AN EVALUATION OF VARIOUS NECK INJURY CRITERIA IN VIGOROUS ACTIVITIES

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ABSTRACT

Several different injury criteria have been proposed to predict the likelihood of AIS 1 neck injury in automobile collisions. The purpose of this study was to evaluate the ability of various injury criteria to predict the presence or absence of minor injury in human volunteers subjected to vigorous activities in which the neck was loaded in a direction similar to an automotive rear impact. Twenty (20) volunteers were subjected to five test scenarios: a soccer ball impact to the forehead, a voluntary hand strike to the forehead, voluntary shaking of the head, plopping down in a seat, and a vertical drop while seated supine in a chair. Most criteria predicted a low risk of injury in all activities. However, high values for the Neck Injury Criterion (NIC) (> 15 m^2/s^2) were recorded in a large number of tests in the soccer ball, head strike, and chair tip scenarios. No long-term neck symptoms were reported after any of these tests, which was a significantly more favorable outcome than what would be predicted by the NIC risk curve proposed by Eriksson and Kullgren (2006) ($p < 0.001$). The finding that NIC greatly overpredicts the risk of long-term (> 1 month) AIS 1 neck injury in non-automotive scenarios casts doubt on its biomechanical validity.

Key words: Biomechanics, Injury Criteria, Neck, Volunteers, Whiplash

MINOR (AIS 1) NECK INJURY following a rear end collision is commonly reported. Short-term neck symptoms felt to be the product of muscle strain have been produced in human volunteers subjected to low speed rear impacts (McConnell et al., 1995). However, in cases of long-term neck complaints without radiographic or other objective signs of a lesion, the nature of the injury is much less clear. There is no consensus as to which anatomical structures are involved in long-term AIS 1 neck injury, much less the mechanisms of injury and associated tolerance values. Several theories have been proposed. Aldman et al. (1986) hypothesized that long-term AIS 1 neck injury is associated with damage to cervical nerve roots due to a "water hammer" effect caused by a hydrodynamic pressure change in the spinal canal during the retraction phase of a rear impact. Based on this hypothesis, Boström et al. (1996) proposed the Neck Injury Criterion (NIC), which predicts injury based on the relative A-P acceleration and velocity between the head and the neck. A tolerance value of 15 m^2/s^2 obtained from pig testing by Svensson et al. (1993) was proposed for humans. Recently, Eriksson and Kullgren (2006) proposed a risk curve for long-term AIS 1 neck injury using NIC based on field data and computer modeling, where "long-term" was defined as "symptoms lasting more than one month." Boström et al. (2000) suggested that long-term AIS 1 neck injury in frontal impacts can be correlated not only to $NIC_{protraction}$, but also to neck loads, using lower tolerance limits for N_{ij} or for upper neck flexion moment. N_{ij} is meant to predict injury to the anterior longitudinal ligament at the junction of the skull and upper neck due to combined tension and extension loading. Tolerance limits for AIS 3+ injury were established based on scaled data from out-of-position airbag deployment tests in which loads measured by a 3 year-old child dummy were mapped to injuries observed in piglets in matched tests (Mertz and Prasad, 2000). The combined loading formulation used in the N_{ij} criterion was adapted by Schmitt et al. (2001), who proposed a linear combination of normalized shear force and bending moment that they called N_{km}. N_{km} is meant to predict long-term AIS 1 neck injury in rear impacts that is caused by stretching of the facet joints due to shear (Yang and Begeman, 1996; Deng et al., 2000). Tolerance limits were based on loads and moments that volunteers were able to withstand without injury (Mertz and Patrick, 1967). Panjabi et

al. (1999) hypothesized that long-term AIS 1 neck injury is caused by subfailure ligament injury. They proposed the Intervertebral Neck Injury Criterion (IV-NIC), which predicts injury based on the relative rotation of adjacent vertebrae. Tolerance limits were established by cadaveric testing. No injuries were actually observed in the cadavers; rather, occult ligament damage was inferred from a change in the viscoelastic properties of the spine. Without identifying a specific mode of injury, Viano and Davidsson (2002) hypothesized that long-term AIS 1 neck injury is caused by excessive motion between the head and the neck. They proposed the Neck Displacement Criterion (NDC), which is based on the relative displacement between the occipital condyles and the T1 vertebrae. Tolerance limits were not specified for humans, but initial working performance guidelines were suggested for the Hybrid III and BioRID dummies.

