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The Effect of Collision Pulse Properties on Six Proposed Whiplash Injury Criteria

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ABSTRACT

Recent epidemiological and biomechanical studies have suggested that whiplash injury is related to a vehicle's average acceleration rather than its speed change during a collision. In this study, we explored how six proposed whiplash injury criteria varied with low-speed rear-end collision kinematics. A BioRID II rear-impact dummy was seated on a programmable sled and exposed six times to each of fifteen different collision pulses. Five properties of the collision pulse were varied: peak acceleration (1.3 to 4.4 g), speed change (3 to 11 km/h), duration (52 to 180 ms), displacement (2 to 26 cm) and shape (square, sine and triangular). Linear and angular accelerations and displacements of the head and linear accelerations of the T1 and pelvis were measured in the sagittal plane. Upper neck loads in the sagittal plane were also measured. Variations within the proposed injury criteria between the different pulses were compared using analyses of variance. Five criteria – peak upper neck shear force, peak upper neck moment, peak retraction, the neck injury criterion (NIC) and the normalized neck injury criterion (N_{ij}) – all exhibited graded responses that were most sensitive to the average acceleration of the collision pulse. Peak extension angle between the head and T1 exhibited a negatively graded response and was therefore unsuitable as a whiplash injury criterion for the BioRID dummy. Of the six criteria, N_{ij} was best able to distinguish between the fifteen pulses. If the five positively-graded criteria are related to the risk of whiplash injury, then the results of this study indicated that the risk of whiplash injury can be reduced by bumper and seat designs that prolong the collision pulse and thereby reduce the average vehicle acceleration for a given speed change.

INTRODUCTION

Unlike most automobile-related injuries, the risk of whiplash injury has increased over the last few decades (v Koch et al., 1994; Morris and Thomas, 1996; Temming and Zobel, 1999). In one recent report, the risk of sustaining a whiplash injury with symptoms lasting at least one year was 2.7 times higher in vehicles introduced between 1989 and 1992 than in vehicles introduced between 1981 and 1986 (Krafft, 2002). The reasons for this increase in whiplash injury risk remain unclear, but at least two mechanical reasons have been proposed: first, newer seats are stiffer and may induce greater head and neck dynamics (Parkins et al., 1995; Krafft, 2002); and second, newer bumpers are both stiffer and more resilient than older bumpers and may cause both higher accelerations and higher speed changes (due to increased restitution) for a given closing speed (Siegmund and King, 1997; Krafft, 2002). In the current study, we have focused on the latter proposal and examined how changes in the kinematic properties of the collision pulse affect several occupant kinematic and kinetic variables that are potentially related to the risk of whiplash injury.

Collision pulses can be characterized by discrete parameters such as speed change, average acceleration, peak acceleration, collision duration, displacement during the collision and pulse shape. Although these descriptors of collision severity are inter-related, recent epidemiological studies have shown that increases in the severity and duration of whiplash injury correlated best with increased average acceleration, but also with increased speed change (Krafft et al., 2000; 2002). Previous experimental and modeling studies have shown that changes in some combinations of vehicle speed change, acceleration, collision duration and pulse shape alter the occupant response (Nilson et al., 1994; Siegmund et al., 1997; Eriksson and Boström, 1999; Boström et al., 2000; Brell et al., 2001; Welcher et al., 2001), but it remains unclear how occupant responses and the associated risk of whiplash injury change when multiple combinations of these collision parameters are varied systematically.

Six whiplash and neck injury criteria were considered by NHTSA (2000) for the new head restraint Standard 202: upper neck shear force, upper neck moment, head/neck retraction, head/neck extension, the neck injury criterion (NIC), and the normalized neck injury criterion (N_{ij}). Although NIC has been shown to correlate with the duration of whiplash symptoms (Eriksson and Boström, 1999), it remains unclear which of these criteria correlate best with whiplash injury.

The goal of this experiment was to evaluate how six proposed whiplash injury criteria calculated from occupant kinematic and kinetic data vary with different kinematic properties of the collision pulse. The results of this study will help identify which of the proposed injury criteria best explain recent epidemiology data showing that whiplash injury severity and duration correlate with average vehicle acceleration and, to a lesser degree, speed change (Krafft et al., 2000; 2002).

METHODS

A BioRID II rear-impact dummy was instrumented to measure head, T1 and pelvis kinematics and upper neck kinetics in the sagittal plane. Linear accelerations of the head and T1 were measured using orthogonally-oriented uni-axial accelerometers (Model EGAS-FS-100 and EGAS-F-25 for x and z respectively; Entran Devices, Fairfield, NJ) and linear accelerations of the pelvis were measured using a tri-axial accelerometer (Summit 34103A; ±7.5g, Akron, OH). Angular head kinematics were measured using a uni-axial angular rate sensor (ARS-04E, ATA-Sensors, Albuquerque, NM). A three-axis load cell (Model 2564, R.A. Denton, Rochester Hills, MI) was used to measure upper neck loads.

Horizontal sled acceleration was measured using a uni-axial linear accelerometer (Sensotec JTF3629-05, Columbus, OH). Head restraint contact was detected with a force sensitive resistor (FSR#406, Interlink Electronics, Camarillo, CA) applied to the front of the head restraint. Transducer and contact switch data were acquired for 2 seconds at 10 kHz using a 12-bit, simultaneous-sample-and-hold Win30 DAQ card (United Electronics Incorporated, Watertown, MA). All data channels conformed to SAE J211, Channel Class 1000 (SAE, 1989) except for the sled acceleration channel, which was Channel Class 180.

