NEW AIS1 LONG-TERM NECK INJURY CRITERIA CANDIDATES BASED ON REAL FRONTAL CRASH ANALYSIS

Ola Boström, Katarina Bohman, Yngve Håland
Autoliv Research
Anders Kullgren, Maria Krafft
Folksam Research

ABSTRACT

The AIS1 neck injury is the most frequent disabling injury in frontal impacts. Recent research has shown that, similar to rear impacts, the crash pulse level rather than the speed change influence the risk of sustaining a short or long term injury. Also similar to rear impacts, different injury mechanisms have been proposed.

In this study, new AIS1 neck injury criteria for frontal impacts were proposed and evaluated namely $NIC_{protraction}$ and the established AIS3+ criteria N_{IJ} and upper neck flexion moment $My_{flexion}$. The $NIC_{protraction}$ calculation is analogous to NIC_{max} shown to be applicable and relevant for evaluating the neck load in rear-end impacts.

Totally 172 belted occupants involved in 144 real frontal crashes with recorded crash pulses were simulated and analysed using MADYMO models of the HIII 50th percentile male. The injury outcome in terms of short-term, long-term or no neck injury, as well as the crash pulse and the utilisation of airbag and belt-pretensioner were known.

At least 70% of the $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ values associated with the non-injured occupants were lower than the values for at least 70% of the long-term injured. In the development of frontal impact protection systems $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ should therefore at least be lower than AIS1 long-term neck Injury Assessment Reference Values. The rounded median values for the long-term injured were in this study found to be 25 m^2/s^2 for $NIC_{protraction}$, 0.2 for N_{IJ}, and 40 Nm for $My_{flexion}$.

Key words: NECK, INJURY CRITERIA, FRONTAL IMPACTS, DATABASES.

PROTECTIVE MEASURES, such as the Head Injury Criterion (HIC) and Viscous Criterion (VC) have been used successfully in the design of today's cars to mitigate the occupant risk of death or serious injury. However, protecting the head and the torso does not guarantee that disabling AIS1 neck injuries are avoided (Krafft, 1998a). Although the exact injury mechanisms responsible for disabling AIS1 neck injuries in frontal collisions remain unknown, there is an urgent need for vehicle design guidelines and tools to mitigate disabling neck injuries for all crash circumstances. Research papers concerning AIS1 neck injuries in frontal collisions are rare and neck injury criteria for short and long-term AIS1 neck injuries in frontal impacts do not yet exist.

The appropriateness of using the standardised Hybrid III (HIII) anthropomorphic test device to evaluate AIS1 neck injury in both rear and frontal impacts has been questioned (Thunnisen et al., 1995; Kullgren et al., 1999). To improve the evaluation of rear impacts, the new Biofidelic Rear Impact Dummy (BioRID) was developed (Davidsson et al., 1998) and Eriksson (2000) has since developed and validated a MADYMO alternative of the BioRID. Although the BioRID has not yet been validated against volunteer frontal impact tests, the greater flexibility of its neck compared to the HIII neck should allow a preliminary evaluation of how neck flexibility influences dummy dynamics.

Since 1992, more than 140 000 Swedish cars has been equipped with crash pulse recorders (installed by Folksam insurance company). Data from frontal impacts have shown that the deceleration of the occupant directly after belt contact is related to the risk of sustaining long-term neck injury (Kullgren et al., 1999, 2000a). Moreover, computer simulations of a HIII dummy have

shown that occupant kinematics were significantly influenced by the shape of the crash pulse (Kullgren et al., 1999). In a recent study by Kullgren et al (2000), where the probability of a reported AIS1 neck injury versus crash severity measured with crash pulse recorders was studied, it was found that an airbag in combination with a seatbelt pretensioner had a significant reducing influence on the neck injury risk. Morris et al (2000) also found similar results. In both studies the combined effect of airbags and seatbelt pretensioners was studied. It should be noted that only reported neck injuries were considered in these studies, irrespective of the long-term consequences. Several studies have shown the importance of separating the analyses for short and long-term disability resulting from a reported AIS1 neck injury (Krafft, 1998b; Kullgren et al., 1999). The injury mechanisms as well as the parameters influencing injury risks may differ for short and long-term disabilities.

In Figure 1, a schematic view of four possible extreme shapes during frontal or rear-end collisions is shown. In frontal impacts, AIS1 neck injuries may take place in the initial phase when the neck may perform protraction (reversed S-shape) motion, meaning local upper neck hyperextension or/and in the late phase when the neck is forced into hyperflexion.

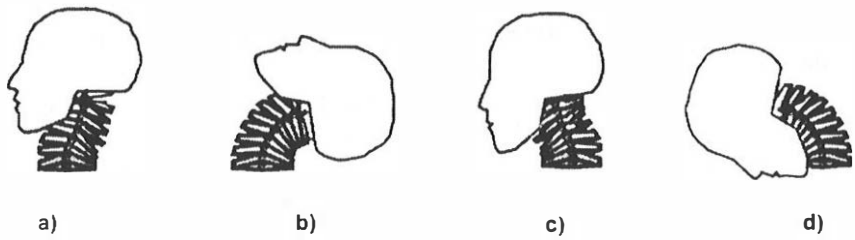

Figure 1 - Schematic view of four possible extreme shapes during a collision: a) maximum retraction (with lower-neck in hyperextension and upper-neck in hyperflexion), b) hyperextension, c) maximum protraction (with lower-neck in hyperflexion and upper-neck in hyperextension), and d) hyperflexion.

The protraction motion was manifested in the volunteer tests reported by Thunnisen et al (1995) and Deng et al (1998) by a temporal delay of the head rotation with respect to the neck rotation. This delay resulted in shear force peak of the upper neck. Other researchers are currently documenting the protraction motion in volunteer and BioRID tests (Davidsson, 2000). Although the HIII neck is not optimal for studying protraction motion, the relative acceleration between the HIII head and the torso might still be related to the violence to the neck in the human protraction motion.

The neck injury criterion N_{ij} peak value, which takes into account the combined load of upper neck moment and axial force, may occur both initially and later. If the injury is occurring in the late phase, the traditional AIS3+ upper-neck flexion-moment criterion (but with lower reference values) may be used also to evaluate the risk of AIS1 neck injuries. The head-to-torso motion in frontal and rear-end impacts exposes portions of the neck to similar types of motions (Figure 1 a) versus c) and b) versus d)). If the injury is occurring in the initial phase, NIC_{max} (Boström et al., 1996, 2000a), already used in rear end impacts, can be used in a revised version to evaluate the risk of injury in frontal collisions. NIC_{max} has been shown to be sensitive to the influence of the major risk factors: crash pulse, seat force-deflection characteristics, car model (disability risk-list) and head-to-head restraint gap. NIC_{max} predicts impact conditions that can result in soft tissue injuries with acceptable accuracy and a high NIC_{max} is always related to excessive relative motion between the head and torso (Boström et al., 1997, 2000b; Eichberger et al., 1998, 2000). Therefore, NIC_{max} seems to be a useful tool for assessing the severity of the initial relative motion between the head and torso in rear-end crashes. This could also be the case in frontal crashes.