The injury criteria described above are based on some profoundly different assumptions regarding the mechanism of long-term AIS 1 neck injury. Most of the criteria have been developed specifically to aid in the evaluation of seat design in the context of rear impact sled testing using a crash test dummy. The NIC, in conjunction with the BioRID dummy, is currently being used for consumer testing in Europe (Viano and Davidsson, 2002). The N_{ij} and NDC have tolerance limits that are specifically mapped to the Hybrid III dummy response. The N_{km} can also be measured in standard dummies, but the IV-NIC cannot. In the context of automotive rear impacts, the various injury criteria generally correlate with collision severity and with each other, and are therefore generally predictive of injury in the field. However, each of the above criteria purports to be biomechanically based, and therefore should also be able to accurately predict the presence or absence of long-term AIS 1 neck injury in other loading scenarios besides automotive rear impacts. The purpose of this study was to obtain measurements of various injury criteria in human volunteers subjected to vigorous activities in which the head was forced rearward with respect to the neck similar to an automotive rear end collision. This information provides a basis for comparing the injury potential of a rear impact, which is an unusual event, to more common activities where the injury potential is better understood.

METHODS

Twenty (20) human volunteer subjects participated in the study, all of whom were employees of Biodynamic Research Corporation. Volunteers were selected to obtain a representative sample of the general population, and included twelve (12) males and eight (8) females spanning a wide range of ages (26 – 58 yrs, mean 44 yrs), heights (150 – 191 cm, mean 172 cm), and weights (54 – 99 kg, mean 80 kg). Plain film lateral lumbar, cervical, and head x-rays were taken of each subject to rule out significant pathology. Written consent was obtained from each participant and the study protocol was approved for human use by both an internal Research Review Board and an external Institutional Review Board (IRB). The study protocol involved five test scenarios, some of which involved progressively increasing levels of intensity. Participants were free to abstain from any test for any reason. All tests for a particular scenario were completed at one time, and different test scenarios were spaced at least one week apart. Subjects reported any symptoms related to testing to a physician (CEB).

Volunteers were instrumented with tri-axial accelerometers (Endevco 7596, 30 g) attached to custom-fit mounts that were securely strapped around the lumbar and upper thoracic regions. Bite blocks made from dental impressions of each volunteer were instrumented with two accelerometers (Endevco 7265-HS, 20 g) and one angular rate sensor. Initially, a magneto-hydrodynamic (ATA ARS-01) angular rate sensor was mounted to the bite block and positioned outside the mouth. This arrangement caused unacceptable vibrations in the angular rate signal in initial tests involving a head impact, so the MHD was replaced with a smaller angular rate sensor (DTS ARS1500k) that was rigidly mounted to the bite block and fit inside the mouth (Figure 1) and the tests were repeated. For hygienic reasons, the instrumented bite block was wrapped in a thin plastic bag before being placed in the subjects' mouths. Instrumentation data were collected using a TDAS-PRO 32-channel rack (DTS) at 10 kHz for the head impact scenarios, and 2 kHz for the other scenarios. All instrumentation data from human subjects were digitally filtered to CFC180. Instrumentation data from the Hybrid III dummy was filtered to CFC600. High speed digital video (Phantom v7.1, Vision Research, Inc.) was collected at 3 kHz for the head impact scenarios, and 1 kHz for the other scenarios. Video data were analyzed by tracking various points using WINanalyze Tracking 3D Software (Mikromak, Inc.).

