BIOMECHANICAL SIGNIFICANCE OF THE REBOUND PHASE IN LOW SPEED REAR END IMPACTS.

Markus H. Muser, Felix H. Walz
Working Group on Accident Mechanics, University / ETH Zürich, Switzerland

Harald Zellmer
Autoliv GmbH, Elmshorn, Germany

ABSTRACT

In rear end collisions, the movement of the occupant can be subdivided into three phases: first, a relative rearward movement (towards the seat) occurs. Second, due to seat elasticity and deceleration of the car after the collision, the occupant reverses the direction of movement. Third, the forward-moving occupant is caught by the seat belts. The latter two phases are generally termed 'rebound phase'. It has long been assumed that the rebound phase in low speed rear end impacts is of a minor biomechanical significance where soft tissue neck injuries (also termed whiplash associated disorders or WAD) are concerned. In view of the increasing elasticity and stiffness of modern seat back designs, which lead to an increase of the biomechanical loads imposed on the occupant during the rebound phase, it is deemed necessary to re-examine this assumption.

Results from a total of 25 sled tests with a standard crash pulse, various car front seat models, and using a Hybrid III/TRID anthropomorphic test device are examined with respect to relative displacements and relative displacement velocities of the c.g. of the head and the uppermost thoracic (T1) vertebra. These values are assessed for the different phases of the occupant relative motion, and brought into correlation with mechanical properties of the various seats tested.

Furthermore, we propose ways of quantifying 'elasticity' of the seat back, and discuss possible injury mechanisms occurring during the rebound phase.

Key words: whiplash, rear impacts, energy absorption, injury criteria, seats sled tests

THE INJURY MECHANISMS which may cause soft tissue neck disorders of occupants of the struck car in rear end collisions are multiple and not all of them are fully understood biomechanically (Huelke, 1986; Backaitis, 1993; Walz, 1995 and 2000; Yoganandan, 1998). Although these theories vary considerably with respect to the anatomical location and the nature of damage, they are all based on relative movement of the cervical vertebrae against each other, resulting from different motion vectors of the head and the thorax during an impact.

Since relative motion observed e.g. in volunteer tests often barely exceeds motion corridors of voluntary movement of the head and neck, it is also clear that the time derivatives of the relative motion, i.e. relative velocity and relative acceleration, must play a role in the injury mechanism. On the other hand, there is wide agreement that the absence (or at least minimisation) of relative motion, as attainable by pressing one's head against the head restraint prior to the impact, greatly decreases the loads on the cervical spine.

Most of the hypotheses on injury mechanisms are focused on the first phase of the impact, where the head moves backwards relative to the torso. This is based on the assumption that the kinetic energy involved in the deformation process of the cervical spine is highest in the first phase.

However, depending on various properties of the seat, sometimes violent relative motion of the head vs. the thorax can also occur during later phases of the collision; the aim of this study is to examine the biomechanical significance of these later phases, and to identify methods to quanitfy the protection potential of various seat models with respect to the rebound phases.

We first discuss the possible motion patterns during the entire collision. Subsequently, based on a series of 25 sled tests conducted with various car front seats always subjected to the same crash pulse, we try to identify parameters that can be measured on anthropometric test devices and that can be used to quantitatively assess the occurrence and 'violence' of the motion patterns in these tests. Finally, test results are discussed with respect to the seat design principles that may aggravate or ameliorate relative motion during the rebound phases.

PHASES OF OCCUPANT MOTION DURING LOW SPEED REAR END IMPACTS

FIRST PHASE: RETRACTION/EXTENSION

First, the upper thorax is pushed forward in the shoulder area by the seat back, while the occupant's head, due to its inertia, remains at its original location in space, since it is not in contact with any parts of the car. In this phase, the head (due to its moments of inertia) also does not rotate, and therefore, together with the purely translational forward movement of the thoracic column, the upper cervical spine is forced into an anteflexion and the lower cervical spine into an extension (Penning 1994). This S-shaped deformation of the cervical spine has been observed in experiments with special dummy necks (Svensson 1992) as well as in volunteer and cadaver tests (McConnell 1993, Svensson 1998, Eichberger 1996, Geigl 1994, Castro 1997, Ono 1997 and 1998, Yoganandan 1998a and 1998b, Wheeler 1998).

In a second sub-phase, due to the tangential components of the forces exerted onto the head through the cervical spine, the head starts to rotate in a rearward direction, and the entire cervical spine will eventually participate in a retroflexion. The upper cervical vertebrae, therefore, change their orientation from the previous anteflexion into a retroflexion, albeit with a larger bending radius. At this point in time, i.e. at the end of the 's-shape' phase and the beginning of the extension, the smallest bending radii can be observed along with the highest bending moments of torque and compressive and shearing force levels. A head restraint which is positioned adequately can mitigate the violence of these forces considerably.

The retraction/extension phase is concluded at the point in time when the rearward movement of the head is stopped by the head restraint cushion (assuming a head restraint is present and correctly adjusted). Depending on the initial head restraint position, head extension angles may vary widely, but at least a small s-shaped retraction is observed in almost cases; for comfort reasons, the head restraint can hardly be placed horizontally nearer than 40 - 50 mm from the head surface. Even with automatically positioned head restraints (Muser 1994), a distance of 30 mm seems to be the minimum.

SECOND PHASE: FORWARD MOVEMENT

After the head has been caught by the head restraint, its motion relative to the car is reversed, i.e. it begins to move forward. In a completely inelastic system, both head and thorax would now stop their movement relative to the car, and the dummy would remain 'stuck' to the seat. Rebound only occurs because the energy stored in e.g. the bent seat back, the compressed foam materials of seat back and head restraint, and other elastic components of the seat, is transferred back to the respective body parts. It should be noted here that, in contrast to the human body and also advanced dummies, the Hybrid III dummy neck and, to a somewhat lesser amount, also the TRID neck, elastically stores a certain amount of energy as well. If the dummy head is brought into an extension, a righting moment that would bring the head back to its original position always exists.

