during the tests.

In order to have numeric values to compare the sensitivity and the repeatability of the different criteria, a method has been defined using the Russell criterion for comparison of curves [9]. This criterion is normally used to compare two different series of data $f_1(i)$ and $f_2(i)$ defined by N points each, giving a numeric value ε_c closer to 0 when the curves are similar and greater when the curves are different. The expressions used are the following:

$$A = \sum_{i=1}^{N} f_1(i)^2 \qquad B = \sum_{i=1}^{N} f_2(i)^2 \qquad (1).$$

$$C = \sum_{i=1}^{N} f_1(i) f_2(i)$$

$$m = \frac{A - B}{\sqrt{AB}}$$

$$p = \frac{C}{\sqrt{AB}}$$

$$\varepsilon_p = \frac{\cos^{-1}(p)}{\pi}$$

$$\varepsilon_m = sign(m)\cdot\log^{10}(1 + |m|)$$

$$\varepsilon_c = \sqrt{x\left(\varepsilon_m{}^2 + \varepsilon_p{}^2\right)}$$

The values ε_m and ε_p represent respectively the errors associated to differences in magnitude and phase, and x is a reference constant that, in this case, has been defined as $\pi/4$.

To get an indicator of the repeatability of the injury criteria the first three tests have been compared to each other (001 to 002, 001 to 003 and 002 to 003), obtaining three ε_c values as results. The average of these values has been considered to be representative of the repeatability. The indicator for sensitivity has been calculated in a similar way, comparing the three first tests with the fourth one and calculating the average of the three obtained ε_c. As defined, the repeatability is assumed to be better when its indicator is closer to zero, and the sensitivity is better when its indicator is higher. To be used as a reference, the indicators of repeatability of the acceleration pulses (high repeatability and low sensitivity) and the rotations of the backrest (relative low repeatability and high sensitivity) were 0.028 and 0.108 respectively, while its sensitivities were 0.022 and 0.482.

HEAD-NECK MOVEMENTS DURING A REAR-END IMPACT

In order to be able to analyze the results of the tests and to try to identify the time when the possible injury mechanisms happen, it is indispensable to understand the kinematics of the neck and the head during a typical low speed rear-end impact. This movement is well documented and has been described by several authors using different techniques ([5], [10], [11], [12] and [13] among others). The main phases of the motion are shown in Figure 3.

In the initial state the subject is seated on the seat in normal position. When the vehicle is struck, the acceleration of the structure is transmitted to the seat through its anchorages, producing a movement forward with regard to the occupant. The first zone of the subject in receiving the pressure of the seat is normally the pelvis and the lumbar zone, followed by the thorax. When the spine, originally curved according to its physiological shape, is pushed forward, it tends to straighten, moving the base of the neck (vertebra T1) upwards and producing some compression on it. This phenomenon can be amplified by the movement upwards of the whole thorax due to the angle of the seat and the acceleration of the base. This is commonly called "ramping up". Although the thorax begins to move, the head at this point remains in its original position. The T1 vertebra, which was originally situated behind the centre of gravity of the head, passes to be in front of it, and the previous compression of the neck becomes traction, with the thorax pulling on the head. The movement of T1 makes the cervical vertebrae work as a chain, transmitting the motion from the lower end upwards, while at the upper end the inertia of the head produces resistance to the movement. The combination of these effects produces a transitory biphasic state known as "s-shape" in which the lower part of the neck (vertebrae C5-C7) presents a very pronounced extension, while the upper part is in flexion. The rearwards movement of the head referred to T1 is called retraction

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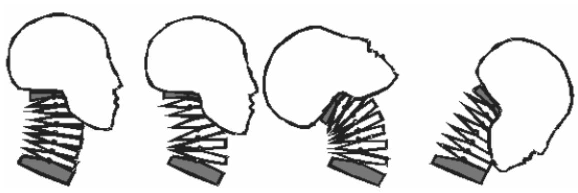

Original S-shape Extension Flexion (rebound)

Figure 3. Different phases of the motion of the head in a typical rear-end impact.

When finally the head begins to rotate, the whole neck arrives in a state of extension with the head being pulled on by the thorax. When the acceleration of the base drops, the elastic energy stored on the seat and the occupant begins to be released, producing a rebound movement with a rotation forward of all the torso of the subject around the pelvis. The seat belt begins to tense over the pelvis and the thorax approximately when the body returns to its original position, producing a violent flexion of the neck. Finally, due to the tension of the belt, the body is stopped, and returns to the backrest.

THEORIES ABOUT WHIPLASH INJURY MECHANISMS

Up to the present a wide number of research works have been done trying to identify the origin of the symptoms related to whiplash associated disorders. As a result of these studies several injury mechanisms have been proposed, the coexistence of some of them being the most accepted hypothesis. If we want to analyze the different injury criteria it is necessary to understand the origin of the lesions as well as possible, in order to be able of relate them with the magnitudes measured in the lab. A review of the most accepted mechanism has been done keeping this idea in mind. Some of the main ones appear below:

Hyperextension

The hyperextension of the neck was the first hypothesis trying to explain the whiplash phenomenon. It was proposed in the sixties by Macnab [14], and suggested the movement of extension of the neck to be the cause of the whiplash injuries, producing lesions on the lower cervical spine. In 1969 the incorporation of head restrains in the new cars sold in USA was made compulsory, trying to limit this movement. However, this fact did not reduce the number of reported whiplash cases in the expected proportion, making evident the necessity for further research. Although hyperextension is still a possible cause of injuries, today the extended use of head restraints has limited it to particular cases, such as misuse or failure of the headrest.