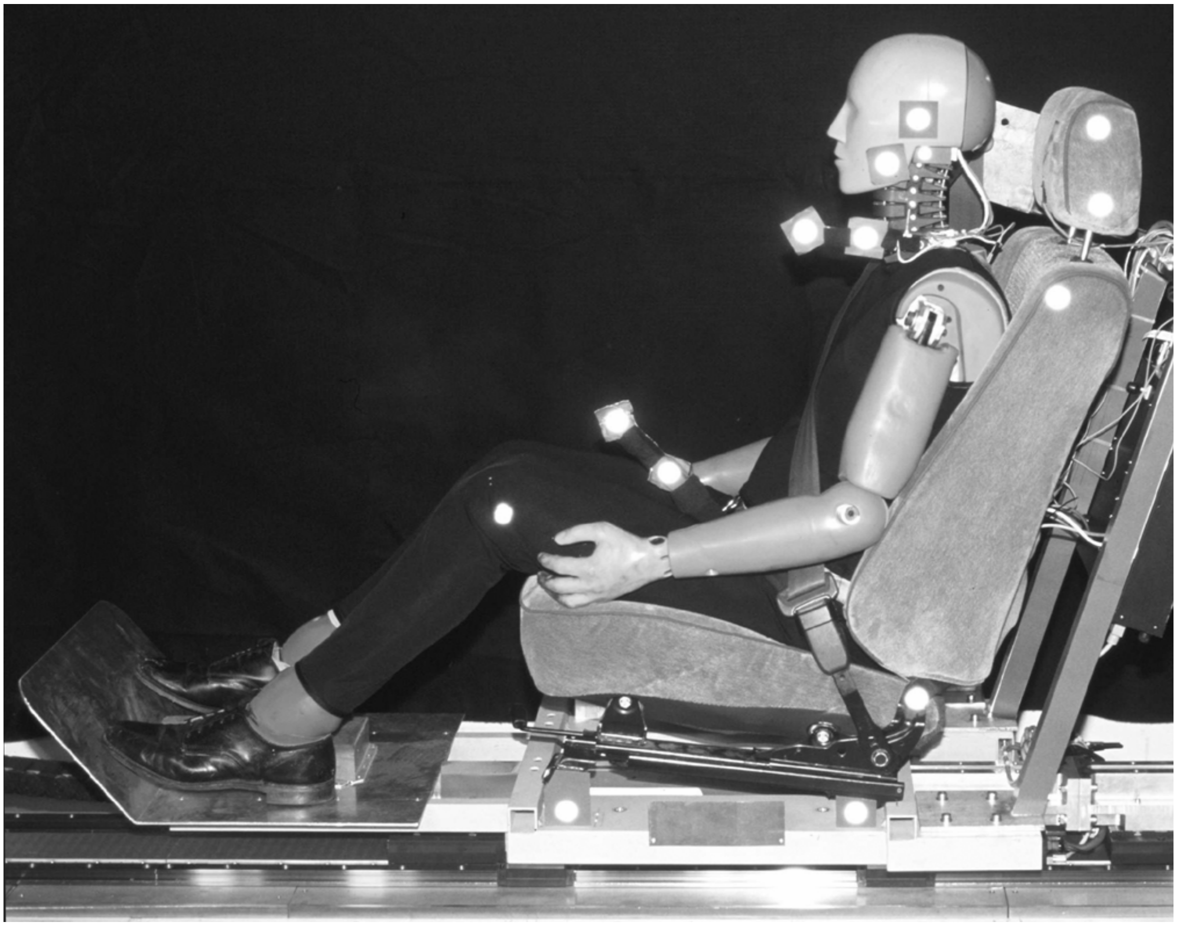


Figure 1: Experimental setup showing the pre-impact position and posture of the BioRID II dummy seated on the sled.

Digital video of the sagittal plane motion was captured using an OmniSpeed HS motion capture system (Speed Vision Technologies, Solana Beach, CA) and high-speed camera (JCLabs 250; 512 x 216 lines resolution, Mountain View, CA). Video data were recorded at 250 frames per second using a shutter speed of 1/1000 s. Reflective targets were applied to the sled, seat and dummy (Figure 1) and subsequently digitized. The digitized video data had an accuracy of ± 2 mm at the vertical plane containing the seat centerline. Synchronization was achieved by simultaneously triggering the data acquisition and video systems. The initial position of the dummy was measured using a three-dimensional digitizer (FaroArm B08-02, Lake Mary, FL; single-point accuracy of ±0.30 mm) to ensure a repeatable initial position and posture.

Test Procedures

The BioRID was dressed in two layers of lycra (Davidsson, 1999) and placed in the front passenger seat from a 1991 Honda Accord LX 4-door sedan. The seat back angle was set to 27 degrees from the vertical and the head restraint was locked in the full-up position. The distance between the back of the BioRID's head and front surface of the head restraint, i.e., the backset, was 8 cm. A 3-point seat belt was fastened snugly for all tests.

The seat was mounted to a feedback-controlled linear sled that accelerated the seat and dummy forward (+x) from rest. The sled generated 15 different collision pulses that could be

Table 1. Description of the fifteen pulses and five groups used in this experiment. See Figure 2 for a superposition of the pulses within each of the 5 groups. Tri-Iso, isosceles triangle; Tri-Asc, triangle with ascending ramp; Tri-Des, triangle with descending ramp.