The analysis of the relative motion between head and thorax becomes complicated at this time, since the head does not necessarily inverse its movement at the same point in time as the thorax. If we assume similar degrees of elasticity in the head restraint and the seat back, a reversal of movement of the thorax prior to the head will result in a prolonged retraction phase, since the head is still moving backwards while the thorax has already begun moving forward. This is the scenario most often seen in our tests, and was also observed by States (1970), and by Svensson (1993) who noticed an 'increasing violence of the head-neck motion'.

On the other hand, if the head precedes the thorax in the reversal of movement, the retraction phase will be shortened; less total retraction occurs, but the relative forward velocity and acceleration of the head vs. the thorax may be higher than in the first case.

Depending on the ratio of the energy absorption capabilities of the head restraint and the seat back, the two scenarios mentioned will yield more or less significant relative motion between head and thorax.

THIRD PHASE: PROTRACTION/FLEXION (BELT RESTRAINT)

Since completely inelastic seats do not exist, the occupant will always move forward relative to the car in a third phase. After having reached a position longitudinally equal to the position prior to the collision, the belt system begins to restrain the occupant. Belt forces begin to act on the pelvis and the thorax. By restraining the thorax via the clavicle and the rib cage, the upper part of the thoracic spine is stopped while the head continues to move forward. The result is an 'inverse s-shape' of the cervical spine, now consisting of an anteflexion of its lower part and an extension of the upper part. However, in contrast to the first (retraction/extension) impact, the effect is less pronounced because the restraining forces are damped by the thoracic cage and, in addition, for geometric reasons more vertebrae can participate in the flexion and extension motion, thereby allowing for larger bending radii and smaller loads on the individual vertebrae and intervertebral structures.

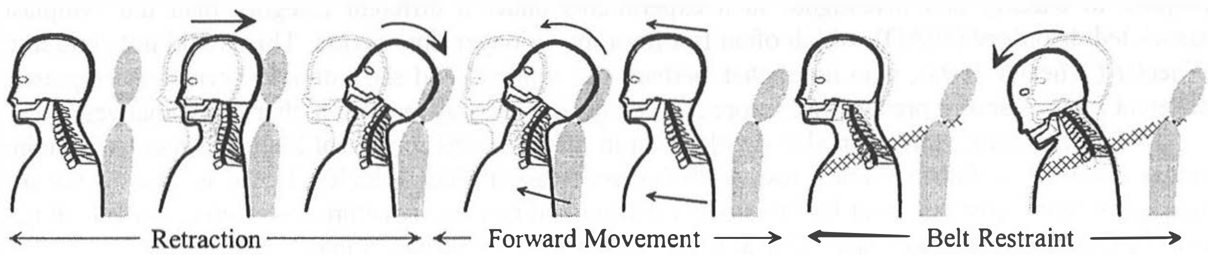

Figure 1: Three phases of occupant movement during rear end impact.

RAMPING EFFECTS

During the first phase of the impact, acceleration forces from the seat back acting on the thoracic spine provoke a straightening of the spine and eventually also a 'ramping' upwards movement. Especially with a hard seat back, this generates a compressive force on the vertebrae, which, in turn, facilitates shearing movement of the vertebrae between each other (Yang, 1997). This 'sliding' movement leads to an impingement of the zygapophyseal joints (facets of the intervertebral joints) (Ono 1998a). The knowledge of this injury mechanism probably has implications for the therapy of neck trauma related headache and neck pain: the neurosurgical denaturation of the nerves innervating the painful joint has been proven to be effective (Bogduk, 1998). Anthropomorphic test devices based on the Hybrid III thorax cannot reproduce this effect, because the parts modelling the 'thoracic spine' do not reproduce the thoracic kyphosis and are, under the relatively low loads discussed here, virtually undeformable.

TEST METHOD

The assessment of the biomechanical protection potential of current car seats is strongly impeded by the fact that, as of today, no standardised test protocol for low speed rear end impacts exists. If applied, such a protocol would allow comparison of current 'standard' seats against each other. Furthermore, the potential of advanced protective devices (Lundell 1998, Jakobsson 2000, and Wiklund 1998) that have entered the market recently, could be assessed in a direct comparison using a standardised test.

In a collaboration between the Institute of Vehicle Safety (GDV), the Working Group on Accident Mechanics of the University and ETH Zürich, and Autoliv (Germany) GmbH, a test protocol designed to alleviate this limitation has been developed (Hell 1999, Muser 1999). In order to assess the practicability of this protocol, as well as to compare the performance of a wide spectrum of front seats of currently circulating cars, 25 sled tests have been performed. Subtracting those tests that were performed to assess repeatability or to experiment with different crash pulses, a total of 20 tests remains that were executed with the same crash pulse, dummy type and position, and measurement parameters.

The complete specification of the test set-up, which was enhanced by additional suggestions from the TU Graz (on behalf of the Brite-Euram 'Whiplash' project) can be found at http://www.biomed.ee.ethz.ch/~agu/pdf/ritp006.pdf .

CRASH PULSE

A trapezoidal crash pulse with rise and fall times of 10 - 20 ms and an average sled deceleration of 6 ± 1 g was used in all tests. The sled delta-v was set at 15 ± 1 km/h. The sled deceleration pulse was chosen such that it would represent a delta-v range where a majority of rear end collisions occur (Hell 1998). Most experts today agree that the threshold level below which, in a normal case (i.e. no prior damage to the cervical spine, normal seating position, age below 55 yr.), injuries to the cervical spine are deemed improbable, lies between a delta-v of 10 and 15 km/h for the struck car (e.g. Castro 1997). With respect to the delta-v value, the sled tests represented the upper border of this range and thus were in our opinion well suited to expose the advantages and disadvantages of different seat designs. Some authors also report acute 'whiplash' symptoms following volunteer tests conducted at much lower speeds; however, these symptoms lasted, to our knowledge, in all cases only for a few hours. Therefore, we propose to classify and investigate such experiments under a different category than the 'whiplash associated disorders' (WAD), which often last for a much longer time period. This would not contradict Wheeler (Wheeler 1998), who noted that 'perhaps our minimal and short duration symptoms represent different injuries and/or precursors to more severe injuries that may result at higher speed changes'.

On the other hand, the mean sled deceleration in our tests was somewhat higher than corresponding values obtained in full size crash tests with (in most cases) older vehicles. However, due to the no-damage or repair cost tests that lead to stiffer car front and rear-end structures, acceleration levels in real world collisions with newer cars arrive at a level of 6 g (or even surpass it in the near future).