Cervical flexion during the rebound phase

Opposite to the previous mechanism, Macnab also proposed the flexion of the neck due to the movement produced by the head when the seat belt acts on the rebound phase as a probable origin of injuries [15]. This was suggested after the observation of a higher frequency of cervical injuries on people using seat belt, and later confirmed by other authors ([16], [17] and [18] among others).

Pressure gradients on the spinal canal

In 1986 Aldman [19] predicted that volume changes produced inside the spinal canal during sudden movements of the cervical spine on the sagittal plane could be the origin of injuries in the intervertebral tissues. In 1993, Svensson et al. [20] confirmed this hypothesis, measuring the pressure changes on the spinal canal of anesthetized pigs and reporting damage to the spinal ganglia that could explain many of the typical symptoms of whiplash. In these experiments the highest pressure oscillations were related to the phase shift from the s-shape to the extension, and the highest pressures were registered at the level of the C4 vertebra during the s-shape.

Localized cervical compression and tension during the s-shape

Nowadays the most accepted cause of whiplash injuries is probably the one related to the hyperextension observed in the lower part of the neck during the formation of the s-shape (vertebrae C5, C6 and C7). In 1998 Panjabi et al. [21] reported that the intervertebral movements observed at these levels during in vitro tests exceeded their physiological limits, being the cause of lesions in the capsular ligaments and facet joints at the C5-C6 level. Similar findings have been done later by other authors ([22], [23] and [24] among others).

COMPARISON OF THE MOST USED CRITERIA

Figure 4 shows the sensors that at present are being included in a BioRID-II dummy as normal instrumentation. The signals of these sensors and the measurements done by image analysis on the sequences registered with high speed cameras are the current available tools to quantify the ability of a seat to protect the neck of an occupant during a low speed rear impact. Several criteria have been developed in order to quantify the risk of having whiplash related disorders, based either on accelerations, displacements or loads. The most accepted among these criteria have been evaluated critically by the authors trying to understand their virtues and defects. Below the results of the evaluation and its application to the tests are presented:

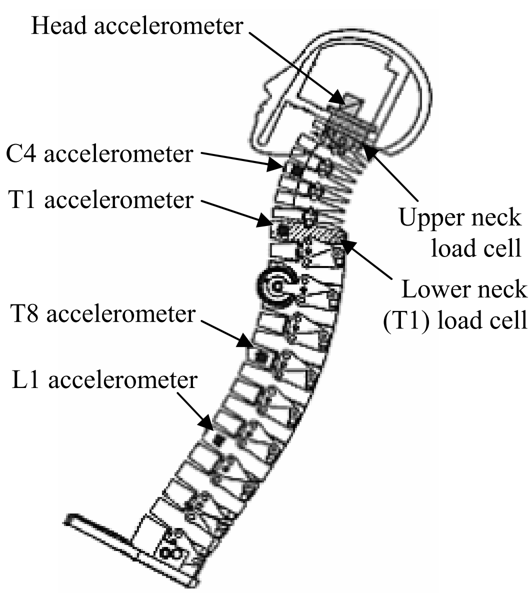

Figure 4. Standard instrumentation in spine and head of BioRID-II (Adapted from R. A. Denton drawing 5834 www.dentonatd.com).

NIC

NIC (Neck Injury Criterion) was proposed by Boström et al. in 1996 [25], as a value to correlate the movement of the head related to the base of the neck (T1 vertebra) with the damage found in the cervical spinal ganglia produced by transient pressure changes in the spinal canal. It uses the difference of accelerations in the longitudinal direction (x axis) between the centre of gravity of the head and the T1 vertebra, being therefore representative of the movement of the neck during the retraction phase. NIC is calculated as follows:

$$\begin{aligned} NIC &= a_{rel} \cdot 0.2 + v_{rel}^{2} \\ a_{rel} &= a_x^{T1} - a_x^{Head} \\ v_{rel} &= \int a_{rel}\, dt \end{aligned} \qquad (2).$$

The maximum reached by this expression during the first 150 milliseconds of the test is called NIC_{max}, and for years has been considered as one of the main indicators of whiplash.

Figure 5 and Table 1 show the NIC values achieved during the tests. The repeatability of the results of the first three tests is very good (with an indicator of 0.084), and even impressive looking at the maximum values. It is necessary to mention here that such a high repeatability of the maximum values is not that common in practice. On the other hand, the different behavior of the seats A and B has been well characterized, having a value of 0.407 on the sensitivity indicator.

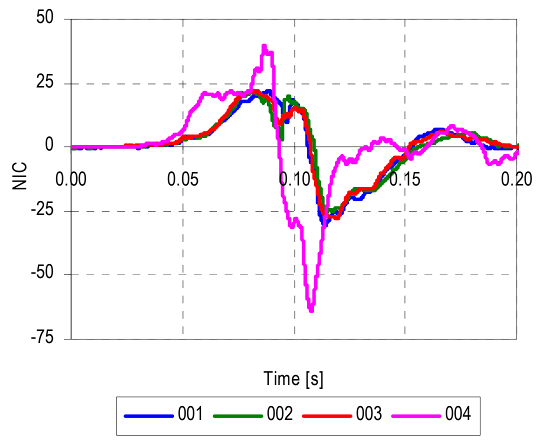

Figure 5. NIC.