In the present study, a set of frontal crashes with known car acceleration pulses and neck injury outcomes were assessed to evaluate possible injury criteria for AIS1 neck injuries. For this study, neck injuries were divided into three categories: no injury, injury, and long-term injuries. Both mechanical and validated mathematical (MADYMO) models of the BioRID and HIII dummies seated in a standard seat were used. The aim of this study was to identify injury criteria that correlated to the duration of neck injuries sustained by occupants in real-world frontal crashes.

METHOD/MATERIALS

In the main part of this paper, an HIII mathematical model was used to simulate real-life frontal crashes with known injury outcome. The loading of the dummies in terms of the below described parameters, were used to interpret the injury outcome of the human occupants. Also, a BioRID mathematical model was used to see whether or not the found HIII conclusions were significantly influenced by the difference in flexibility of the spine, especially of the cervical spine. The reason for mainly using the HIII dummy was that the mechanical BioRID dummy was not robust enough to withstand the complete series of the mechanical validation sled tests.

REAL-WORLD DATA - The most recent version of the Folksam database of recorded crashes with a frontal overlap of more than 25% and with a repair cost exceeding 7000 USD, was used with the inclusion of surviving belted occupants with known injury outcome. Totally 144 frontal crash pulses and 172 occupants where included in the study. 127 of the 172 occupants were drivers while 45 were passengers. The mean car Δv and acceleration were 22 km/h and 5g respectively, see also Figure 2 for the cumulative distributions.

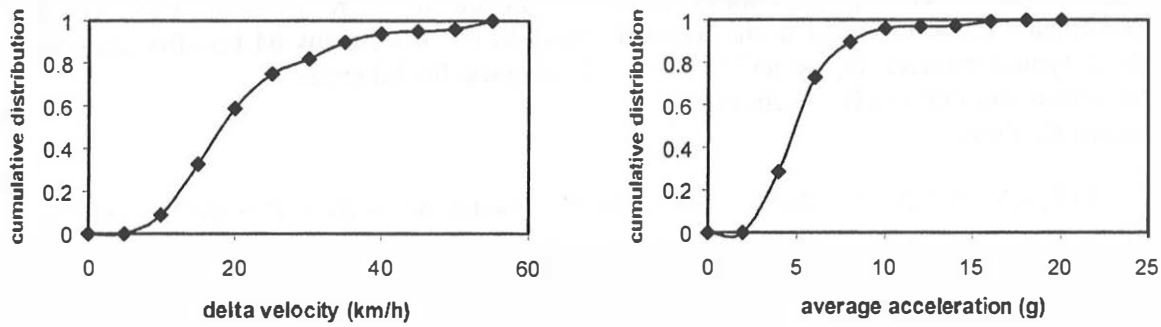

Figure 2 - The cumulative distributions of the car Δv's and mean accelerations.

Out of the 172 occupants 50 sustained an AIS1 neck injury and 11 still suffered from a neck injury after 6 months (long-term injury). Only one sustained an AIS3+ neck injury.

PHYSICAL MODELS - A series of mechanical sled tests were performed to complement the prior validation of the mathematical models. Two crash pulses with same speed change (28 km/h) were used, a typical no-neck injury and a typical long-term injury crash pulse, according to the Folksam database, see Figure 3. Both HIII and BioRID 50th percentile male dummies were used. The dummies and the seats were positioned in design positions, according to standardised frontal crash sled test procedures. The belt system consisted of different combinations of a belt with a standard retractor, belt pretensioner and for the HIII also a driver airbag. The dummy had a possibility to interact with a knee bolster, toe pan and a steering wheel. Intrusion of the compartment was not simulated as the maximum level of intrusion for the investigated crashes were low and unlikely to have a major impact on the conclusions drawn in this study.

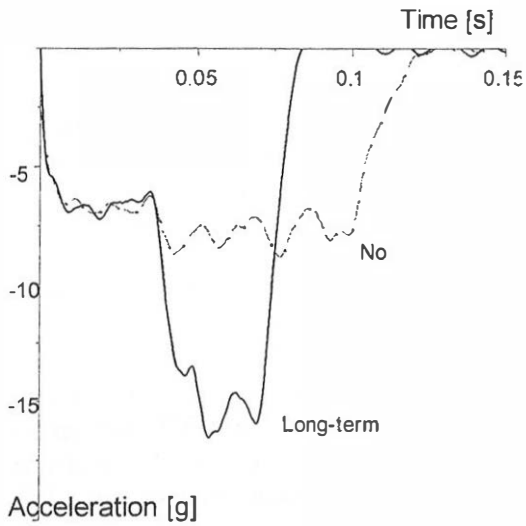

Figure 3 – The two crash pulses with the same speed change (28 km/h) used in the physical tests, a typical no-neck injury and a typical long-term injury crash pulse, according to the Folksam database.

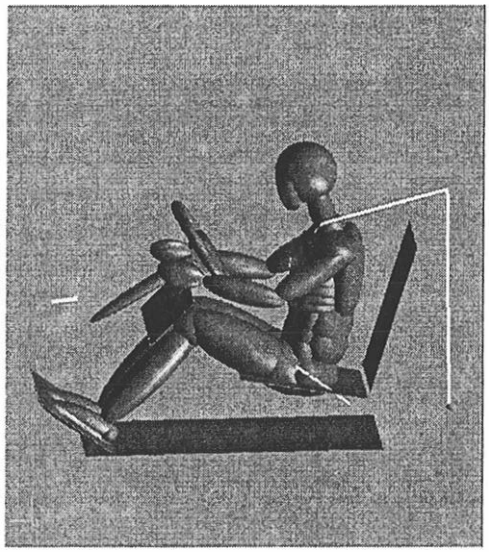

Figure 4 – A pictorial view of a MADYMO HIII dummy 64 ms after the start of a typical frontal crash.

MATHEMATICAL MODELS - The MADYMO models of the BioRID (Eriksson, 2000) and the HIII dummies seated in a standard seat, restrained by a three-point belt with belt pretensioner and for the HIII a driver airbag as option and with the possibility to interact with a knee bolster, toe pan and steering wheel, were used in the mathematical simulations, see Figure 4.