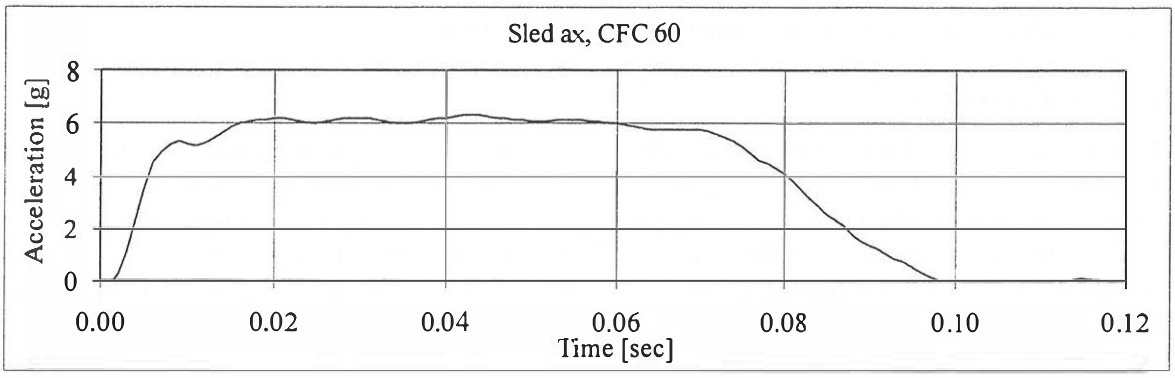

Figure 2: Typical sled deceleration pulse.

ANTHROPOMETRIC TEST DEVICE

The biofidelity of anthropometric test devices, when used in low speed rear end impact tests, has been the subject of a large number of publications. Most authors agree that the Hybrid III series of dummies does not offer an adequate biofidelity for use in rear end impact tests. This is mainly due to the relatively high stiffness of the neck, but also due to the design of the torso that does not allow deformation similar to a human thoracic spine. Efforts to build a dummy specially for use in rear end impact, such as the BioRID dummy (Davidsson 1998) or the dummy resulting from the European (Brite-Euram) Whiplash project (van den Kroonenberg 1999) have recently been completed, but their availability and wide-spread use in test facilities is considered to be several years away. Therefore, we chose the compromise of using a Hybrid III 50th percentile dummy equipped with a softer, segmented rear end impact neck (Thunissen 1996) for the tests discussed here. (Some additional tests, whose results are not presented here, were also conducted using a BioRID dummy. A detailed comparison of the two ATD's is beyond the scope of this paper; however, we note that, even though the absolute values of e.g. NIC measurements differed between the two ATD's, the Hybrid III/TRID combination was useful for our purposes, since mainly relative values, i.e. comparisons between tests with the same dummy and different seats, were evaluated).

TEST SET-UP

The seats were directly mounted on a sled using welded fixtures bolted to the sled surface. Recliner and seat base adjustments were made to ensure a 25° ± 2° torso line and a 12° ± 1° seat ramp using a H-point machine according to SAE J 826. A foot rest that offered a 45° inclined plane for the positioning of the dummy feet was also mounted onto the sled.

In addition to the standard instrumentation, one additional accelerometer was placed on the neck bracket in a position corresponding to the first thoracic vertebra (T1) for the measurement of the neck injury criterion (NIC, Boström 1996), and another biaxial accelerometer was mounted on the top of the dummy head in order to quantify head rotation in conjunction with the head c.g. accelerometer. In addition, the head-neck joint (C0) was instrumented with a six-axis load cell (upper neck load cell). In 7 tests, a lower neck load cell inserted in place of the neck bracket was also used.

The dummy was placed on the seat according to the procedures set forth in ECE R 94. The neck bracket was adjusted to ensure an exact alignment of the dummy head accelerometer's x-axis to the horizontal. Such an adjustment is essential in order to prevent artefacts in the measurement of the neck compressive forces if the head is inclined forward prior to the collision. In the cases where the presence of the lower neck load cell prevented this adjustment, an appropriate steel wedge was inserted between the load cell and the TRID neck in order to ascertain the correct head position.

Mainly in order to secure the dummy from completely falling off the seat during the test, a seat belt was mounted using a standard 3-point configuration and geometry. Since we did not implement the same belt type and geometry as in the respective target vehicles, dummy measurements during the third phase (protraction/flexion due to belt restraint) can be interpreted only to a limited extent.

Finally, head restraints were (if possible) brought into a position where the top of the head and the top of the head restraint aligned vertically.

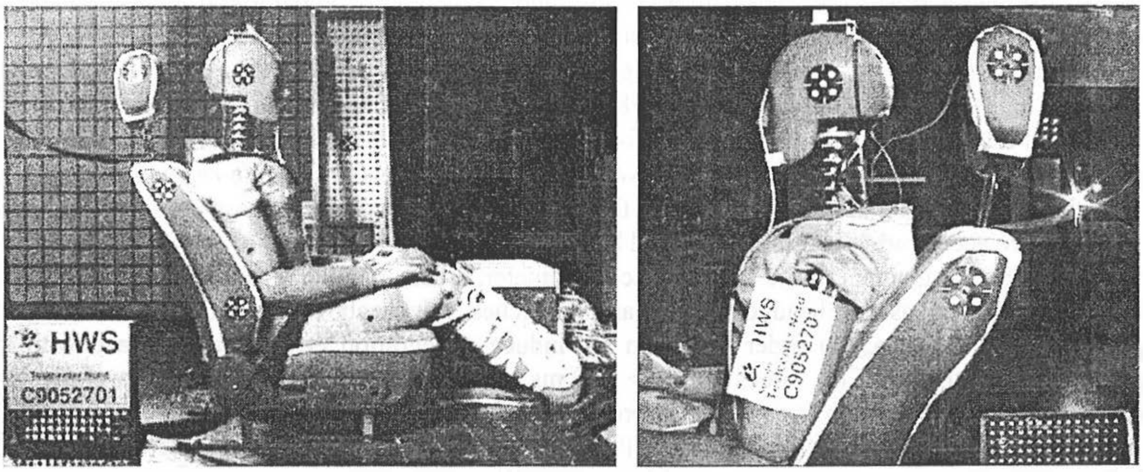

Figure 3: Test set-up (seat M), identical for all tests described here. A stationary high speed video camera was used for an overview of the entire test rig. A second camera was fixed to the sled, and positioned to show a more detailed view of the head, neck, and upper thorax.