Table 1. Maximum NIC values.

Test	001	002	003	004
NIC_{max}	21.94	21.94	21.93	40.08
Time (ms)	88.6	79.9	81.8	86.7

When analyzing the causes that can produce different accelerations in the longitudinal axis between the head and the T1 vertebra and, therefore, cause a modification on the value of the NIC, we observe that this difference can not only be produced by distortions in the neck, but also by any rotation of the head and T1 around the transversal axis (Y) as a rigid body. This movement does not cause any deformation in the neck, and, apart from extreme cases, should not be a direct cause of injury. We can see a scheme of this in Figure 6.

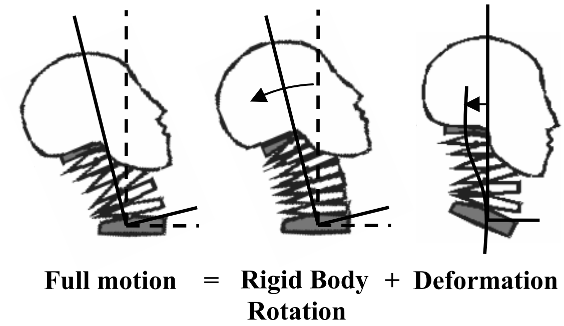

Figure 6. Decomposition of movements producing NIC values.

The influence of this effect can be estimated dividing the relative acceleration used in the NIC definition in two terms:

$$a_{rel} = a_{rotation} + a_{deformation} \qquad (3).$$

If we refer to the angular acceleration of T1 as α and the distance between the centre of gravity of the head and the accelerometer at T1 as d, we can then calculate the acceleration term corresponding to the deformation:

$$a_{deformation} = a_{rel} - \alpha \cdot d \qquad (4).$$

Although d is not fixed for all the configurations of the neck (it is deformable) we can consider 0.2 metres as an average, and we can estimate α from the double derivation of the angle of the T1 vertebra measured on the images (Figure 7).

If we use $a_{deformation}$ instead a_{rel} in expression (2) we get the curves shown in Figure 8. We will refer to these values as NIC*, calculated only with the term related to deformation. Table 2 shows that the maximum values obtained in this way can differ up to 30% from the original NIC values. This variation is produced by factors not directly related to the distortion of the neck.

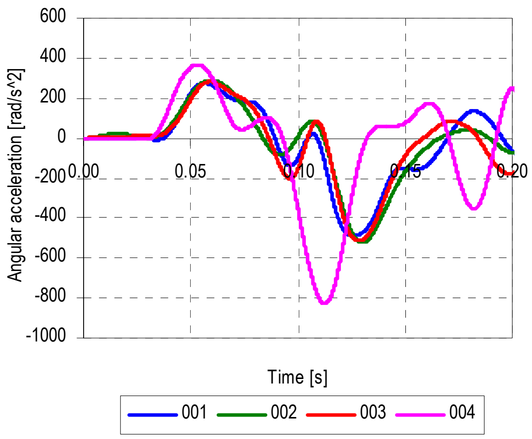

Figure 7. T1 angular acceleration.

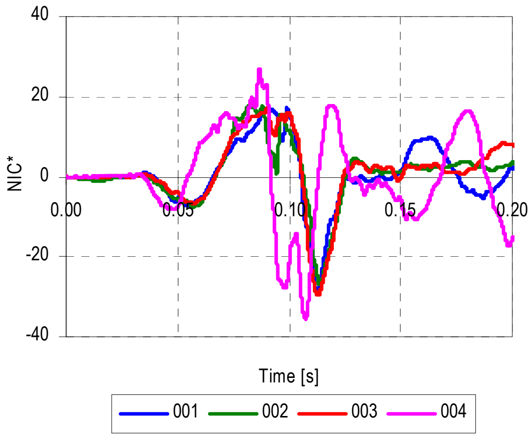

Figure 8. NIC* (without effect of T1 rotation).

Table 2. Maximum NIC* values and deviation with regard to the original NIC.

Test	001	002	003	004
NIC*$_{max}$	17.30	17.72	17.20	26.95
Deviation	21.2%	19.2%	21.6%	32.8%

This fact supports the observations made by Suffel during the fourth BioRID User Meeting [26], who reported the carrying out of some tests blocking the movement of the neck relative to the T1 vertebra, but obtaining NIC values around 8 m^2/s^2.

In short, NIC has shown to have a good repeatability and distinguishes well between the two different seats. It also takes into account the kinematics of the head with regard to the thorax trying to describe the retraction movement, but on the other hand, it is sensitive to effects not related to the distortion of the neck, due to the use of accelerations for its calculation (for instance, the rotation of the seatback produces the effect previously described).

Nkm

In 2001 Schmitt et al. [27] proposed the N_{km} criterion, based on the linear combination of shear forces (F_x) and sagittal bending moments corrected to the occipital condyle ($M_{y\ OC}$), measured with the upper neck load cell. This criterion distinguishes among four possible situations depending on the sign of My and F_x (see Table 3)

Table 3. Cases of Nkm.