The airbag and the belt pretensioner were triggered 17 ms after the start of the crash (at 10mm of free flying mass displacement relative to the car for the typical long-term injury pulse according to Figure 3). The MADYMO BioRID, not previously validated for frontal impacts, was slightly changed (tuned) to the mechanical tests. To ensure robustness of the HIII model, a sensitivity analysis was performed. The evaluation of all crashes were repeated with a change of either the elastic stiffness of the seat cushion, the position of the belt pillar loop, the horizontal distance to the knee panel and for the driver only, the occupant horizontal distance to the steering wheel. The seat stiffness was increased (s+) respective decreased (s-) 20% of its original force characteristics. The knee panel was moved 100 mm forward (k+) respective backwards (k-). The belt pillar loop was moved 100 mm forward (b+) respective 100 mm backwards (b-), while the height was kept the same during all tests. These changes were calculated for both the driver and the passenger. Exclusively for the driver, the influence of the distance to the steering wheel was estimated by moving the steering wheel 100 mm backwards (st-) compared to the standard model.

INJURY CRITERIA CALCULATION - In 1996, Boström et al (1996) proposed a new neck injury criterion based on the relative motion between the head and the lower neck, to be used in rear impacts and potentially also in frontal impacts. In rear impacts, the current standard for the retraction (rearward S-shape motion) the NIC calculation is,

$$NIC_{max} = Maximum_{first\ 150ms}\ (a_{rel} * 0.2 + (v_{rel})^2) \qquad (1)$$

where a_{rel} and v_{rel} = the relative T1-to-head centre of gravity x-acceleration/velocity according to SAEJ211 conventions. Due to the non-influence of the sign of the relative velocity in this formula, a generic formula for extreme NIC values is needed. Such a generic formula is naturally expressed as,

$$NIC_{generic} = a_{rel} * 0.2 + v_{rel} * |v_{rel}| \quad (2)$$

The expression $|v_{rel}|$ means the absolute value of the relative velocity. The following formula for frontal impacts was used as a criterion candidate and used in the validation (tuning of the mathematical models),

$$NIC_{protraction} = |\text{Minimum } NIC_{generic}| \quad (3)$$

In addition to $NIC_{protraction}$ the traditional AIS3+ injury criteria, the N_{IJ}, the upper neck flexion moment $My_{flexion}$, and the upper neck shear force Fx were evaluated. In the upgrade of FMVSS 208, the neck injury criterion N_{IJ} will replace the traditional individual criteria for compression, tension, shear, flexion and extension of the upper neck. The N_{IJ} criterion combines the upper neck axial force and the upper neck moment around the y-axis,

$$N_{IJ} = Fz/Fz_{intercept} + My/My_{intercept} \quad (4)$$

The N_{IJ} can be divided into,

- N_{TE} – tension - extension
- N_{TF} – tension - flexion
- N_{CE} – compression - extension
- N_{CF} – compression – flexion

The used intercept values, 4500N in tension and compression force and 310Nm and 125Nm in flexion and extension moment respectively, in combination with a peak N_{IJ} value to 1 corresponds to a 22% risk of sustaining an AIS3+ neck injury (Eppinger et al., 1999). There also exist injury risk curves for AIS2+ injuries, but none for AIS1+ injuries.

The validation of the BioRID moment and axial forces values were not taken into priority as these are, in contrast to the HIII ones, not validated to volunteer tests. Still, the BioRID moment values can be used to study the influence of the spine flexibility for the conclusions made.

RESULTS

Selections of the validation curves are shown in the Appendix. The MADYMO HIII $My_{flexion}$ values were found to be linearly related to the Fx shear force values with a multiplication factor of 70 mm. Therefore, the shear force values are not further presented.

The simulations showed that all evaluated criteria were correlated to AIS1 short/ long term as well as AIS1 long-term injury outcome. The third quartile $NIC_{protraction}$ and $My_{flexion}$ values for the non-injured were lower than the first quartile values for the long-term injured. For the N_{IJ}, the situation was almost the same, however with a slightly larger overlap between the long-term and non-injured.

In Table 1-3 the main results of the first sub study are collected in terms of mean, minimum and maximum values (Q0 and Q4), first, second and third quartiles (Q1-Q3), and the standard deviation for the no injured/injured and long-term injured. Note that Q2 equals the median value.

In order to present the facts of Table 1-3 more pictorially, cumulative diagrams were drawn. In Figures 5 to 7, the cumulative diagrams of the $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ values are shown.

Table 1 – The mean, minimum and maximum values (Q0 and Q4), first, second and third quartiles (Q1-Q3), and the standard deviation for the simulated $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ values of the non-injured (N=122).

Non injured	Mean	Q0	Q1	Q2	Q3	Q4	Stdev
$NIC_{protraction}$ [m²/s²]	15	4	8	11	18	77	11
N_{IJ}	0.15	0.05	0.08	0.10	0.17	0.92	0.12
$My_{flexion}$ [Nm]	27	11	20	23	28	86	14

Table 2 – The mean, minimum and maximum values (Q0 and Q4), first, second and third quartiles (Q1-Q3), and the standard deviation for the simulated $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ values of the short and long term injured (N=50).

Injured	Mean	Q0	Q1	Q2	Q3	Q4	Stdev
$NIC_{protraction}$ [m²/s²]	19	5	10	15	23	66	12
N_{IJ}	0.19	0.06	0.10	0.16	0.23	0.94	0.15
$My_{flexion}$ [Nm]	34	13	22	25	40	111	20

Table 3 – The mean, minimum and maximum values (Q0 and Q4), first, second and third quartiles (Q1-Q3), and the standard deviation for the simulated $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ values of the long-term injured (N=11).

Long-term	Mean	Q0	Q1	Q2	Q3	Q4	Stdev
$NIC_{protraction}$ [m²/s²]	30	9	18	25	37	66	19
N_{IJ}	0.31	0.09	0.15	0.21	0.39	0.94	0.25
$My_{flexion}$ [Nm]	49	19	28	39	65	111	19

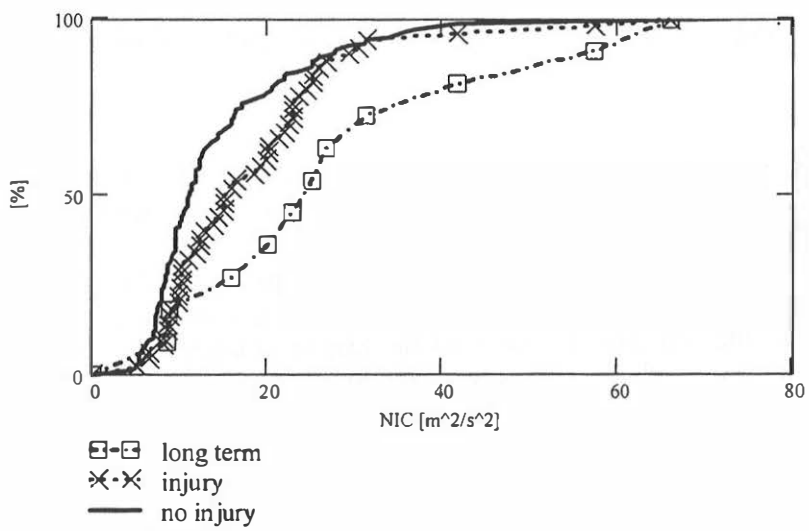

Figure 5 - The cumulative diagram of simulated HIII-$NIC_{protraction}$ values for non-injured, injured and long-term injured occupants. Note that the long-term injured are included in the injured group.