SEATS

A total of 17 different seat models were tested. In two tests, head restraints were equipped with an additional cushion designed to lower the horizontal distance of the head vs. the head restraint; in one test, a whiplash protection device (based on a cantilever that uses the inertial force of the thorax to push the head restraint forward and upward) was blocked in order to compare results against the test with a functioning device. Out of the 17 different seat models, 15 seats were new, while 2 seats (L,M) were taken from used vehicles. Three seat models were equipped with systems that move the head restraint forward and upward during the collision, and one model was equipped with a recliner that employs a special kinematic joint in order to deform in a controlled way during the acceleration phase.

PARAMETERS FOR THE COMPARISON OF THE PROTECTION POTENTIAL

For a comparison of the biomechanical protection potential of the various seats, a reduction of the numerous measurement channels recorded during the test down to approximately 4 - 5 significant resulting values is necessary. For most of the physical quantities that can be measured on a dummy during a test, biomechanical tolerance criteria that indicate serious injuries (e.g. fractures of vertebrae, ligament ruptures, cerebral concussions) have been defined. Indications how to reduce these tolerance thresholds to make them apply for (minor) soft tissue neck injuries are much more difficult to find. We can therefore only use measurements of e.g. shear forces and moments of torque in the neck for a comparison of the seats against each other, and not for conclusive classification of the seats into 'good' and 'bad' groups.

NECK INJURY CRITERION (NIC)

The neck injury criterion (Boström 1996) is an exception to the remarks made above, since it was explicitly developed and validated to quantify the risk for whiplash associated disorders. The formula for calculating NIC as a function of time:

$$NIC(t) = a_{rel}(t) \cdot 0.2 + (v_{rel}(t))^2$$

takes into account the relative acceleration and velocity of the highest (occipital condyle) and lowest (T1/C7) point of the cervical spine. NIC_{max} is the maximum of the NIC(t) curve during the retraction phase as described in (http://www.biomed.ee.ethz.ch/~agu/pdf/Nic.calc.005.pdf). The idea of this criterion is based on pressure measurements in the spinal canal in animal experiments where the head of an anaesthetised animal was retracted with a velocity and acceleration comparable to measurements on human necks during rear end impacts. It was found that the pressure gradient that develops in the venous and cerebrospinal fluid in the spinal canal during the retraction could possibly injure nerve root ganglia (Aldman, 1986, Boström, 1998; Svensson, 1998). In correlation with accidentological and theoretical studies, the limit above which a significant risk for WAD injuries exists was set at 15 m^2/s^2.

It should be noted that, due to the way NIC was derived, it applies only for the *retraction phase*, i.e. where both relative acceleration and velocity of the head vs. the thorax have a rearward direction. Also, since the relative acceleration is usually acquired by subtracting the x-axis signal of the T1 accelerometer from the corresponding signal of the head c.g. accelerometer, a considerable error is introduced into the NIC(t) curve as soon as the head extension angle reaches values above 20 - 30°. In addition, the integration over time of a_{rel} in order to obtain v_{rel} induces integration errors that increase with the integration period; thus, NIC(t) values with t > 150 ms approx. need to be interpreted carefully with respect to those facts. Research is currently in progress to define NIC(t) that is valid for the assessment of head-neck relative movements in frontal collisions. For this situation, the original NIC(t) formula needs to be adapted with respect to the sign of the v^2 term (Boström 2000):

$$NIC(t) = a_{rel}(t) \cdot 0.2 + v_{rel}(t) \cdot abs(v_{rel}(t))$$

NIC_{min}, the minimum of the NIC(t) curve, could possibly be used to quantify 'violence' of the head-neck motion during the second phase of a rear end collision as well as for frontal collisions; however, in view of the aforementioned problems we faced in determining a correct NIC(t) curve, we include these values for reference only.

NECK MOMENTS AND SHEAR FORCES

During a rear end collision where the dummy is in a standard position and its head hits the head restraint centrally, the only significant moments of torque are measured on the y-axis (lateral axis) of the load cell. Significant forces can be observed along the z-axis (vertical), i.e. compression/tension of the neck, and along the x-axis (forward), i.e. shear forces between the vertebrae. With a correctly positioned TRID-neck, compression of the neck is very small, since, as mentioned above, ramping due to straightening of the thoracic spine does not occur. Tensile forces along the z-axis are mainly due to rotational effects. Thus, axial forces measured in the TRID neck, especially compressive forces, have little correlation with those thought to occur in human beings.

In a first step, we have therefore taken into account only the moments of torque on the y-axis, i.e. those moments that appear due to flexion/extension of the neck. However, depending on the amount of 's-shape' that occurs (depending on the seat in question as much as on the kind of dummy neck used), shear forces may also play an important role. The simultaneous presence of high shear forces and moments of torque would appear to be most injury critical. Thus, in a similar way to the Nij criterion (Kleinberger 1998) that combines moments of torque and compression/tension force by normalising the values with corresponding intersection values and then adding the two terms, shear forces and moments of torque could be combined:

$$Nkm = \frac{F_x}{F_{int}} + \frac{M_y}{M_{int}}$$

with the intersect values:
$F_{int} = 860N$ *(negative) / 1200N (positive)*
$M_{int} = 57Nm$ *(Extension) / 80Nm (Flexion)*
(a positive shear force occurs when the head is pressed forward relative to the thorax)

'k' and 'm' in the above formula are indices into a 2x2 matrix, i.e. in analogy with the Nij criterion, 4 combinations (forward/rearward shear force and extension/flexion moment) must be made and interpreted separately. Using 'e' for extension and 'f' for flexion as the first index and 'a' for anterior (positive) and 'p' for posterior (negative) shear force, the 4 combinations would be N_{ea}, N_{ep}, N_{fa}, and N_{fp}. In our tests, the simultaneous occurrence of positive (anterior) shear forces and extension moments (N_{ea}) did not occur, nor did the combination of posterior shear forces and flexion moments (N_{fp}). Thus, only two (N_{ep} and N_{fa}) combinations need to be examined. It should be noted that the intersect ('limit') values used for this formula are not validated as yet. Thus, this criterion is included for reference only.