Case	M_y	F_x
N_{fa} (Flexion Anterior)	> 0	> 0
N_{fp} (Flexion Posterior)	> 0	< 0
N_{ea} (Extension Anterior)	< 0	> 0
N_{ep} (Extension Posterior)	< 0	< 0

The criterion is calculated as follows:

$$\begin{aligned} N_{km} &= \frac{|F_x|}{F_{\text{int}}} + \frac{|M_{yOC}|}{M_{\text{int}}} \\ F_{\text{int}} &= 845N \\ M_{yOC} &> 0 \Rightarrow M_{\text{int}} = 88.1N\cdot m \\ M_{yOC} &< 0 \Rightarrow M_{\text{int}} = 47.5N\cdot m \end{aligned} \qquad (5).$$

Figure 9 shows two possible representations of the results of N_{km} applied to the tests, and Table 4 the maximum values achieved. After these results we can see that the criterion distinguishes both models of seats very well. With regard to the repeatability, it seems to be lower than that observed on the NIC. The maximum on the test 002 is reached during the phase of extension anterior (N_{ea}), instead of during the phase of flexion anterior (N_{fa}), as happens in the tests 001 and 003. This makes the time of the maximum differ between them. The indicators of repeatability and sensitivity have worse values than the ones obtained for the NIC, being 0.137 and 0.307 respectively.

Table 4. Maximum N_{km}.

Test	001	002	003	004
N_{km} max.	0.33	0.20	0.27	0.62
Time (ms)	128.6	112.6	128.1	108.7
Case	FA	EA	FA	FA

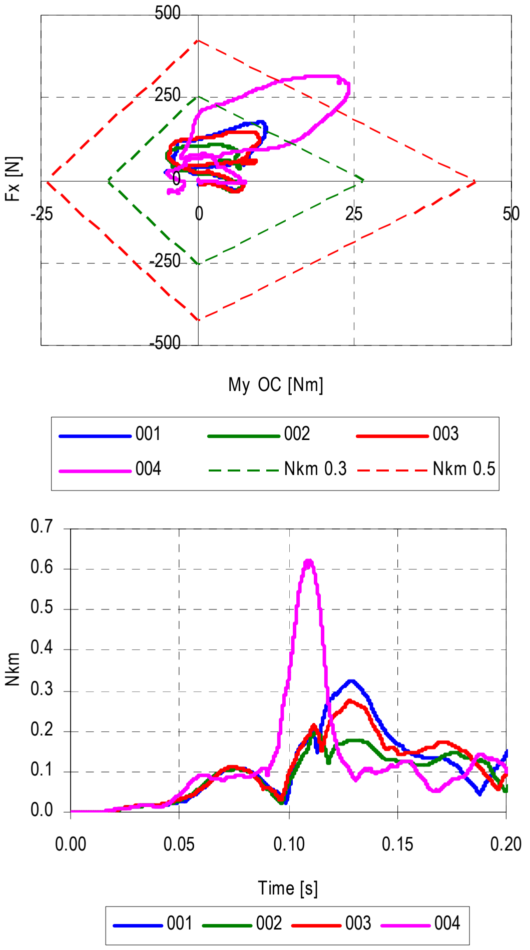

Figure 9. Two representations of N_{km} criterion.

The main advantage found for this criterion is the use of forces and moments, directly related to the loads of the neck, not being affected by other effects such as rotations. Another positive point is its definition in cases, depending on the sign of F_x and $M_{y\ OC}$. This allows the criterion to consider different values and limits depending on the load case. On the other hand, a possible disadvantage is related to the use of the signals measured only on the upper neck load cell, located at the occipital condyle, while the most common injuries have been described between the vertebrae C5 and C7, nearer to the base of the neck. Despite this, the combination of F_x and $M_{y\ OC}$ seems to correlate well with the time in which the s-shape is produced, at least in the studied cases.

Additionally, although the observed influence is not high, it was noticed that the mathematical definition of the criterion as a lineal combination depending on the load case can produce local minimums, oscillations or variations on the tendencies (discontinuities on the derivatives of the curves) at the points of change of case. This can be understood more easily by looking at the first representation of the criterion in Figure 9. The rhomboidal lines represent the points with 0.3 and 0.5 constant values of Nkm. If we intended to have a continuous value of the criteria on the zones of the corners (change of case) we should follow a line with this shape, producing a change in the tendencies of some of the magnitudes (force or moment, depending on the corner) when changing the case, and therefore a discontinuity on its derivative. In practise, the change of the definition of the N_{km} results in discontinuities on its derivative and possible local minimums related only to its mathematical formulation, although, as mentioned above, the influence of this effect has not been decisive in any of the studied cases.

LNL

In 2002 (one year after the proposal of the N_{km} criterion) the prototype of a new load cell placed on the T1 vertebra of the BioRID-II dummy was presented, designed to give information about the loads on the lower end of the neck, next to the vertebrae that had been more often related to injury mechanisms (C5-C7). In March 2003 the version "f" of the dummy was released, already equipped with this load cell. Taking advantage of this new instrument the LNL criterion (Lower Neck Load) was proposed, defined as follows:

$$LNL = \frac{|M_{y\,lw}|}{C_{moment}} + \frac{|F_{x\,lw}|}{C_{shear}} + \frac{|F_{z\,lw}|}{C_{tension}} \qquad (6).$$

In this expression $M_{y\ lw}$, $F_{x\ lw}$ and $F_{z\ lw}$ are the moment and forces measured with the T1 load cell, and C_{moment}, C_{shear} and $C_{tension}$ reference values (15 N·m, 250 N and 900 N respectively). The value to be used for the evaluation of the seats is the maximum of this curve.