Pulse	Shape	? V	? T	a_{peak}	s_{accel}	Groups				
		km/h	ms	g	cm	I	II	III	IV	V
A	Square	8.00	136	1.7	15.1	✓		✓	✓	
B	Sine	8.00	136	2.6	15.1	✓				
C	Tri-Iso	8.00	136	3.3	15.1	✓	✓			✓
D	Tri-Des	8.00	136	3.3	20.1		✓			
E	Tri-Asc	8.00	136	3.3	10.1		✓			
F	Square	8.00	52	4.4	5.8			✓		
G	Square	8.00	76	3.0	8.4			✓		
H	Square	8.00	100	2.3	11.1			✓		
I	Square	8.00	180	1.3	20.0			✓		
J	Square	3.06	52	1.7	2.2				✓	
K	Square	4.47	76	1.7	4.7				✓	
L	Square	5.88	100	1.7	8.2				✓	
M	Square	10.60	180	1.7	26.5				✓	
N	Tri-Des	6.93	118	3.3	15.1					✓
O	Tri-Asc	9.80	166	3.3	15.1					✓

grouped in five ways (Table 1, Figure 2): Group I pulses had the same speed change, duration and displacement, but varied in shape and peak acceleration; Group II pulses had the same speed change, duration and peak acceleration, but varied in shape and displacement; Group III pulses had the same shape and speed change, but varied in duration, peak acceleration and displacement; Group IV pulses had the same shape and peak acceleration, but varied in speed change, duration and displacement; and Group V pulses had the same peak acceleration and displacement, but varied in shape, speed change and duration. Displacement refers to the distance traveled during the collision pulse. The dummy was exposed to six blocks of 15 pulses: each of the 15 different pulses was presented once per block and presentation order within each block was randomized.

Data Reduction

Accelerations were determined from the transducer data and displacements were determined from the high speed video data. All dummy accelerations and loads were reported in local head, T1 and pelvis coordinates and all displacements were reported in global coordinates. The x-axis was oriented positive forward, the z-axis positive down and angular motion about the y-axis was positive in extension. Accelerations measured by the T1 accelerometers were rotated so that the local x-axis was initially horizontal before any subsequent calculations were made. The sagittal plane moment measured by the upper neck load cell was corrected to the AO pin, and load cell data were reported as forces and moments applied to the head by the neck. Angular rate data were digitally-compensated to reduce the sensors' high-pass frequency to 0.002 Hz (Laughlin, 1998), and angular accelerations were then computed by finite differences (16 ms window). All signals were zeroed based on their pre-impact values.

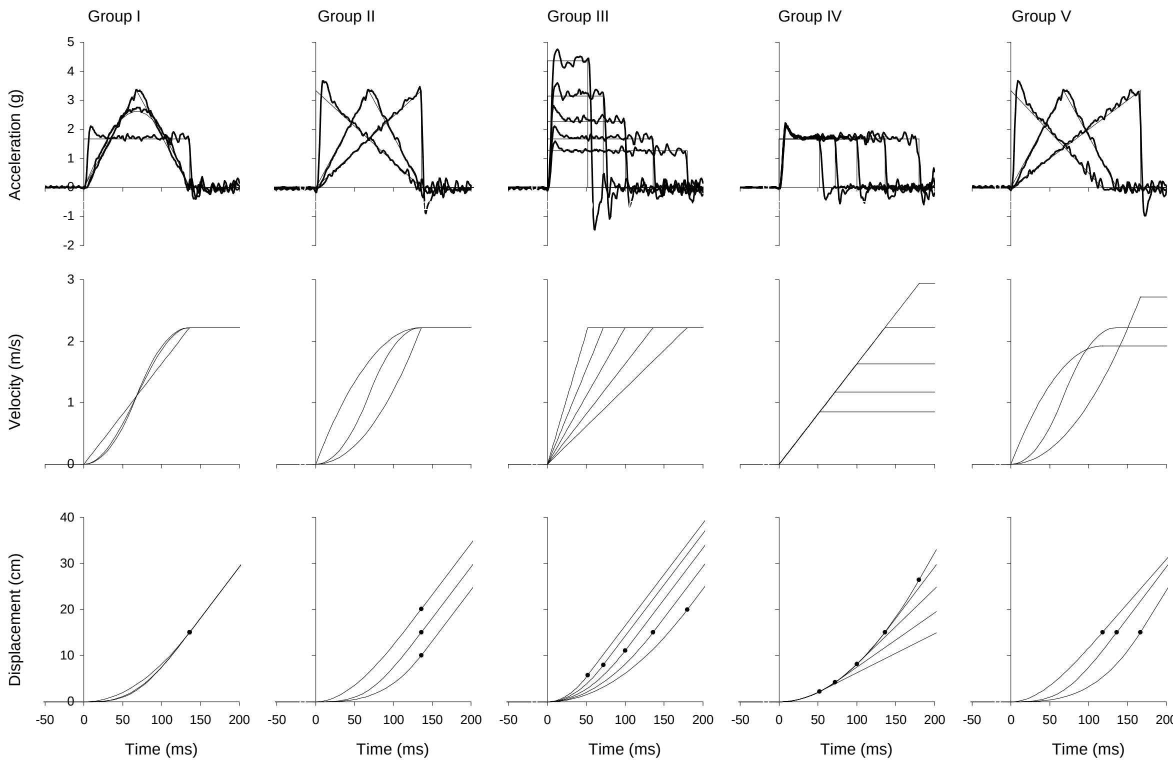


Figure 2: Idealized acceleration, velocity and displacement versus time graphs (thin lines) for the fifteen different collision pulses grouped into the five experimental groups. The thick lines in the acceleration versus time graphs depict the actual accelerations measured on the sled. The solid circles on the displacement versus time graphs depict the end of the collision pulse.