Lesions caused by excessive moments of torque would, in principle, occur whenever a corresponding tolerance limit is exceeded, independent of the three phases of relative head-neck motion defined above. Such values could thus be used to quantify the injury risk during the rebound phase; however, moments of torque are highly dependent on the bending stiffness of the neck used, and thus, in the case of the TRID-neck, might be overestimated in comparison to a human neck.

QUANTIFYING 'ELASTICITY' AND 'REBOUND' IN THE SECOND PHASE OF MOTION

In view of the fact that, on the one hand, NIC is so far only validated for the first (retraction) phase, and, on the other hand, problems exist in the interpretation of neck shear forces and moments of torque during the second and third phase, we chose to primarily use the velocities measured at the head c.g. and at the position of the first thoracic vertebra (T1) relative to each other and relative to the sled to quantify biomechanical influences of the elasticity of the seat. The following values have been measured by analysing high speed video sequences of the tests:

- By comparing velocities before and after contact with the head restraint (head) and seat back (T1), the amount of kinetic energy dissipated in the deformation of the seat back can be estimated; the following formula delivers an approximation of the ratio between the kinetic energy of e.g. the head before and after head restraint impact:

  $$el = \left[\frac{v_{peak}}{v_{peak}'}\right]^2$$

  vpeak = peak velocity of e.g. the dummy while moving towards the seat back,
  vpeak' = peak velocity while moving away from the seat back

- Time difference between zero intersection points of the dummy head and T1 velocity vs. time. By subtracting the time point when the velocity of the chest relative to the sled changes its sign from the corresponding time point for the head, the degree of additional rearward x-displacement caused by elasticity (cf. section 'Second phase: forward movement') can be characterised. In Figure 4, this parameter is denoted Δt.

- Velocity of T1 relative to the sled at the point in time where the retraction of the dummy head vs. T1 reaches its maximum. Also by this value, the degree of additional rearward x-displacement of the head c.g. caused by elasticity (cf. section 'Second phase: forward movement') is characterised. In Figure 4, this parameter is denoted v_{ch}.
- Peak relative velocity along the x-axis of the dummy head vs. T1 [$v_{rel}+$] after contact with the head restraint: this parameter characterises the motion in which the head moves from its most retracted position at the beginning of the second phase back to its original position. The corresponding value from the first phase [$v_{rel}-$] is also included for reference. This parameter is expressed in a co-ordinate system that remains fixed to T1, i.e. the rotation of the chest due to seat back yielding is taken into account. In the following tables, this parameter is denoted $v_{rel}+$.
- Velocity of the dummy head and of T1 relative to the sled at the beginning of the third phase, i.e. just before the dummy is caught by the seat belt: since we do not have a realistic belt geometry and therefore do not have reliable measurement results for e.g. belt forces or neck moments of torque during the third phase, we use the velocity at which the dummy is being caught by the belt system. Nilsson (1994) used the chest velocity (similar to T1) for this purpose. In Figure 4, these parameters are denoted $v_{head}+$ and $v_{T1}+$

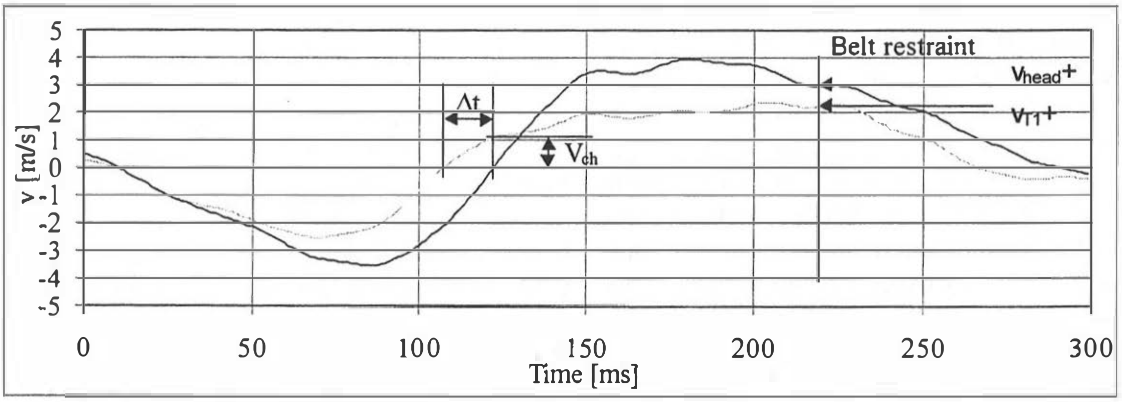

Figure 4: Velocity of the dummy head (black) and T1 (grey) relative to the sled along the horizontal axis, test Q. Graphical definition of the aforementioned parameters Δt, v_{ch}, $v_{head}+$, $v_{T1}+$ as presented in the tables below. A positive velocity signifies movement in the car forward direction (away from the seat back).

- Dynamic and remaining seat back deflection: the seat back angle of deflection was determined from the video sequences using film targets mounted on the seat back, 'dynamic' deflection denotes the maximum angle reached during the test, while 'remaining' deflection denotes the deflection angle measured on the seat after the test, i.e. due to plastic deformation of the seat back structures or the recliner mechanism.

TEST RESULTS

In the following tables, the results from 20 sled tests with respect to the measurement parameters described above can be found. The tests are denoted only with capital letters A..T. Table 1 shows measurements mainly related to the first phase of the impact. The values presented there provide a reference for a 'conventional' protection potential assessment, i.e. without particular consideration for the rebound phases. In contrast to this, Table 2 provides results that could possibly lead to a more detailed assessment of the rebound phases and, thus, the elastic properties of the seat systems.