The curves obtained when applying this expression to the data of the tests are shown in Figure 10, and the maximums in Table 5. Looking at these results, we can see that the repeatability for the first three tests is excellent throughout the curves (with an indicator value of 0.044), including the maximums, but the criterion has not been able to differentiate well between seats A and B, at least in the maximum values. The indicator for sensitivity has a value of 0.250.

Table 5. Maximum LNL.

Test	001	002	003	004
LNL max.	3.98	4.09	4.01	3.88
Time (ms)	119.3	123.3	120.7	107.3

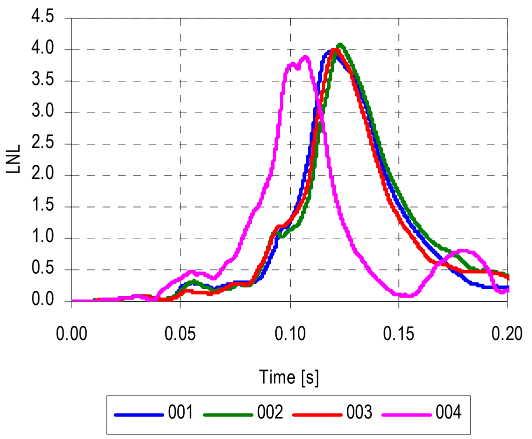

Figure 10. LNL.

The advantages found for this criterion are very similar to the ones found for the N_{km}. It is a criterion based directly on loads, and therefore easy to implement, and does not have the influence of other effects. It is also defined by segments (because of the modulus in the mathematical expression), although it only changes the sign of the reference values for positive and negative data. Besides, the load cell used is the nearest one to the vertebrae where the incidence of injuries is supposed to be higher, and the repeatability shown is very good. On the other hand, the definition by segments presents the same problem already mentioned for the N_{km}, and it has not been able to differentiate between two seats supposed to be very different in terms of whiplash protection.

Neck displacement based criteria (ND)

Viano and Davidsson have proposed a criterion based on the displacements and rotations of the occipital condyle with regard to the T1 vertebra [13]. This criterion, called Neck Displacement Criterion (NDC), was developed from the analysis of the kinematics of volunteers, and is based on two graphs, with the vertical displacement and rotation of the occipital condyle in abscissa and the rearwards horizontal movement of the occipital condyle in ordinate, all of them referred to the T1 vertebra (Z_{OC-T1}, θ_{OC-T1} and X_{OC-T1} respectively). According to the zones occupied by the curves the behaviour is classified as excellent, good, acceptable or poor. This classification was done considering the natural range of motion of both dummies and volunteers.

In order to get numeric values to compare with other criteria, Tencer, Mirza and Huber [28] have defined $Nd_{distraction}$, $Nd_{extension}$ and Nd_{shear} as the quotient between the data used by the NDC criterion and reference values, as described in (8):

$$Nd_{distraction} = \frac{Z_{OC-T1}}{-15mm}$$
$$Nd_{extension} = \frac{\theta_{OC-T1}}{25^{o}} \qquad (7).$$
$$Nd_{shear} = \frac{X_{OC-T1}}{35mm}$$

Using experimental results with volunteers and in vitro tests, and comparing several criteria, they arrived at the conclusion that the best predictors of injury are Nd_{shear}, $Nd_{extension}$ and $Nd_{distraction}$, following this order, instead of other criteria such as N_{km} or NIC, and therefore they recommended the use of criteria based on displacements.

Figure 11 shows the Nd_{shear} calculated for the tests. We can see that the curves of the tests 001, 002 and 003 have a repeatability worse than the previous criteria (0.163), and seat B has been well differentiated (sensitivity of 0.343). Table 6 shows the relative maximums achieved during the formation of the s-shape (100-150 ms).

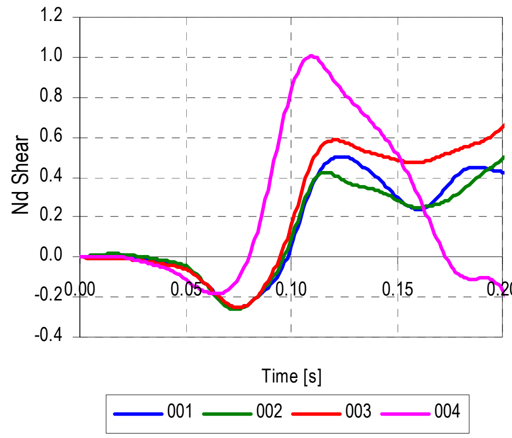

Figure 11. Nd_{shear}.

Table 6. Maximum Nd_{shear}.

Test	001	002	003	004
Nd_{shear} max.	0.50	0.42	0.59	1.00
Time (ms)	124	116	121	109

The main advantage of these criteria is that they represent the real kinematics of the neck, taking into account the whole movement of the head with regard to T1. On the other hand, the main disadvantage seems to be the necessity of displacements measurement using motion analysis software, which, although available, represents additional operations, time of analysis and cost in practise.