The six injury criteria were computed from the BioRID response data as follows: peak neck shear force (F_x) and peak neck moment (M_y) were measured with the upper neck load cell; peak rearward translation (retraction R_x) of the atlanto-occipital joint (pin) with respect to the T1 joint axis and peak rearward extension (θ_{rel}) of the head with respect to T1 were determined from the high-speed video data; the whiplash neck injury criterion (NIC) was computed from the relative horizontal acceleration and velocity of the head centre of mass with respect to the T1 joint axis (Boström et al., 2000); and the peak normalized neck injury criterion (N_{ij}) was calculated from the neck axial force (F_z) and the neck moment (M_y) using the intercept values for the Hybrid III mid-sized male (Eppinger et al., 1999).

Statistical Analysis

Within each group of collision pulses, a one-way analysis of variance (ANOVA) was conducted for each parameter. Post-hoc Scheffé tests were used to determine homogeneous groups, i.e., groups of pulses that were not significantly different from each other. All analyses were conducted using Statistica (v.6.0, Statsoft, Tulsa, OK) and a significance level of 0.05.

RESULTS

The time-varying dynamic responses of the head, T1 and pelvis of the BioRID were of similar form, though of variable magnitude, for the fifteen different pulses (Figure 3). In all but pulses I, J and K, all of the injury criteria peaked within 49 ± 26 ms of each other. Peak

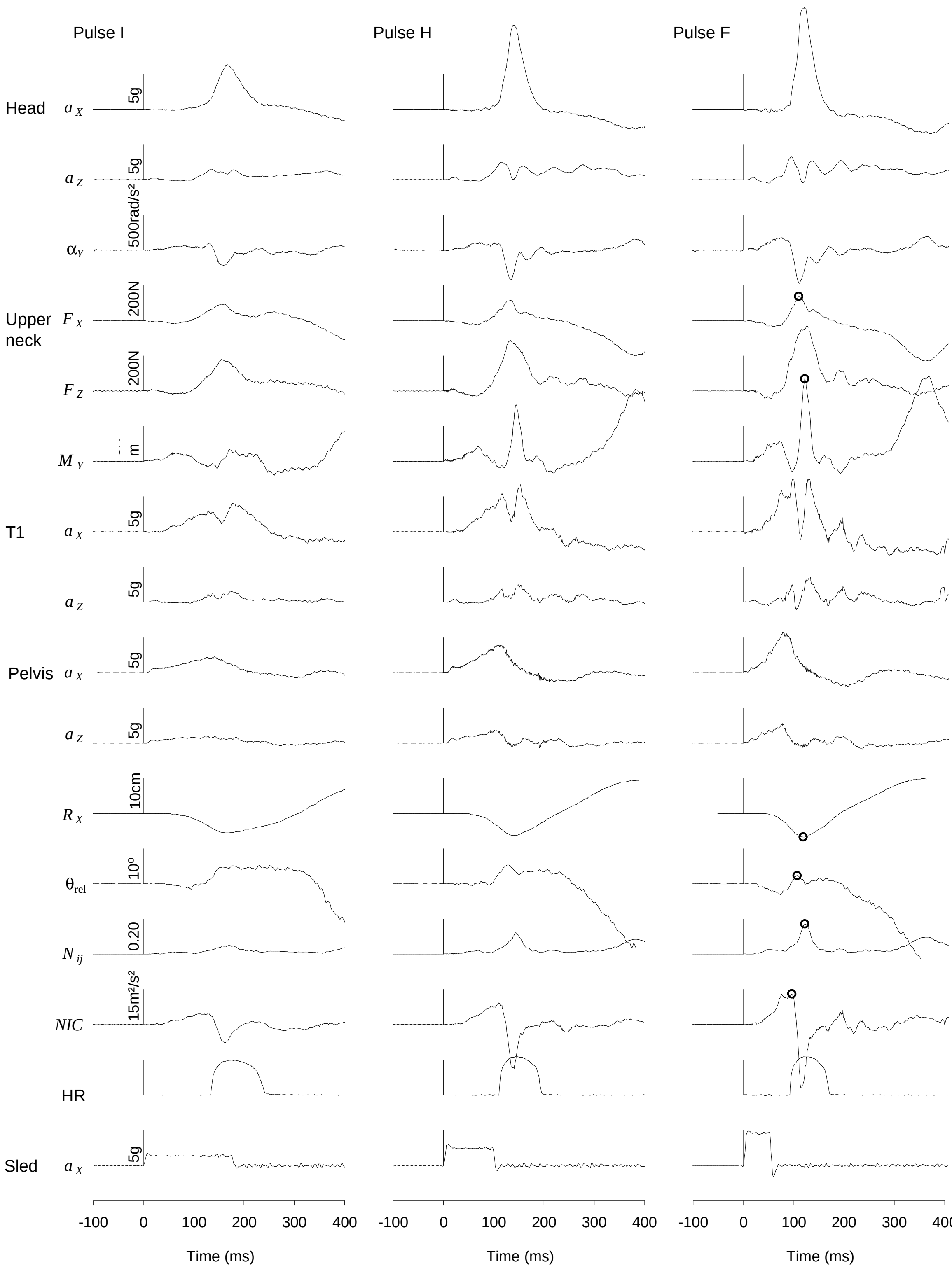


Figure 3: Sample kinematic data, kinetic data and calculated injury criteria as a function of time for three square collision pulses: on the left, pulse I (ΔV=8 km/h, Δt=180ms, a_p=1.3g, s=20cm); in the middle, pulse H (ΔV=8 km/h, Δt=100ms, a_p=2.3g, s=11.1cm); and on the right, pulse F (ΔV=8 km/h, Δt=52ms, a_p=4.4g, s=5.8cm). The scale bars for each graph are aligned with the onset of the pulse and the open circles shown for pulse F (right column) depict the peak values for the six injury criteria used in the subsequent analysis. HR depicts head restraint contact and Sled a_x depicts the sled acceleration.