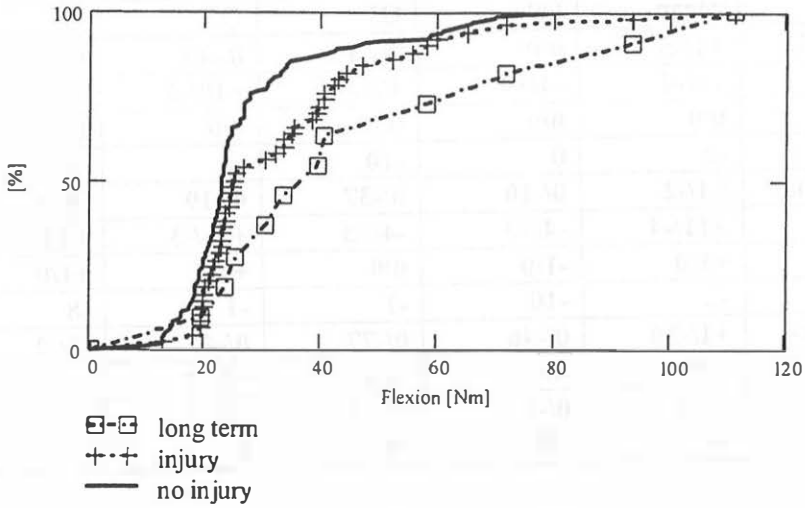

Figure 6 - The cumulative diagram of simulated HIII-$My_{flexion}$ values for non-injured, injured and long-term injured occupants. Note that the long-term injured are included in the injured group.

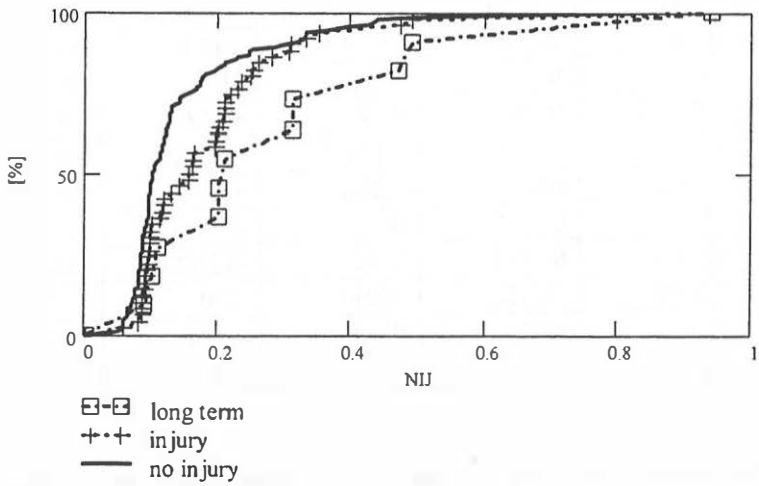

Figure 7 - The cumulative diagram of simulated HIII-N_{IJ} values for non-injured, injured and long-term injured occupants. Note that the long-term injured are included in the injured group.

In the sensitivity analysis the Q3 non-injured and the Q1 long-term injured changes were less than 17% and 14% respectively. The $NIC_{protraction}$ and N_{IJ} mean variations were less than 13% while the highest $My_{flexion}$ mean change was 20%. In Table 4 and 5, the parameter-change influences expressed in percent are shown for the set of non- and long-term injured. For example, +10/-5 in Table 4 cell, $NIC_{protraction}$ b+/b-;Q2, means that the $NIC_{protraction}$-median value for the 122 non-injured occupants was increased with 10% when the pillar loop was moved towards the occupant (horizontally 100mm forward) and was decreased with 5% when the pillar loop was moved away from the occupant.

Table 4 – The influence of a knee bolster move (k+/k-), pillar loop move (b+/b-), seat stiffness change (s+/s-) and steering wheel move (st-) on Table 1.

Non injured	Mean	Q0	Q1	Q2	Q3	Q4
$NIC_{protraction}$ k+/k-	+1/-5	0/0	-1/-15	0/-15	0/-8	0/-10
b+/b-	+8/-4	-4/+9	+7/-6	+10/-5	+9/-7	0/0
s+/s-	0/0	0/0	0/0	0/0	0/0	0/0
st-	-2	0	-10	-4	-4	+10
NIJ k+/k-	+1/-2	0/-10	0/-37	0/-19	-1/-17	-4/+16
b+/b-	+11/-4	-4/+3	-4/+3	+18/-3	+13/-11	+11/-9
s+/s-	+1/0	-1/0	0/0	+1/0	+1/0	+1/0
st-	-2	-10	-1	-1	-8	-10
$My_{flexion}$ k+/k-	+1/-20	0/-46	0/-77	0/-47	-2/-2	+19/+15
b+/b-	+2/+2	-6/+2	-5/+2	+2/-2	+4/+3	+16/+20
s+/s-	+3/0	0/-1	+1/-4	+2/-3	+1/+1	+14/+12
st-	-9	-30	-8	0	-3	-16

Table 5 – The influence of a knee bolster move (k+/k-), pillar loop move (b+/b-), seat stiffness change (s+/s-) and steering wheel move (st-) on Table 3.

Long-term injured	Mean	Q0	Q1	Q2	Q3	Q4
$NIC_{protraction}$ k+/k-	+4/-3	0/-11	+1/-6	0/-4	+2/-2	+2/+1
b+/b-	+5/-1	+3/-21	+12/-3	+5/-5	+4/0	+1/0
s+/s-	0/+1	-3/+6	-1/+1	+1/-1	+1/+1	+1/-2
st-	-3	0	0	-10	-7	0
N_{IJ} k+/k-	+8/+13	0/-54	0/-14	+1/-6	+1/+23	-3/+11
b+/b-	+12/-4	-5/+2	+12/-8	+23/-2	+17/+1	+9/-6
s+/s-	+1/+3	0/0	+2/-1	+4/-4	0/+5	+1/+6
st-	-2	0	0	0	0	0
$My_{flexion}$ k+/k-	-5/-14	0/-101	0/+2	-11/-23	-1/-11	-5/-5
b+/b-	+2/+2	-2/+2	+5/+1	-2/0	+1/+2	-2/+4
s+/s-	0/0	0/-2	0/+3	-1/+1	+2/-1	-4/0
st-	-8	0	-5	-29	-16	0

When the BioRID dummy was used, the influence of airbag and steering wheel contact was not simulated and therefore not taken into account. Moreover, the MADYMO BioRID $NIC_{protraction}$ values had to be calculated according to Equation 3 with an additional restriction,

$$\text{BioRID-}NIC_{protraction} = |\text{Minimum}_{\text{the first 40 mm}}\ NIC_{generic}| \qquad (4)$$

That is, just as the first 150 ms were considered for NIC_{max} in rear-end collisions, only the first 40 mm of relative T1-to-head displacement was considered for the BioRID. As a result, the BioRID compared for the HIII was more sensitive to the influence of a belt pretensioner. Nevertheless, except a scaling difference and the influence of the pretensioner, all HIII and BioRID $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ values were more or less the same (r^2>0.8), see Figure 8 to10.