Rebound

Several authors ([15], [16], [17] and [18] among others) have reported the risk of injury during the rebound phase when the seat is not able to absorb energy during the impact. This phase can be divided into two different stages. In the first one the dummy receives the released elastic energy from the seat, moving forward freely. The second phase starts when the seat belt begins to act on the dummy, stopping the pelvis and the thorax, and producing a sudden flexion of the neck. Figure 12 shows the data measured with the seat belt load cell during the tests and the rotation of the occipital condyle referred to the T1 vertebra measured by image analysis. We can observe how a violent flexion of the neck is produced when the forces in the seat belt grow. This is reflected also in the loads of the neck, as can be seen in the N_{km} values on this phase (Figure 13). It can be observed also that the maximum values in some of the cases (tests 001, 002 and 003) are considerably higher than the ones registered when observing only the first stages of the movement.

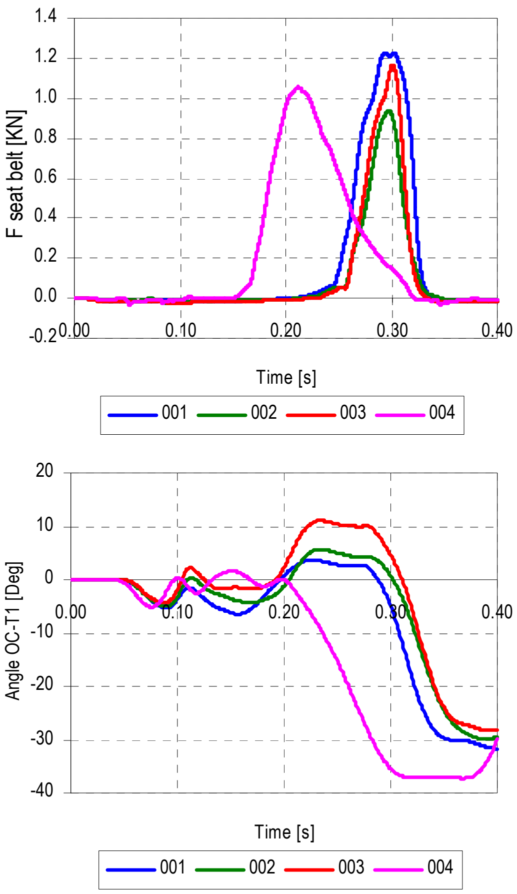

Figure 12. Forces measured at the seat belt and angle of the occipital condyle referred to T1.

At present the capacity of the seats to prevent injuries during the rebound phase is evaluated mainly by measuring the speed of the centre of gravity of the head when it comes back to the position that it occupied at the beginning of the movement (the results of this operation for the fulfilled tests are shown in Table 7). This is supposed to happen just before the seat belt begins to work, so the behaviour of the seat belt is not taken into account. Normally this approximation should be enough, when using seat belts with similar mechanical characteristics on the strap and spool out (the loads are too low so as to be affected by load limiters working at common levels), but this can change in special cases, such as when using pretensioner systems or, as in the case of the seat "B" (test 004), when the points of fixation of the seat belt are fixed to parts of the seat that displace during the impact. Having a look at Table 7 and Figure 13 we can see that, while the rebound speeds are similar for all the tests, the loads on the neck at the rebound are somewhat higher on the fourth test. This fact points to the convenience of reproducing the seat belt configuration in the injury assessment in this phase, at least in the mentioned particular cases.

Table 7. Rebound velocity and time of measure.

Test	001	002	003	004
Rebound velocity (m/s)	3.98	3.96	3.75	4.04
Time (ms)	242	260	259	184

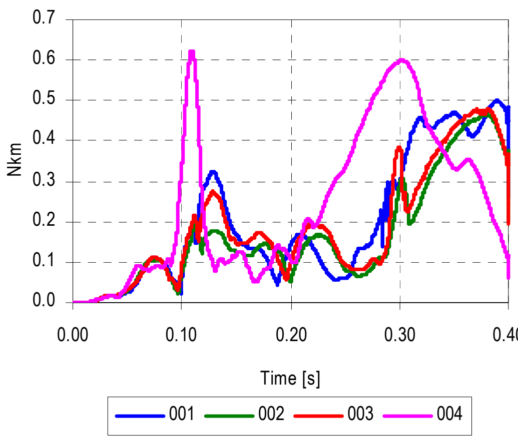

Figure 13. N_{km} extended to the rebound phase.

In conclusion for the rebound phase, a criterion based on loads seems to give more information for injury assessment than the calculation of the speed at a particular point. Considering that the possible injuries of the neck in this phase are better understood (the movements are similar to the ones produced in frontal crashes), a general criterion could be used, such as maximum loads at occipital condyle or N_{km}. Besides, the current method to calculate the rebound velocity supposes normally the use of image analysis, with the practical disadvantages already commented on in the case of displacement based criteria.

RESULTS

The study of these criteria has evidenced weak and strong points in all the cases. The advantages more esteemed by the authors on the underlying concepts of the different criteria have been the following:

1. Capacity to describe the dynamics of the whole neck, taking into account the upper and the lower parts (NIC and ND). Conceptually this should provide a better description of multiphasic states of the neck, particularly the s-shape.