Table 2. Mean (SD) of the minimum and maximum values of each injury criterion. Also shown are the range/noise ratios and the coefficients of variation.

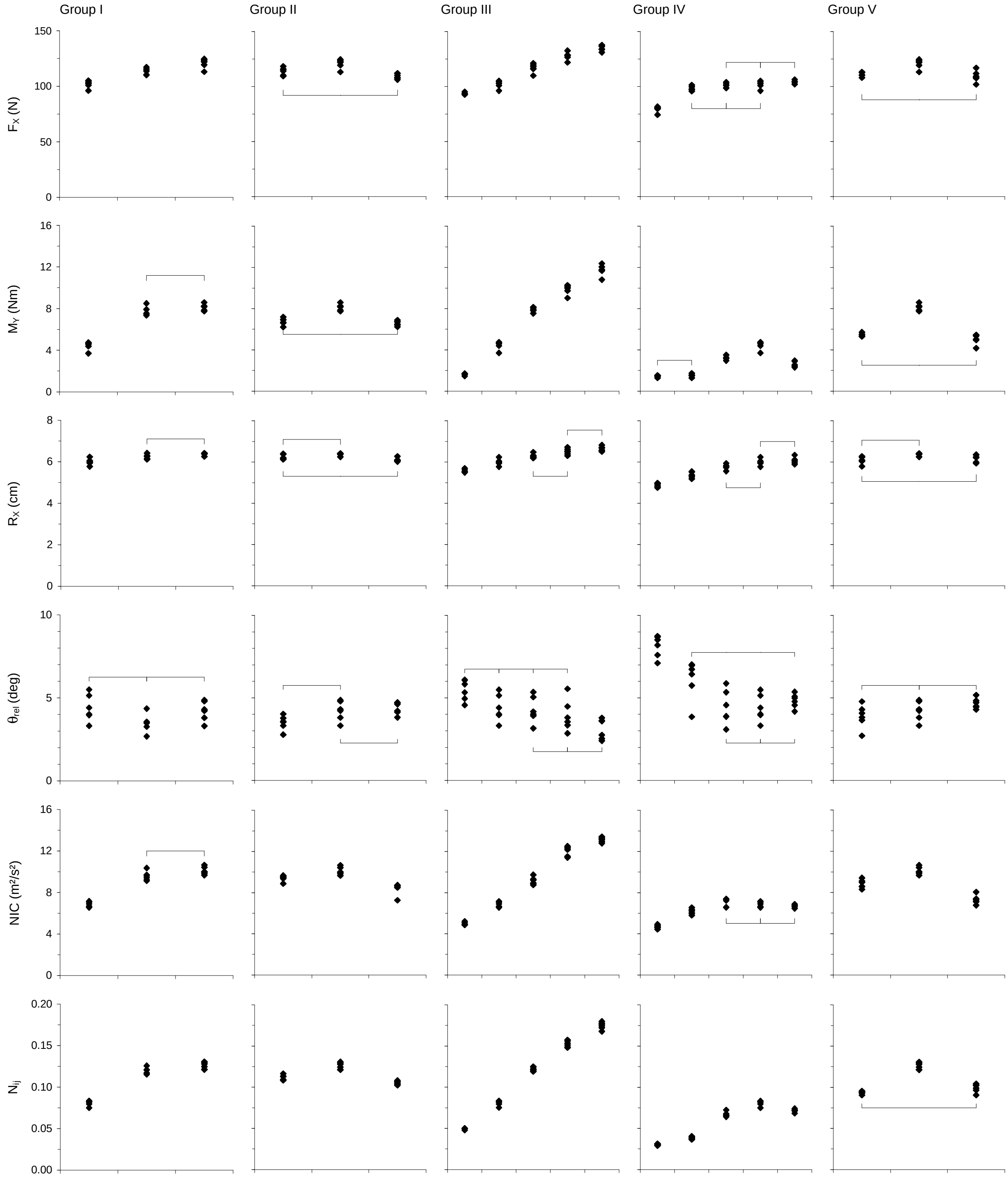
Injury Criterion	Minimum (Pulse J)	Maximum (Pulse F)	Range/Noise Ratio	Coefficient of Variation (%)
F_x N	79.5 (2.6)	135.0 (2.5)	19.6	2.6 (0.9)
M_y Nm	1.44 (0.08)	11.7 (0.5)	35.4	5.9 (2.5)
R_x cm	4.89 (0.09)	6.64 (0.11)	14.3	2.0 (0.6)
θ_{rel} deg	8.14 (0.66)	3.11 (0.62)	-7.5	15.1 (5.7)
NIC m²/s²	4.68 (0.16)	13.15 (0.24)	25.6	4.0 (1.3)
N_{ij}	0.030 (0.001)	0.175 (0.004)	51.6	3.0 (1.0)

head/neck extension (θ_{rel}) and neck moment (M_y) occurred over a more variable period of time than the other criteria (F_x, R_x, NIC and N_{ij}), all four of which peaked within 37 ± 6 ms of each other. NIC peaked first in all pulses except pulse J, in which the neck moment peaked before NIC.

Five of the six injury criteria exhibited a similarly graded response across the fifteen different collision pulse (see F_x, M_y, R_x, NIC and N_{ij} in Figure 4) and had similarly low coefficients of variation (Table 2). Based on the fewest number of homogeneous groups (horizontal bars in Figure 4), N_{ij} and NIC were best able to distinguish between the collision pulses. Peak head/neck extension (θ_{rel}) was least able to distinguish between collision pulses (Figure 4) and also had the highest coefficient of variation (15 ± 6 percent, Table 2). T1 extension angles in the global reference frame increased more rapidly with collision severity than head extension angles and thus peak relative extension of the head with respect to T1 was not positively graded to collision severity (see Groups III and IV in Figure 4). This behaviour was not suitable for an injury criterion and θ_{rel} was dropped from the analysis.