Regarding the relevance of real frontal-crashes, the $NIC_{protraction}$, N_{IJ} and $My_{flexion}$ results did not indicate any qualitatively difference between the HIII and the BioRID.

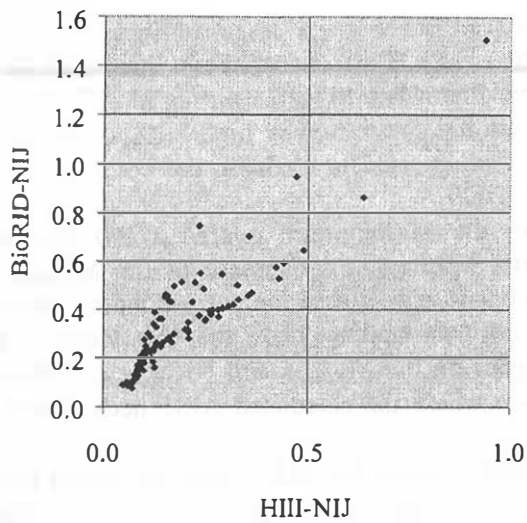

Figure 8 - BioRID-$NIC_{protraction}$ values versus HIII-$NIC_{protraction}$ values for the 172 occupants (r^2=0.86). The double linear relationship is due to the difference between the dummies regarding the influence of the belt pretensioner. The influence of airbag and steering wheel contact was not modeled and therefore not taken into account.

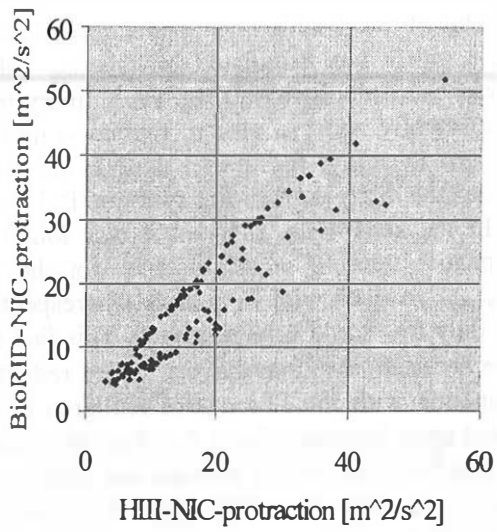

Figure 9 - BioRID-N_{IJ} values versus HIII-N_{IJ} values for the 172 occupants (r^2=0.87). The influence of airbag and steering wheel contact was not taken into account.

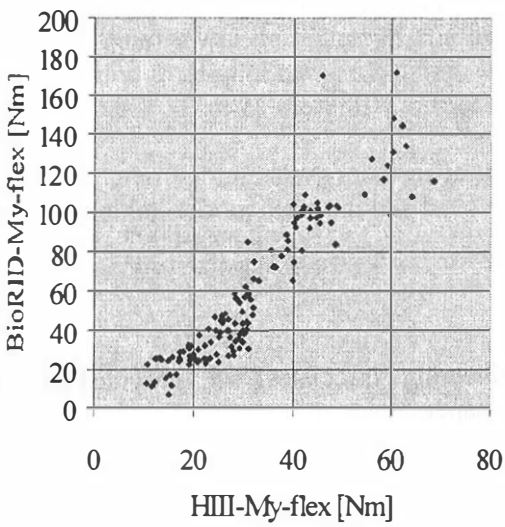

Figure 10 - BioRID-$My_{flexion}$ values versus HIII-$My_{flexion}$ values for the 172 occupants (r^2=0.84). The influence of airbag and steering wheel contact was not taken into account.

DISCUSSION

For all 172 occupants the HIII N_{IJ} and $My_{flexion}$ values were lower than the AIS3+ reference values proposed by Eppinger et al (1999) and Mertz et al (1997). As only one of the occupants in the analyzed crashes sustained an AIS3+ neck injury (N_{IJ} = 0.31), the results of this study did not contradict the established AIS3+ levels.

The mechanical BioRID was not validated to volunteer frontal impact simulations. For both the HIII and the BioRID only one dummy size, posture, seat position and belt configuration were used. Moreover, no individual biological or psychological parameters were taken into account. These and other shortcomings hindered an individual in-depth crash analysis. Nevertheless, the simulated HIII and BioRID $NIC_{Protraction}$, N_{IJ} and $My_{flexion}$ values, similar apart from the influence of a belt pretensioner and a scale difference in N_{IJ} and $My_{flexion}$, showed a remarkable relation to the overall AIS1 neck injury outcome, especially regarding the long-term consequences. That is, the present model can be used to reconstruct and evaluate a set of frontal crashes but not a single crash.

Out of the disabled occupants, three had experienced rather low average car acceleration (<6g) with a rather low simulated $NIC_{protraction}$ value (<20 m^2/s^2). Two out of these three had in turn experienced a deployed airbag. Preliminary tests show that a late triggered airbag in a frontal crash could actually cause a violent retraction motion of the neck and thereby high NIC_{max} values. In a forthcoming study, the model used in this study is further used to evaluate the complex balance between the different restraint systems (Bohman et al., 2000).

In the sensitivity analysis it was found that for the 11 long-term injured a 100 mm closer occupant-to-steering wheel distance for the 7 drivers, resulted in a decrease of the $My_{flexion}$ and $NIC_{protraction}$ values with 29% and 10% respectively. For the non-injured the corresponding decreases were only 0% and 4% respectively. This fact and preliminary findings show that the steering wheel, above a certain Δv, may have a major reduction impact on the $My_{flexion}$ and $NIC_{protraction}$ values, in conjunction with the findings of Kullgren et al (2000) where the calculated AIS1 neck injury risk stopped up to increase after a Δv of about 30 km/h.