In 8 tests, a time delay (Δt) between the reversal of movement of T1 and the head greater or equal than 10 ms was observed. 10 ms may seem a short time, but, in the case of test G, the upper thorax had already accelerated to a speed of 1.6 m/s (parameter v_{ch}) in the forward x-direction at the time when the head reversed its motion. Δt and v_{ch} have to be interpreted together, since at least in one case (seat D) relatively high Δt values went along with low v_{ch} measurements.

Seats equipped with protection systems (B,J,K,N) showed low or even (B) negative Δt values. In the case of the moving-head restraint-systems (J,K,N), this was easily understandable because the system is based on a balancing effect of the forces acting on the upper thorax and the head. The recliner-based protection system (B) achieved the same goal through the kinematics of the recliner mechanism and its energy absorption capability.

A high initial head-to-head restraint distance of approx. 100 mm produced, in test R, the effect that the head, although exhibiting a high extension angle, barely touched the head restraint, since T1 was moving forward already 16 ms before the head. In other cases with similar initial distance, but lower Δt values (e.g. M) this effect was not observed.

The seats that were equipped with additional head restraint cushions (G, S) showed consistently better results for the first phase of motion than the original seats (F and O, respectively), in that NIC_{max} was reduced from 22.7 to 15.5, and from 23.1 to 16.6, respectively. However, due to the elastic properties of the material used for the cushion, a higher relative velocity of the head vs. T1 in the rebound phase was observed in comparing tests F and G.

Test	NIC_{max}	d [mm]	M_y [Nm]		N_{km}		Seat deflection [°]	
			Flex.	Ext.	N_{ep}	N_{fa}	Dynamic	remaining
A	23.4	*95*	18.4	13.8	*0.79*	*0.42*	-9.3	1
B*	**10.3**	*45*	**6.6**	**6.7**	*0.06*	*0.22*	-12.3	-3
C	20.9	*55*	**15.8**	23.3	*0.48*	*0.45*	-20	-7.2
D	19.9	*85*	**10.7**	20.1	*0.41*	*0.38*	-12	-1
E	16.2	*50*	**8.8**	**5.3**	*1.26*	*1.03*	-11	-2
F	22.7	*120*	17.9	15.3	*0.31*	*0.45*	-14	-4
G	15.5	*50*	**11.8**	17.9	*0.37*	*0.39*	-14	-3
H	**13.5**	*45*	**8.7**	16.6	*0.38*	*0.28*	-18.4	-9.3
I	24.2	*58*	28.3	21.8	*0.61*	*0.61*	-6.3	0
J*	17.1	*115*	22.4	**9.3**	*0.25*	*0.47*	-10.9	-1
K*	**12.5**	*60*	20.5	**10.6**	*0.3*	*0.43*	-16.3	-6.4
L	18.2	*90*	**7.7**	30.3	*0.65*	*0.27*	-15.2	-6.2
M	16.7	*110*	**6.0**	**5.8**	*0.35*	*0.21*	-11.6	-5.7
N*	**10.2**	*75*	**9.2**	**2.4**	*0.07*	*0.17*	-13.1	0
O	23.1	*100*	**13.9**	24.5	*0.56*	*0.4*	-10	-1
P	17.3	*65*	28.3	14.8	*0.72*	*0.32*	-7.5	0
Q	22.3	*65*	23.4	17.8	*0.55*	*0.35*	-9.5	1
R	18.1	*95*	19.8	12.2	*0.36*	*0.35*	-13.2	-2
S	16.6	*55*	**9.8**	14.3	*0.38*	*0.26*	-8.3	0
T	**11.5**	*30*	**10.7**	**6.9**	*0.41*	*0.25*	-11.8	2

Table 1: Measurement results of tests A...T, parameters relating to the first phase of the impact. Letters with an asterisk (*) denote seats with 'whiplash protection systems'. NIC_{max} is the maximum NIC value during the first (retraction) phase, i.e. while the head is moving backwards relative to the torso. d is the initial head-to-head restraint distance. M_y is the peak moment of torque around the y-axis at the occiput. N_{km} denotes the combined shear force – moment criteria as defined above. In test G, an additional head restraint cushion was mounted to seat F. In test S, the same was done for seat O. In test O, the protection system of seat J was blocked.

With respect to the third phase (Protraction/Belt restraint phase), forward velocities of T1 relative to the sled (v_{T1}+) between 2.1 m/s (tests I, O, S) and 3.7 m/s (test C) were observed. These values correspond to 7.5 and 13.3 km/h, respectively. The forward velocities of the head relative to the sled (v_{head}+) in the same phase ranged from 1.2 m/s to 4.2 m/s, covering a larger range than v_{T1}+ because of the rotation of the entire dummy as well as the rotation of the head vs. the thorax.

Test	Δt [ms]	v_{ch} [m/s]	$v_{rel}+$ [m/s]	NIC_{min}	$v_{rel}-$ [m/s]	$v_{T1}+$ [m/s]	$v_{head}+$ [m/s]	el head	el T1
A	13	1.0	1.8	*41*	*2.9*	2	4.2	*0.9*	0.8
B*	-7	-0.1	0.9	*23*	*2.1*	1.5	2.6	*0.4*	0.5
C	14	0.6	1.8	*26*	*3.7*	2	3	*0.5*	0.3
D	14	0.4	2	*34*	*3.3*	1.8	2.9	*0.7*	0.4
E	-7	-0.1	1.4	*30*	*2.5*	2	2.7	*0.6*	0.6
F	9	1.0	1.6	*30*	*3.1*	2.1	2.9	*0.7*	0.5
G	10	1.6	2.2	*30*	*3*	2.1	3.8	*1.0*	0.5
H	-11	-0.2	1	*22*	*2.9*	1.2	1.6	*0.3*	0.3
I	5	0.5	1.8	*45*	*3*	1.7	2.9	*0.8*	0.7
J*	3	1.0	1.2	*17*	*2.3*	1.9	2.5	*0.6*	0.8
K*	6	1.1	1	*16*	*1.7*	1.5	1.8	*0.7*	0.6
L	18	0.8	1.3	*19*	*2.5*	1.3	1.2	*0.5*	0.3
M	7	0.5	1.1	*20*	*2.6*	2.4	3.6	*0.5*	0.4
N*	3	0	1.2	*20*	*2*	1.8	2.5	*0.6*	0.9
O	10	0.8	1.7	*34*	*2.6*	2	2.5	*0.9*	0.8
P	7	0.8	1.2	*24*	*2.4*	1.5	2.3	*0.8*	0.7
Q	14	1.5	2.1	*21*	*2.4*	2.2	2.9	*1.2*	0.8
R	16	1.2	2.3	*21*	*2.4*	1.7	2.7	*1.1*	0.8
S	7	1.0	1.6	*25*	*2.3*	1.9	2.3	*0.9*	0.8
T	0	0.4	1.5	*22*	*1.6*	2.6	3.5	*1.1*	1.3