Some properties of the collision pulse had a greater effect on the injury criteria than others. Simultaneous increases in acceleration and decreases in collision duration produced the largest increase in the five remaining injury criteria (Group III, Figure 4). Simultaneous increases in both collision speed change and duration also increased the five remaining injury criteria, though this gradation was limited to pulses less than 100 ms long (Group IV, Figure 4). With collision speed change, duration and displacement held constant, increases in peak acceleration alone also increased the injury criteria values (Group I, Figure 4). And finally, within the triangular-shaped pulses, the isosceles pulse (C) generated higher injury criteria values than either the ascending-ramp or descending-ramp pulses, despite having the same speed change, duration, peak acceleration and displacement (Groups II and V).

All five injury criteria reached a minimum value in pulse J and a maximum value in pulse F. To objectively compare the sensitivity of the different criteria, the overall range of each criterion (i.e., the difference between the average value observed for pulse F and the average value observed for pulse J) was divided by a measure of that criterion's noise (i.e., the average of the standard deviations computed for each of the 15 pulses). This range/noise ratio showed that N_{ij}, which combined a large range with low noise, was the most sensitive of the whiplash injury criteria examined here (Table 2).



Pulse	A	B	C	D	C	E	I	A	H	G	F	J	K	L	A	M	N	C	O
Shape	Square	Sine	Tri-Iso	Tri-Des	Tri-Iso	Tri-Asc	Sqr	Sqr	Sqr	Sqr	Sqr	Sqr	Sqr	Sqr	Sqr	Sqr	Tri-Des	Tri-Iso	Tri-Asc
ΔV (km/h)	8	8	8	8	8	8	8	8	8	8	8	3.1	4.5	5.9	8	10.6	6.9	8	9.8
Δt (ms)	136	136	136	136	136	136	180	136	100	76	52	52	76	100	136	180	118	136	166
a_{peak} (g)	1.7	2.6	3.3	3.3	3.3	3.3	1.3	1.7	2.3	3.0	4.4	1.7	1.7	1.7	1.7	1.7	3.3	3.3	3.3
s_{accel} (cm)	15.1	15.1	15.1	20.1	15.1	10.1	20.0	15.1	11.1	8.4	5.8	2.2	4.7	8.2	15.1	26.5	15.1	15.1	15.1

Figure 4: Values for the six injury criteria (in separate rows) as a function of the collision pulses arranged into groups (in separate columns). All six tests for each criterion/pulse combination are shown separately. The horizontal bars depict homogeneous groups of pulses, i.e., groups of pulses that were not significantly different from each other in post-hoc testing.

DISCUSSION

The current experiment showed that five candidates for a whiplash injury criterion (F_x, M_y, R_x, NIC and N_{ij}) generated graded responses to kinematic changes in the collision pulse. These gradations were consistent with previously-observed increases in whiplash injury severity and duration (Krafft et al., 2002). All five injury criteria were more sensitive to changes in average vehicle acceleration than in speed change – consistent with epidemiological data in which a better correlation of whiplash injury was observed with average acceleration than with speed change (Krafft et al., 2002). Although the current results cannot be used to conclude which of these five injury criteria correlate best with whiplash injury, the results indicated that the normalized neck injury criterion (N_{ij}) had the highest range/noise ratio and was best able to discriminate between the collision pulses – both of which are important attributes for an experimental injury criterion. The other four criteria – the neck injury criterion (NIC), neck shear force (F_x), neck moment (M_y) and head/neck retraction (R_x) – also had the graded response required of a useful injury criterion, though with slightly lower range/noise ratios.

Of the six criteria, peak extension angle (θ_{rel}) of the head relative to T1 was the only criterion that appeared to be unsuitable for assessing whiplash injury under the current experimental conditions. It not only was negatively graded to collision pulse severity, but its variance was also more than double the next highest variance (Table 2). Peak extension of the head relative to the torso is the current dynamic criterion in the Federal Motor Vehicle Safety Standard 202 and appears to be the favoured criterion for the proposed revision (NHTSA, 2000). Although the collision pulses used in the current study were below the existing and proposed pulse corridors specified in Standard 202, a more likely explanation for the poor performance of θ_{rel} in the current study was the use of the BioRID dummy rather than the Hybrid III dummy specified in Standard 202. The coupling of the T1 and torso in the Hybrid III dummy is rigid, whereas the coupling of the T1 and torso in the BioRID dummy is flexible. This flexibility means that the relative head extension measured in the BioRID varies with the vertebral level against which it is referenced. This difference between dummies suggests that dummy-specific whiplash injury criteria may be required and raises questions regarding which dummy is best for evaluating whiplash injury protection.

Injury criteria are ultimately design criteria that are used to improve vehicle safety and reduce occupant injury. From this perspective, a whiplash injury criterion that is based on both a dummy with a human-like response and the actual mechanism of whiplash injury may give vehicle and seat designers the best chance of reducing or eliminating whiplash injuries. Using the Hybrid III dummy, which lacks biofidelity in rear-end collisions (Scott et al. 1993; Davidsson et al., 1999a, 1999b; Cappon et al., 2000; Siegmund et al, 2001), and an injury criterion based on head/neck extension, which may be unrelated to whiplash injury (McConnell et al., 1995; Brault et al., 1998), may lead designers to optimize seats and head restraints to pass the standard rather than minimize actual whiplash injuries.