2. Avoidance of distortions due to facts not directly related to the studied injury mechanism (NDC and Rebound speed), such as the angular accelerations found in the NIC (produced by the use of accelerometers at different points) or the mathematical definition in the change of case for the N_{km} and LNL.

3. Facility of calculation (NIC, LNL and N_{km}), avoiding the use of image analysis or complicated algorithms. For practical reasons, the results of the criteria should be available to be analyzed immediately after the test without extra operations.

Other considerations, such as the repeatability or the capability to distinguish different seats are not chosen in the design of the criterion, but are a consequence of the selection of the magnitudes or expressions used in the calculation.

Considering all this, we can draw some guidelines to be applied in the definition of a whiplash injury criterion, focusing on the advantages and avoiding the disadvantages of the studied ones:

1. It should be representative of the dynamics of the whole neck. Taking into account the importance given to the s-shape by the currently accepted injury mechanisms, it should work with values at both ends of the neck in order to be able to detect and quantify this biphasic state.

2. It should avoid the use of accelerations in more than one point, in order to eliminate the sensitivity to the rotations of the seatback.

3. For practical reasons, it should also avoid the use of displacements or velocities measured by image analysis.

4. It would be desirable that its mathematical expression was simple, avoiding the definition in segments.

Taking into account these guidelines and the current instrumentation of the BioRID-II dummy, the simplest solution seems to be the use of the two load cells that the dummy has in the upper and lower ends of neck within only one simple mathematical expression.

PROPOSAL OF A NEW WHIPLASH INJURY CRITERION (WIC)

Having described the previous guidelines, the next step was to determine whether the complex movement of the neck during a rear-end impact could be described by only one mathematical expression using just load magnitudes. As most authors coincide in pointing to the s-shape of the neck as the most probable cause of whiplash injuries, it was decided to look for a function that had a maximum when it happened. As we have seen, the s-shape is a biphasic stage in which the upper end of the neck suffers a flexion at the same time as a hyperextension occurs at the lower end. When using the sign convention stated by the SAE J1733 recommended practice [8], the extension movement is characterized by positive moments in the sagittal axis (Y) of both neck load cells, while the flexion moment is defined by negative moments. Therefore, during the s-shape of the neck, there must be a positive Y moment on the upper end of the neck and a negative Y moment on the lower end (see Figure 14). Taking this into account, the function WIC (Whiplash Injury Criterion) was defined as the most evident solution to the problem:

$$WIC = M_{y_{OC}} - M_{y_{lw}} \qquad (8).$$

In this expression $M_{y\ OC}$ represents the Y moment around the Occipital Condyle (at the upper end of the neck), and $M_{y\ lw}$ represents the Y moment measured at the T1 load cell.

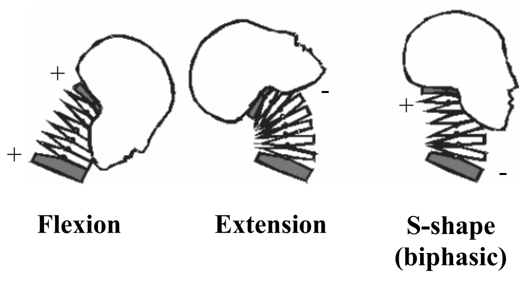

Figure 14. SAE J1733 sign convention for neck moments in "Y" axis.

Figure 15 shows the result of the application of this function to the data obtained in the tests. The maximum values registered were 25.10 Nm, 19.34 Nm and 22.32 Nm respectively for the three first tests (seat "A") and 38.67 Nm for the fourth test (seat "B").

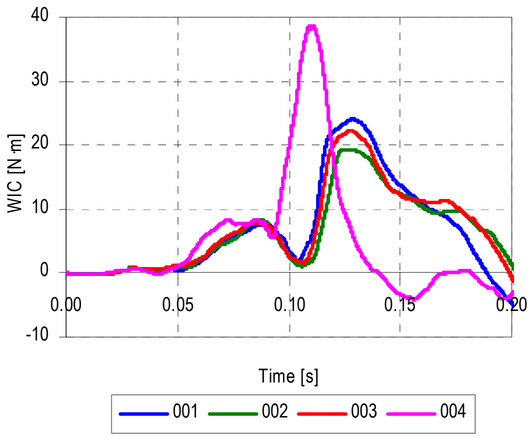

Figure 15. WIC.

After the evaluation of the results we can make the following observations:

1. The curves are very clear and easy to understand. There is a first peak corresponding to the time of the ramping up and spine straightening effects (50-100 milliseconds), coincident with the compression force measured on the lower neck load cell, and a second one, much more marked, during the time when the s-shape is more accentuated (100 to 150 milliseconds).

2. The repeatability in the curves for similar seats (tests 1, 2 and 3) is quite good, having an indicator value of 0.097 (Table 8 shows a comparison of the different values achieved by the indicators of repeatability and sensitivity by the different criteria). We can see also in Table 9 that maximum values for these first three tests happen at very similar times, within a range smaller than that observed by any other criterion.

3. There is a clear differentiation between the curves of the two different seats (sensitivity value of 0.359). The criterion has proved to be sensitive to the seat used and has indicated correctly the inferior seat with regard to neck protection.