The findings did not indicate any preference regarding when the neck injuries in frontal crashes occur. None of the three HIII predictors was demonstrably better than any of the others at predicting injury. All evaluated parameters were more or less linearly related to each other. For high N_{IJ} and $NIC_{protraction}$ values, N_{IJ} (N_{TE}) occurs at the same time as $NIC_{protraction}$ while for low values N_{IJ} (N_{TF}) occurs at the same time as $My_{flexion}$.

Further analysis, based on more real-life crashes, is required to present risk curves and a more precise injury prediction evaluation.

CONCLUSION

Only one of the 172 occupants involved in the assessed 144 real frontal crashes sustained an AIS3+ neck injury and all simulated dummy loading were below the established AIS3+ neck injury levels. A fraction of 6% sustained an AIS1 long-term neck injury. The parameters $NIC_{protraction}$, N_{IJ} and $My_{flexion}$, occurring at different phases of the crash sequence, predicted long-term AIS1 neck injuries with reasonable accuracy. The third quartile values for the non-injured were lower (almost for N_{IJ}) than the first quartile values for the long-term injured. The rounded median long-term criteria values (50% risk Injury Assessment Reference Value candidates) for the HIII were 25 m^2/s^2, 0.2 and 40 Nm respectively.

To mitigate disabling AIS1 neck injuries in the design of restraint systems for frontal impacts it is therefore recommended achieving a balance between the head and the torso motion initially as well as during the flexion phase.

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APPENDIX - VALIDATIONS OF THE MATHEMATICAL MODELS

HIII 50th percentile dummy - A set of mechanical sled tests was performed. The test set-up included a HIII dummy, a reinforced seat, floor, toe-pan, knee panel and restraint system. There was no intrusion of the toe-pan, floor or knee panel. In all tests the dummy was restrained with a standard buckle and a three-point seat belt, with a webbing elongation of 12%. In some of the tests an airbag and a pretensioner was added to the restraint system. Two different crash pulses were used. The Δv was 28 km/h for both pulses. However, the shape of the pulse was different. (see Figure 2 in the paper).

A MADYMO (version 5.4.1) sled test model of the mechanical sled tests was developed. The HIII 50th percentile dummy version 5.2.1 was used. The belt system was modeled with the conventional belt system available in MADYMO. The model was validated for the two different crash pulses. The model was also validated for different restraint systems, such as pretensioner and airbag.

The measurements that were used to validate the model were head acceleration, lower neck acceleration, chest acceleration, pelvis acceleration, shoulder belt force, upper neck moment (My), upper neck axial force and upper neck shear force.

There was generally good agreement between the predictions of the MADYMO model and the mechanical sled tests (see figure A1 and A2). The predicted accelerations were in general slightly lower than the measured accelerations in the sled tests, but the shape of the accelerations was similar to the measured. The maximum neck moment was slightly delayed in the MADYMO model compared to the sled tests. The prediction from the model indicated the correct trend compared to the results from the sled tests.

BioRID - The mechanical tests with the BioRID were performed in the same test set-up as with the HIII 50th percentile dummy. However, no tests with airbag were performed.
The MADYMO sled test model of the mechanical sled tests was developed. The MADYMO model of the BioRID validated for rear impacts were used. This dummy model was tuned in to better perform the protraction motion according to the mechanical sled tests. The same measurements as in the HIII validation were used to validate the BioRID model.

There was generally good agreement between the predictions of the MADMYMO model and the mechanical sled tests during the first 100 ms (see figure A3 and A4). After this time, the head acceleration was slightly delayed and thereafter, the head acceleration reached too high absolute values, which influenced the upper neck moment as well as the $NIC_{protraction}$. The model predicted the correct trend of the upper neck moment, but the absolute values were too high. About the $NIC_{protraction}$, the maximum neck protraction motion had already occurred within the first 100 ms, therefore, it was possible to study this phenomenon with the BioRID. The upper neck axial force also showed the correct trend, but with too high absolute values.