The current study provides a link between the kinematic properties of a collision and a number of injury criteria computed from dynamic occupant responses. The link between these injury criteria and actual whiplash injuries is less well established, though based on models of 6 actual collisions, NIC values were highest in collisions that generated the longest duration of whiplash symptoms (Eriksson and Boström, 1999). Given that both the risk of long-term whiplash injuries (Krafft et al., 2002) and the injury criteria evaluated here were graded more strongly to average acceleration than to vehicle speed change, the results of the current study support the usefulness of these criteria as measures of whiplash injury risk.

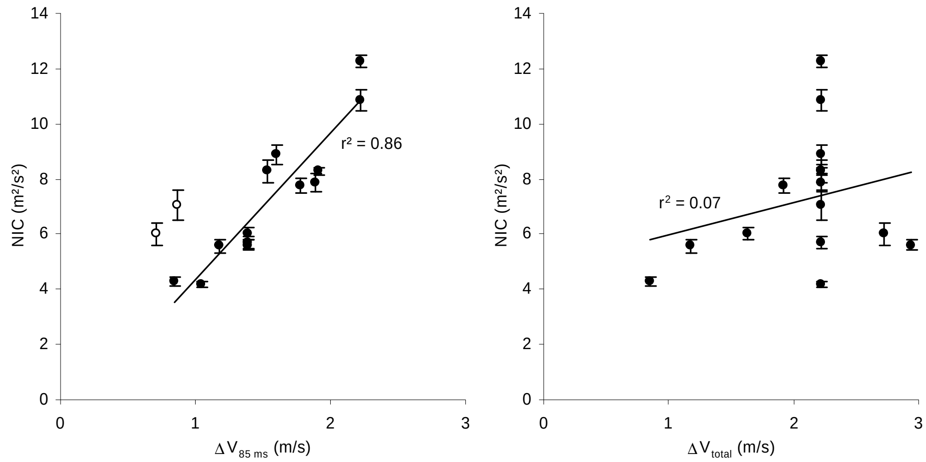


Figure 5: Variations in NIC with speed change showing an improved correlation for NIC v. ΔV_{85ms} compared to NIC v. ΔV_{total}. The open circles represent the ascending-ramp pulses, which have not been included in calculating the coefficient of determination in the left graph.

Vehicle speed change has historically been the favoured measure of collision severity when the whiplash injury potential of rear-end collision is being assessed. Recent epidemiological evidence, however, indicates that long-term whiplash injuries are more common in vehicles with stiff rear components like a towing hitch (Krafft et al., 2000) and that whiplash symptom duration and intensity correlate better with average vehicle acceleration than with vehicle speed change (Krafft et al., 2002). Assuming that the whiplash injury criteria evaluated here are related to the risk of whiplash injury, the results of the current study also indicated that average vehicle acceleration is a better measure of whiplash injury potential than vehicle speed change. No one collision-based parameter, however, captured all of the effects observed in the data. For instance, within the Group I pulses (all of which had the same speed change and average acceleration), peak vehicle acceleration affected the occupant response and injury criteria. The time of peak acceleration within a collision pulse also affected the occupant response and the injury criteria (Group II and V pulses). Other kinematic properties of the collision pulse, i.e., collision duration and displacement, appeared to have only secondary effects.

In real collisions, a vehicle's speed change is governed by the mass ratio of the two involved vehicles, the closing speed of the two vehicles, and the restitution of the combined bumper systems. The acceleration is governed by the mass of the vehicles and the series stiffness of the combined bumper systems. If the injury criteria evaluated here are related to the risk of whiplash injury, then the current data indicated that reductions in whiplash injury could be achieved by reducing bumper restitution and stiffness. Indeed, the current data suggest that the increase in bumper restitution and stiffness brought about by the increased use of rigidly-mounted and foam-core bumpers may be partly responsible for the observed increase in whiplash injury over previous decades (v Koch et al., 1994; Morris and Thomas, 1996; Temming and Zobel, 1999). These data also suggest that safety standards designed to reduce whiplash injury should treat the bumper, seat and head restraint as a system, and thus encourage vehicle manufacturers to optimize bumper/seat combinations to reduce whiplash injury.

The current results are consistent with previous work in which simultaneous changes in acceleration and duration for a fixed speed change altered the occupant responses (Nilson et al., 1994; Eriksson and Boström, 1999; Boström et al., 2000). Eriksson and Boström (1999) also observed the strongest correlation between NIC and speed change when only the speed change over the first 85 ms (ΔV_{85ms}) was used. When applied to the current data (Figure 5), the correlation between NIC and ΔV_{85ms} (r^2=0.86) was a considerably better than the correlation between NIC and total speed change (ΔV_{total}, r^2=0.07). The correlation between NIC and ΔV_{85ms} diminished to r^2=0.68 when the two ascending-ramp pulses (E and O) – a type of pulse not considered by Eriksson and Boström (1999) – were included. These findings suggest that there may be a measure based on speed change that predicts whiplash injury, though further work is required.

The general applicability of the current results is limited by the use of a single seat, a single seated posture and a single head restraint position. These factors are known to affect the occupant response (Håland et al., 1996; Svensson et al., 1996; Siegmund et al., 1999; Boström et al., 2000; Welcher and Szabo, 2001) and further work is needed to assess the affect of these variables on the relationships observed here. The current results are also limited to the BioRID dummy and additional work is needed to determine whether the Hybrid III or other whiplash dummies display similar sensitivities to difference in the kinematics of the collision pulse. The current study also relied on idealized pulses contrived to allow different combinations of kinematics parameters to be held constant or varied. Actual collision pulses likely fall between the square and triangular pulses used here, and therefore the trends observed using actual pulses may be muted compared to the trends observed here.