Table 2: Results of tests A...T, parameters relating to the second and third phases ('rebound'), as defined above. ***NIC_{min}*** *describes the relative motion of the head vs. thorax during the* ***rebound phases where the head moves forward relative to the thorax****. Letters with an asterisk (*) denote seats with 'whiplash protection systems'. In test G, an additional head restraint cushion was mounted to seat F. In test S, the same was done for seat O. In test O, the protection system of seat J was blocked. Since the results presented in this table were acquired using measurements on digital high speed video, their accuracy may be considerably lower that e.g. acceleration measurements. We estimate the error to be in the range of ± 10 %. Values printed in italic are included for reference only.*

DISCUSSION

The classification of the various seat models into groups of seats that perform in a similar way proved to be quite difficult; a differentiation into 'yielding' and 'unyielding' seats was not sufficient. 'Unyielding' seats showed high elasticity in all cases. Some 'yielding' seats, on the other hand, deformed plastically (e.g. seats B, H) and thus showed promising results also with respect to the rebound phase. In contrast to this, other seats showed a considerable amount of elasticity (e.g. seat C) that led to high rebound velocities. Kinetic energy was, during the first phase of the collision, not only stored in the recliner and the seat back frame; also the mechanical links between the seat and its forward anchorage points on the car showed the potential of storing energy through spring loading, e.g. in height adjustment mechanisms. In these cases, a rotation of the entire seat, not only the seat back, was observed during the tests.

Measuring the initial head-to-head restraint distance in order to compare seat designs is useful as a generic classification method (Pedder 1995, IIHS 1997); however, the biomechanically relevant 'initial distance' would have to be composed of the geometric distance and additional parameters such as elasticity and deformability of the seat materials. In addition, the disadvantage of a high initial distance can be (at least in part) made up for by an advanced protection system, as with seat J (NIC_{max} 17.1 in comparison to 23.1 with the protection system disabled, test O). In this comparison, the energy absorption capabilities (el head, el T1) were the same, while, with the active system enabled, a lower relative velocity of the head vs. T1 ($v_{rel}+$) was observed. By blocking the protection system, the seat back was made somewhat stiffer than a (hypothetic) seat back designed entirely without such a mechanism.

For such a seat back, a lower NIC value would probably result, and the comparison would appear less unfavourable than with the blocked mechanism. Thus, low initial distances, in most cases, predicted favourable results in our tests, whereas higher initial distances did not necessarily predict 'bad' results.

It is known that, in frontal collisions with velocity changes (delta-v) of about 13 km/h, lesions of the cervical spine do not occur in belted occupants (for example, public mini-sled tests to demonstrate seat belt effectiveness have been (and still are being) performed in various European countries with thousands of volunteers without producing injuries. On the other hand, belt marks have been observed in occupants of vehicles that were subjected to rear end impacts with delta-v values below 15 km/h. This apparent contradiction can be resolved by examining the velocity of the occupant relative to the belt system; e.g. in test C, the occupant exhibited a relative velocity of 3.7 m/s towards the belt system (very high belt 'slack'), while, in a frontal crash with an initial velocity of 3.7 m/s, the relative velocity of the occupant at the point in time when the restraint force of the belt sets in would be much lower. It was also observed that the velocities of the head vs. the sled were consistently higher than those of T1, in some cases (A,C) even higher than the sled delta-v itself; this might be explained by rotations of the dummy around the pelvis, and additional rotation of the head around a point near T1. Again, as opposed to the frontal impact situation, at the time when belt restraint forces set in, the head already exhibited a velocity forward relative to T1, thus leading to a more violent protraction movement in the third phase.

In principle, belt pretensioners could be triggered also in rear end impacts, and thus serve to alleviate biomechanical loading during the third phase by taking up some of the huge 'belt slack'. In fact, some manufacturers today adapt this solution. A possible disadvantage of this idea is, apart from repair cost considerations, the problem that high output pretensioners may well be capable of taking up all of the belt slack at the point in time when the occupant is in his rearmost (most reclined) position; if, at this point in time, the retractor is blocked, the spring-loaded seat back will act against the occupant's thorax which in turn is restrained by the belt, and considerable forces could thus be exerted on the rib cage. Further studies should investigate whether such forces may amount to biomechanically dangerous levels or not.

It can be observed in Table 2 that NIC_{min} values, i.e. NIC values relevant to a forward motion of the head relative to the thorax, do not correlate very well with $v_{rel}+$ values, although the (squared) relative velocity appears in the NIC formula. This means that the NIC(t) curve is dominated by the relative acceleration term. Again, low $v_{rel}+$ values go along with low NIC_{min} values, whereas high $v_{rel}+$ values do not necessarily indicate high NIC_{min} values (e.g. tests Q and R). These interrelations need further investigation.

COMMENTS TO SELECTED CASES

Seat H was an example of a relatively low-cost seat with no special whiplash protection devices. Due to a low initial head-to-head restraint distance, and due to a considerable amount of plastic deformation in the seat back, this seat showed a good biomechanical protection potential in the first phase (NIC 13.5) and also in the later phases, i.e. relative velocity in the second phase and the ratio of rebound energy both werw in a low range. This showed that, by allowing energy absorption in the seat back, good biomechanical results were obtained with very simple technical measures; however, it should be noted that the additional load on the seat back that would occur e.g. with a heavier occupant or a higher acceleration load of the car would soon lead to a collapse of this seat back.

In contrast, seat I showed a very high stiffness and no remaining deformation, i.e. high elasticity. Although the initial head-to-head restraint distance was relatively low, stiff structures in the shoulder area led to high T1 accelerations and, accordingly, a high (24.2) NIC_{max} in the first phase. Moments of torque and also shear forces reached both their maximum and minimum values between the first and second phases (t=100..120 ms), i.e. during the reversal of motion of the head. The elastic properties of the seat back led to an increase of those values, which were significantly above average.