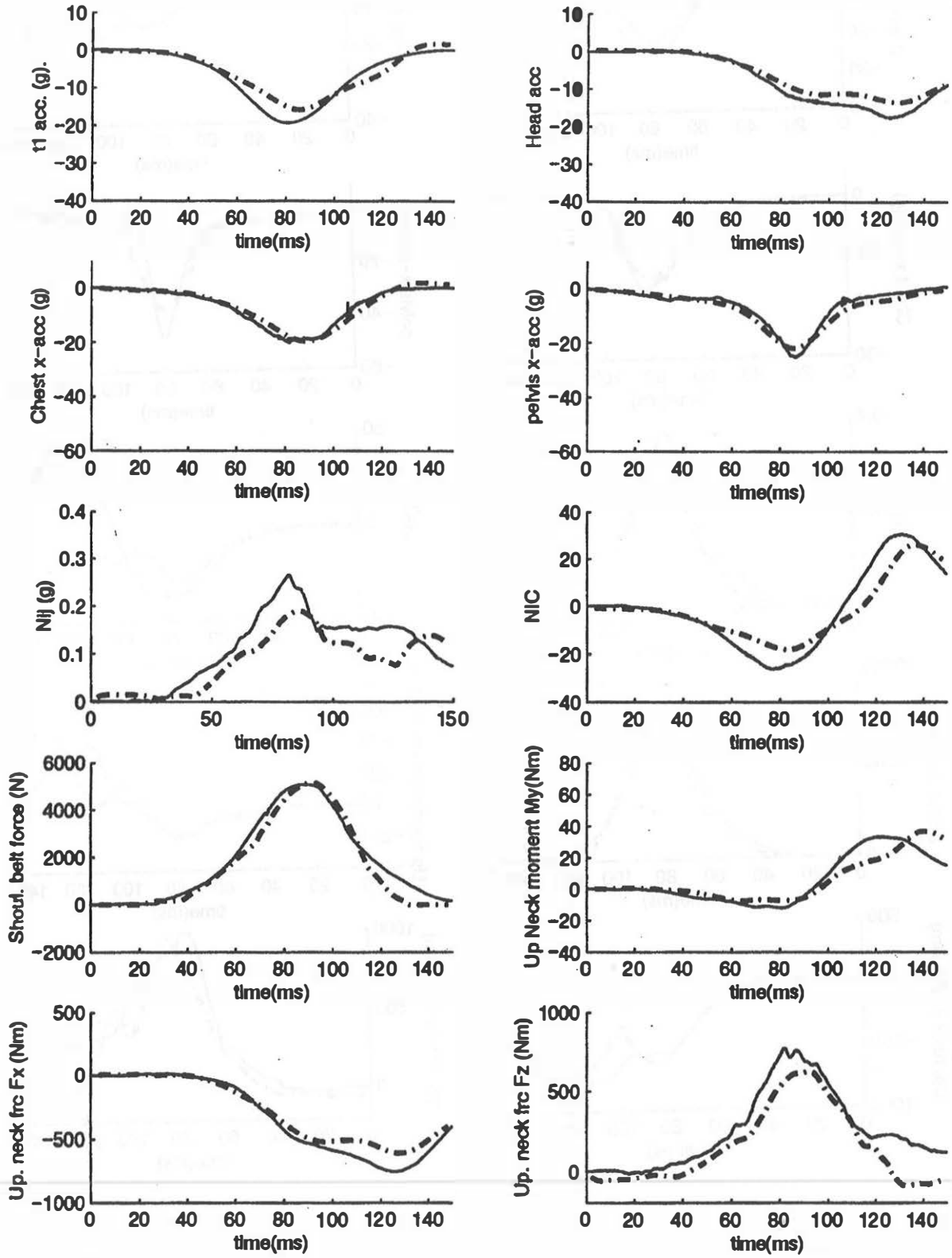

Figure A1 – The HIII mechanical (solid) and mathematical (dotted) T1, head, chest, pelvis x-accelerations and the HIII N_{IJ}, $NIC_{protraction}$, shoulderbelt force, upper neck moment, upper neck Fx and Fz forces for the "no neck-injury pulse".

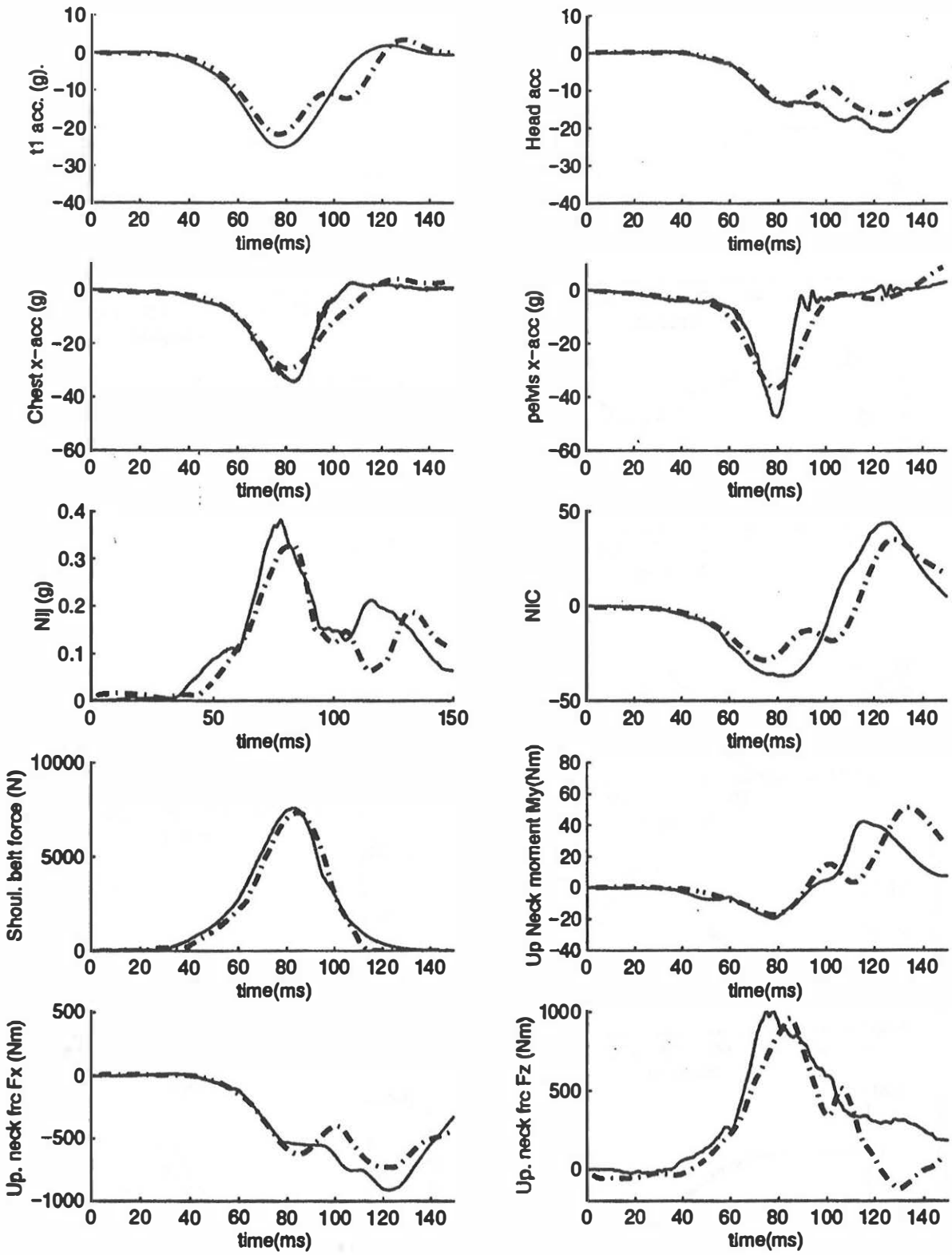

Figure A2 – The HIII mechanical (solid) and mathematical (dotted) T1, head, chest, pelvis x-accelerations and the HIII N_{IJ}, $NIC_{protraction}$, shoulderbelt force, upper neck moment, upper neck Fx and Fz forces for the "long-term neck-injury pulse".