CONCLUSIONS

In summary, five of six proposed whiplash injury criteria exhibited graded responses to changes in collision pulse kinematics that have been previously related to increases in whiplash injury severity and duration. Based on the current data, N_{ij} had the highest sensitivity to increasing pulse severity and was best able to distinguish between the fifteen different collision pulses.

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APPENDIX

Table A1. Mean (SD) of all injury criteria for each of the collision pulses.

Pulse	F_x	M_y	R_x	q_{rel}	NIC	N_{ij}
A	102.1 (3.3)	4.47 (0.40)	5.99 (0.15)	4.39 (0.81)	6.82 (0.26)	0.081 (0.003)
B	113.7 (2.9)	7.79 (0.42)	6.25 (0.11)	3.48 (0.54)	9.57 (0.44)	0.119 (0.004)
C	120.9 (4.2)	8.14 (0.31)	6.37 (0.06)	4.22 (0.59)	10.10 (0.37)	0.126 (0.004)
D	113.9 (3.5)	6.76 (0.33)	6.24 (0.12)	3.49 (0.43)	9.41 (0.28)	0.111 (0.003)
E	109.4 (2.4)	6.58 (0.24)	6.10 (0.09)	4.36 (0.37)	8.39 (0.56)	0.105 (0.002)
F	135.0 (2.5)	11.73 (0.52)	6.64 (0.11)	3.11 (0.62)	13.15 (0.24)	0.175 (0.004)
G	127.8 (3.5)	9.90 (0.46)	6.51 (0.15)	3.93 (0.96)	12.05 (0.47)	0.152 (0.004)
H	116.8 (4.0)	7.93 (0.23)	6.28 (0.11)	4.27 (0.80)	9.14 (0.37)	0.122 (0.002)
I	93.6 (0.9)	1.63 (0.09)	5.57 (0.08)	5.47 (0.63)	4.98 (0.14)	0.050 (0.001)
J	79.5 (2.6)	1.44 (0.08)	4.89 (0.09)	8.14 (0.66)	4.68 (0.16)	0.030 (0.001)
K	98.2 (1.9)	1.52 (0.14)	5.38 (0.13)	6.13 (1.21)	6.15 (0.27)	0.039 (0.002)
L	101.0 (2.2)	3.27 (0.21)	5.77 (0.13)	4.43 (1.03)	7.08 (0.40)	0.067 (0.003)
M	103.9 (1.6)	2.60 (0.27)	6.07 (0.16)	4.82 (0.42)	6.66 (0.17)	0.072 (0.002)
N	111.3 (2.0)	5.44 (0.16)	6.08 (0.17)	3.89 (0.70)	8.83 (0.41)	0.093 (0.002)
O	109.3 (5.0)	5.05 (0.49)	6.18 (0.18)	4.66 (0.32)	7.28 (0.43)	0.099 (0.005)

DISCUSSION

PAPER: **The Effect of Collision Pulse Properties on Six Proposed Whiplash Injury Criteria**

PRESENTER: ***Dr. Gunter Siegmund, School of Human Kinetics, University of British Columbia and MacInnis Engineering Associates of Canada***

QUESTION**:**
I'm curious about the data. I mean, I enjoyed your whole talk, but that caught my attention. If I were to model the human OC-1 joint, I would put a pin in the middle of your head and the dummy has a pin a couple inches remote from that.

ANSWER: That's right.

Q: Where the joint structure actually is.

A: Right.

Q: You're getting strange theta data and a lot of head rotation about that pin. What do we make of that?

A: Well, the dummy's probably not a good model of the human for head rotation, or head versus neck rotation. So, it may not be that the metric's bad. It's just the metric's bad in this dummy.

Q: Absolutely. Yeah. That's why I said–Yeah. We've got a limitation with the BioRID dummy. In the Hybrid III where there isn't the possibility of retraction, it may be a perfectly fine metric. But, I guess the important take-home message is that your metric may be dummy-dependant. Thanks.

Q: *Guy Nusholtz, Daimler/Chrysler*
Along that same line of the, being dummy-dependent, one of the experiences that we've had with the BioRID dummy is the response varies not, not only in terms of the between BioRID and the BioRID II and the Hybrid III, but also between the different BioRID you get different results depending on which BioRID you use.

A: Right.

Q: How, how recent–I mean, I've been told that a lot of that has been corrected.

A: Right.

Q: It may or may not, but it could be just: Your results could be dependent just on that specific BioRID; and you use a different BioRID, and you get a different result.

A: I understand. Agnes Kim presented some data last year at STAPP that showed there was multiple load paths through the BioRID that–

Q: That is correct.

A: That they had. We did the inverse dynamics on this particular BioRID and there was no multiple load path. This is the identical dummy used in all of the Chalmers and Volvo studies. It's their dummy. So from that perspective, it's comparable to those. It may not be–Well, I know it's not comparable to the data that Agnes presented because that was, I think it was a pre-production version of BioRID that had multi-load path.

Q: At the time, it was a production version.

A: Yeah. Okay.

Q: After that, they decided it wasn't a production. So, it became a non-production after the fact.

A: I don't want to get into it.

Q: Okay. Thank you.

Q: *Matt Phillipens, TNO*
It's interesting that you compare the delta v and the acceleration. That one of the most recent information I've seen is that there's quite a relation between, specifically, the NIC and the time duration of acceleration. And, that will be interesting to use these data because you can do that easily to check that if as the time duration of the acceleration is–The dummy need a certain time to come in contact with the seat.

A: Right.

Q: So before the dummy feels the acceleration of the car–And, that's one of the things that are found that could easily be done with the data you have.

A: Yeah. We can talk about that. I only presented half the data. We've got 45 other tests with different pulse shapes, saw tooths, and sign waves and various other things. So, it's likely buried in the data.

Q: Thank you.