4. Looking at the biomechanical aspects, the criterion was designed seeking a function to describe the s-shape, based on the studies that pointed to it as the origin of the more common whiplash injuries. Figure 16 shows a detail of the neck and head of the dummy at the times when the s-shape seemed to be more pronounced visually. We can appreciate that, as expected, the s-shape was significantly more accentuated in the fourth test (the seat was much more rigid than in the other tests, so the thorax accelerated before and the retraction movement happened more violently).

5. Table 9 presents the times in which the different criteria had a maximum, compared to the times when the most accentuated s-shape in the videos were observed (Figure 16). We can see how the proposed criterion was in general, next to N_{km}, the nearest one to the observed times. Besides, it quantified the magnitude of the loads, indicating clearly which seat produced a more pronounced s-shape.

6. Finally, it is easily implemented, neither image analysis being necessary, nor additional instrumentation or complicated algorithms. It can also be easily applied to previously done tests using the version "f" of the dummy (the first one implementing the lower neck load cell) or later.

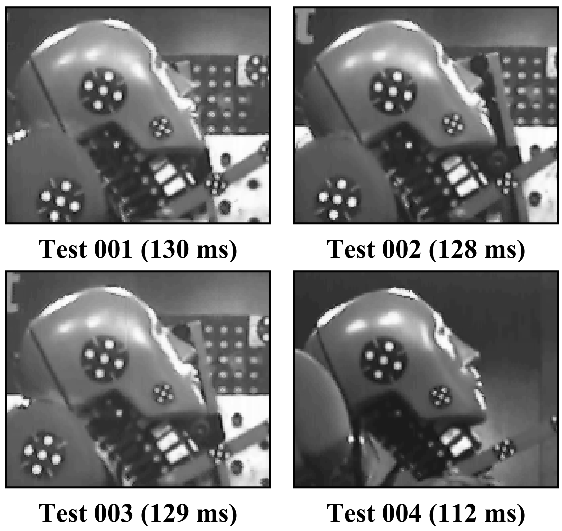

Figure 16. Detail of the head and neck at the time of the most accentuated observed s-shape during the tests.

Table 8. Indicators for repeatability and sensitivity.

	Repeatability (lower better)	Sensitivity (higher better)
WIC	0.097	0.359
NIC	0.084	0.407
N_{km}	0.137	0.307
LNL	0.044	0.250
Nd_{Shear}	0.163	0.343

Table 9. Times of maximum values (ms).

Test	001	002	003	004
Observed S-shape	130	128	129	112
WIC	128.6	126.3	127.7	110.2
NIC	88.6	79.9	81.8	86.7
N_{km}	128.6	112.6	128.1	108.7
F_x (upper)	128.4	121.2	121.2	108.1
F_z (upper)	115.3	123.1	119.9	98.9
LNL	119.3	123.3	120.7	107.3
Nd_{Shear}	124	116	121	109

This study could not have been finished without a critical review of the new criterion. The observed points were the following:

1. This criterion only takes into account the injury mechanisms associated with the formation of the s-shape in the neck. It does not reveal other possible mechanisms such as, for instance, damages produced during the rebound phase or simple hyperextension. To consider them it should be complemented with another criterion for general use (for instance, maximum loads on the occipital condyle or N_{km})

2. The dynamics of the whole neck has only been represented by the two sagittal moments. Of course, this is a simplification, and a more complex criterion could be defined using additional parameters, such as forces, one acceleration or derivative terms. On the other hand, the criterion has shown to be able to detect and quantify the formation of the s-shape, which was its main objective. This could be enough to evaluate the protection for the most accepted whiplash injury mechanisms. Further studies are suggested in order to analyze this point and the convenience of developing a more complete criterion using this one as a base.

CONCLUSIONS

The original aim of this study was the critical review of the commonest injury criteria used to evaluate whiplash protection, analyzing the advantages and disadvantages of each one of them in order to get a more thorough knowledge of their use. A review of the current theories about the motion of the head and the injury mechanisms was done in order to provide a better understanding of the whiplash phenomenon. Four tests with a BioRID-II dummy were fulfilled to provide data to be used in the comparison of the criteria. As a result, some guidelines to define a new criterion were drawn up focusing on the advantages and avoiding the disadvantages of those previously studied. To resume, it should be based on measurements done at both ends of the neck, in order to be able to describe accurately the biphasic state of the s-shape, and, at the same time, it should avoid the use of several accelerometers or image analysis. Therefore, the clearest solution was to use the upper and lower neck load cells at the same time.

Following these directives a new criterion called "WIC" (Whiplash Injury Criterion) was proposed and evaluated under the same conditions that had been used for the study of the other criteria. The results have been very promising, having shown a good repeatability, sensitivity to the seat and capacity to predict and quantify the s-shape of the neck.

In conclusion, some ideas are suggested for future studies:

1. Further evaluation of the new criterion with previously done tests, in order to confirm the first results.

2. Definition of limit values for evaluation of seats, based either on biomechanical studies, on statistical results (taking into account the values given by different types of seats, as done by IIWPG to define their current limits [3]), or using either tests or simulations of real-world accidents with known injury outcomes and recorded crash pulses, as done by Eriksson and Kullgren [29] or Linder et al. [30].

3. More in depth biomechanical analysis, researching into the convenience or not of defining a more complex criterion based on the same guidelines.

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