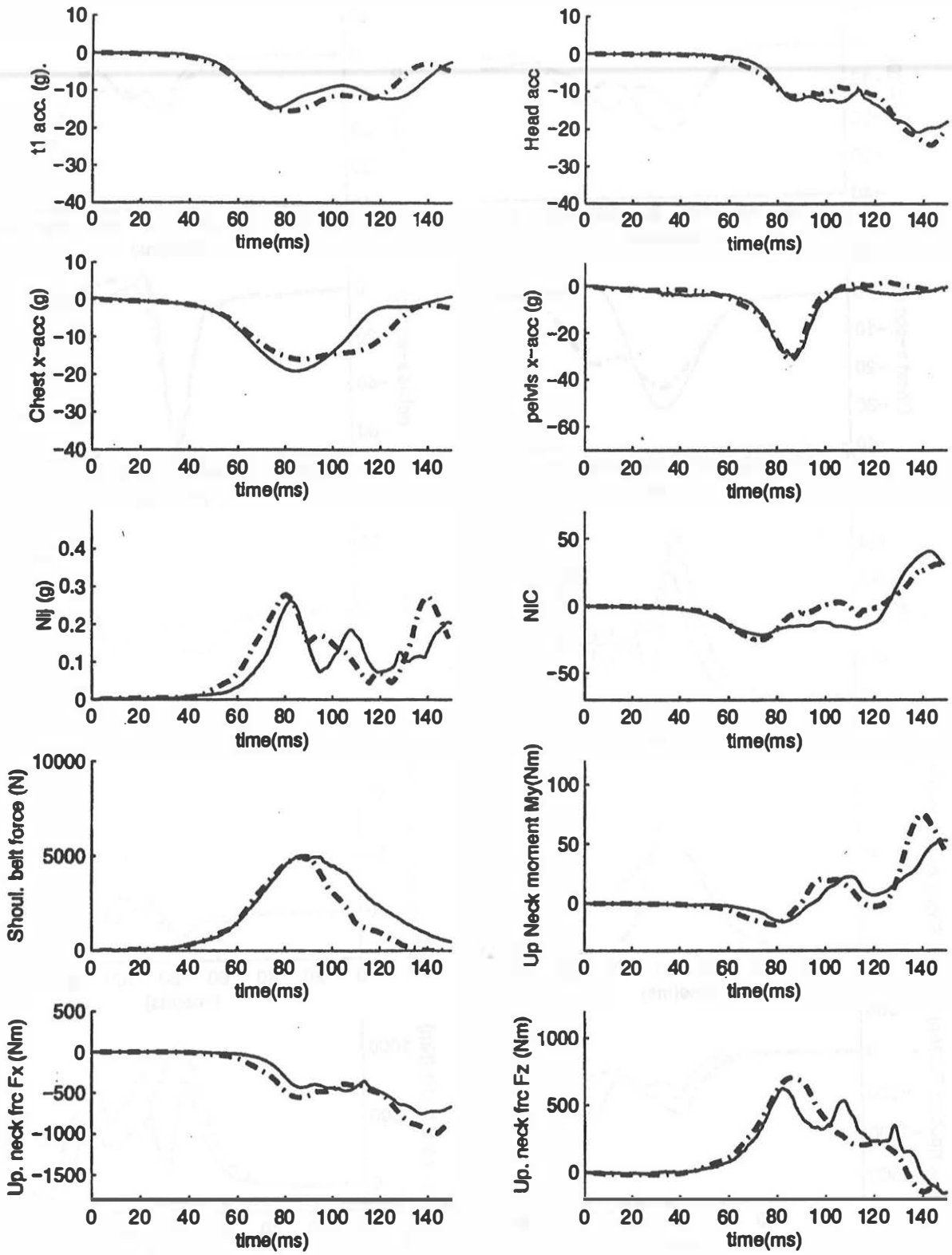

Figure A3 – The BioRID mechanical (solid) and mathematical (dotted) T1, head, chest, pelvis x-accelerations and the BioRID N_{IJ}, $NIC_{protraction}$, shoulderbelt force, upper neck moment, upper neck Fx and Fz forces for the "no neck-injury pulse".

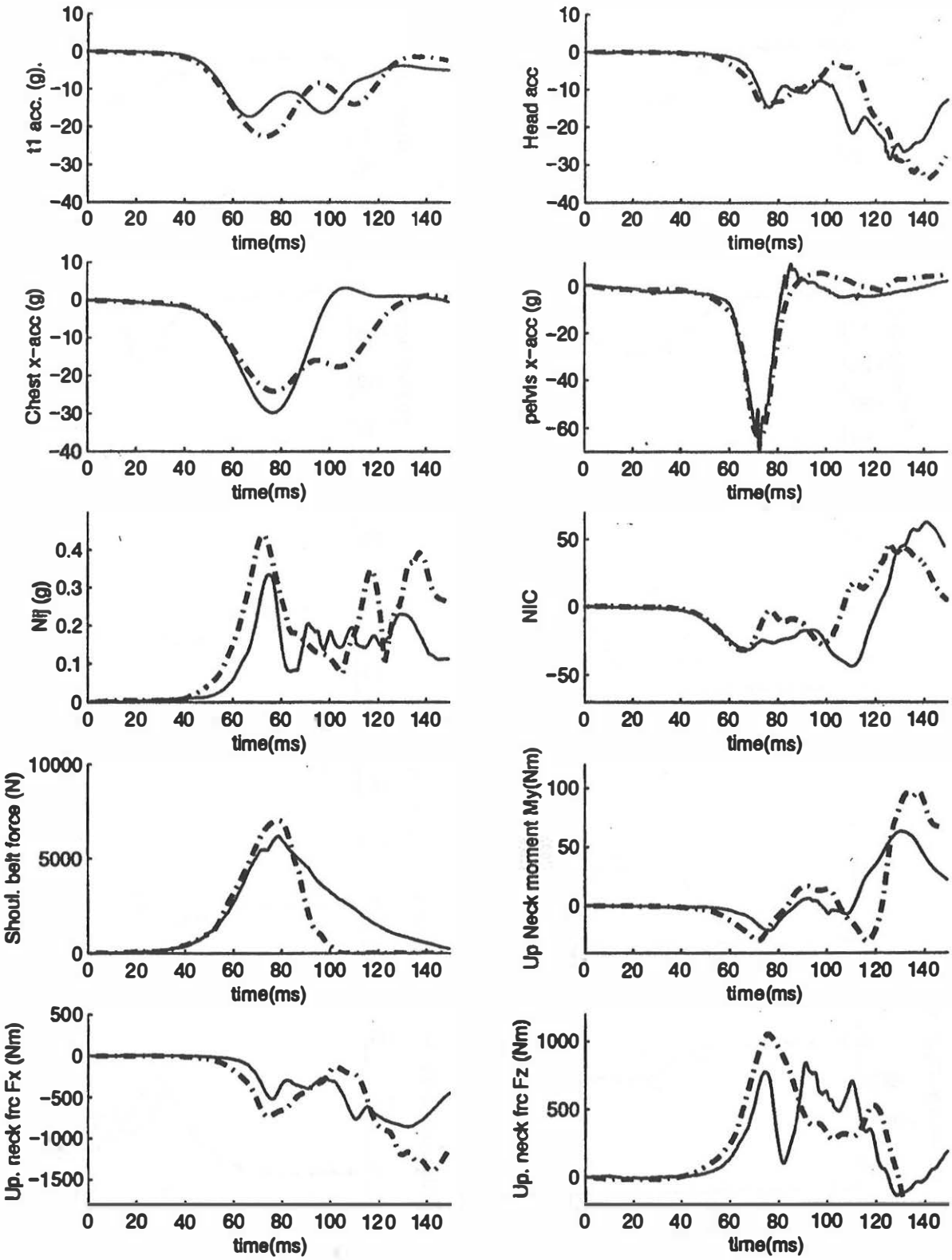

Figure A4 – The BioRID mechanical (solid) and mathematical (dotted) T1, head, chest, pelvis x-accelerations and the BioRID N_{IJ}, $NIC_{protraction}$, shoulderbelt force, upper neck moment, upper neck Fx and Fz forces for the "long-term neck-injury pulse".