Seat L showed a behaviour somewhat similar to seat H. However, due to high initial head-to-head restraint distance, NIC_{max} was higher (18.2) than with seat H. Also, due to a delta-t value of 18 ms, high moments of torque for the head extension (30 Nm) werw observed at the upper neck transducer since an additional extension was provoked by forces transmitted from the thorax (already moving forward) via the neck to the occiput (States 1970). Also, the head restraint, which were not locked vertically and moved downward during the head contact, contributed to this adverse behaviour.

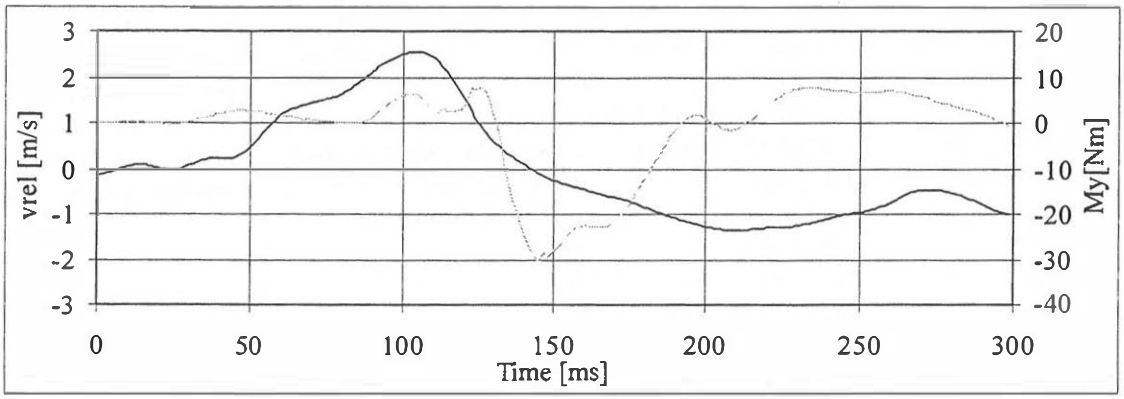

Figure 5: Relative velocity of the head vs. T1 in the x-direction of a co-ordinate system fixed to T1(black), and moment of torque around the y-axis of the upper neck transducer (grey), as measured in test L. As with most seats that exhibit a high 'Δt' value, a maximum extension moment (negative My) occurs at the point in time when the relative motion between T1 and head reverses its direction. This is due to the fact that the head is still rotating backwards when the thorax begins to move forward and, due to the forces transmitted through the neck, 'yanks' the latter into an additional extension motion.

Seat T was again an example with a stiff seat back design. In contrast to seat I, the design of the seat back generated lower T1 accelerations, and the head-to-head restraint distance was very small (30 mm). Thus, favourable biomechanical values were recorded during the first phase of the collision (NIC 11.5). Moments of torque and shear forces between first and second phase were also very low. As with all seats that employed this design principle, we found higher than average values for the elasticity parameters that might have led to an increased biomechanical loading in the third (protraction/belt restraint) phase.

CONCLUSIONS

The forward rebound phase can only be neglected in seat design where seat backs are applied that deform plastically during low speed (delta-v = 15 km/h) rear impacts with relatively low (6 g average) car accelerations. In the stiff seats that are (in part due to no-collapse design requirements for more serious car mean accelerations) becoming more and more popular, the forward rebound phase is biomechanically significant for the risk of soft tissue neck injuries, and should therefore be considered in the seat design. The reason why no-collapse requirements seem to take precedence over whiplash protection in some current seat designs may also lie in the fact that, for seat back strength, technical guidelines (ECE R17, FMVSS 207) exist, while for whiplash protection, a standardised test procedure has so far been lacking. We have described a number of possible measurements during a test according to our procedure in this paper; however, as far as the rebound phase is concerned, we have not yet reached definitive conclusions with regard to the measurement parameters and their respective tolerance limits to be applied for the assessment of the 'violence' of the rebound phase. Before the combined criteria such as NIC_{min} or the elasticity (el) parameters are applied, we find it simpler and more **straightforwad to interpret direct measurements such as Δt and v_{rel}**+ for the second phase and v_{T1}+ for the third phase of motion in rear end impacts.

The engineering conflict between 'no-collapse' requirements and the demand for plastic deformation in the seat back (e.g. Strother 1987, Warner 1991) needs a renewed discussion, even more so since there are indications (Parkin 1995) that neck injuries occur more frequently in vehicles with unyielding (and therefore, in most cases, elastic) seats. Possible approaches to resolve this conflict by constructing seat backs that yield but do not collapse have become evident in our tests. On the other hand, the design and biomechanical performance of a stiff but inelastic seat back would merit closer investigation e.g. in situations where confined space for the rear seat occupants prohibits the use of yielding seat backs.

Obviously, advanced protection systems that address the yielding characteristics of the seat directly show promising results in mitigating the biomechanical consequences of the rebound phase while maintaining sufficient seat back strength in more severe collisions; less obviously, also protection systems like those based on forward moving head restraints, i.e. not based on manipulation of the recliner characteristics, in part exhibit favourable properties with respect to rebound.

With respect to the belt loading to be expected in the third (protraction/belt restraint) phase, our results support the hypothesis that, if the dummy moves with a given velocity of e.g. 15 km/h towards the belt system in this phase, biomechanical loadings will be higher than those that would be found in a corresponding test where a frontal impact with a delta-v of 15 km/h is simulated. However, a numerical relationship between the biomechanical measurement parameters in those two scenarios has at present not beend derived. By using the relative velocity of the chest (or T1) just *before* the occupant is caught by the belts, the performance of a seat system in that respect can be assessed without having to reproduce the exact belt geometry of the target vehicle, at least if belt pretensioners are not triggered in rear end impacts.

Further work should also focus on the biofidelity of the ATD's used in this study, and on ways of quantifying 'elasticity' with respect to the biomechanical responses mentioned above. A broader range of seat models tested could further corroborate the results presented here.

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