The five test scenarios included two head impact scenarios: a soccer ball impact to the forehead and a voluntary hand strike to the forehead, and three other scenarios: voluntary head shaking, plopping down in a rigid seat, and a vertical drop while seated supine in a chair (chair tip). All scenarios involved loading in the sagittal plane without significant off-axis components. For the soccer ball impacts, participants stood near a custom-made apparatus that was aligned to shoot a regulation adult size 5 soccer ball inflated to 55 kPa horizontally such that it struck the subjects in the forehead (Figure 2). Up to four tests were conducted on each subject at increasing ball speeds: one low speed impact at 5 m/s, and three moderate speed impacts at 8.5, 10, and 11.5 m/s. The participants remained stationary and did not attempt to actively head the soccer ball. In fact, the ball was released and struck the subjects' foreheads before they had a chance to react. For the voluntary hand strikes, subjects were asked to strike themselves in the forehead with the heel of their hand as hard as they were willing to do it (Figure 3). Three trials were conducted for each subject. In the voluntary shaking scenario, subjects were asked to shake their head back and forth in the sagittal plane as vigorously as they could (Figure 4). One trial of 5 – 10 head shakes in rapid succession was conducted for each volunteer. In the seat plop scenario, subjects were asked to aggressively sit down on a rigid seat (Figure 5). Three trials were conducted for each subject. In the chair tip scenario, subjects were seated supine in a rigid chair that was dropped from successively increasing heights of 5, 10, and 15 cm measured to the mid-back (Figure 6). There was no head or upper back support, so the subjects had to tense their neck muscles to hold their head up against gravity. A padded headrest was placed on the chair to protect against hyperextension of the neck during testing, but no contact between the head and headrest occurred in any of the tests.

Fig. 1. Photograph of an instrumented bite block.

Fig. 2. Video capture of a soccer ball impact.

Fig. 3. Video capture of a voluntary hand strike to the forehead.

Bite block accelerations were transformed using standard rigid body dynamics equations to determine the acceleration at the center of gravity of the head and the occipital condyles. These locations were identified from anatomical landmarks that were located precisely on scaled images in which lateral head x-rays and photographs of each volunteer were superimposed (Figure 7). The coordinate system of the head was defined using the top of the external auditory meatus (in x-rays) or the tragus (in photographs) as the origin. The x-axis pointed anteriorly through the inferior orbital rim along the Frankfort plane. The z-axis pointed inferiorly, perpendicular to the Frankfort plane, and the y-axis pointed to the right, in accordance with the SAE sign convention. The location of the center of gravity of the head in all volunteers was assumed to be 8.4 mm anterior and 31 mm superior to the origin of the head coordinate system, which is the mean value for a 50th percentile male (Schneider et al., 1983). The location of the occipital condyles was identified radiographically for each volunteer, and ranged from 8 – 22 mm (mean 13 mm) posterior and 49 – 66 mm (mean 58 mm) inferior to the estimated location of the head center of gravity.

Fig. 4. Video capture of a voluntary head shake.

Fig. 5. Video capture of a seat plop.

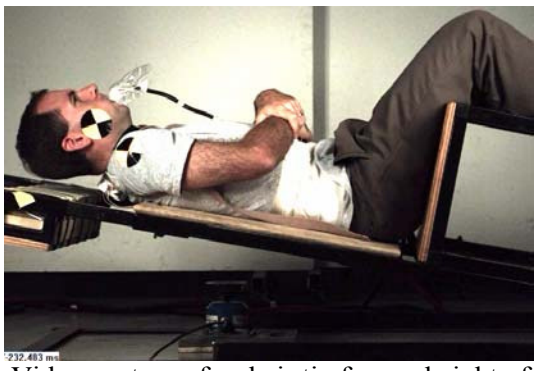

Fig. 6. Video capture of a chair tip from a height of 15 cm.

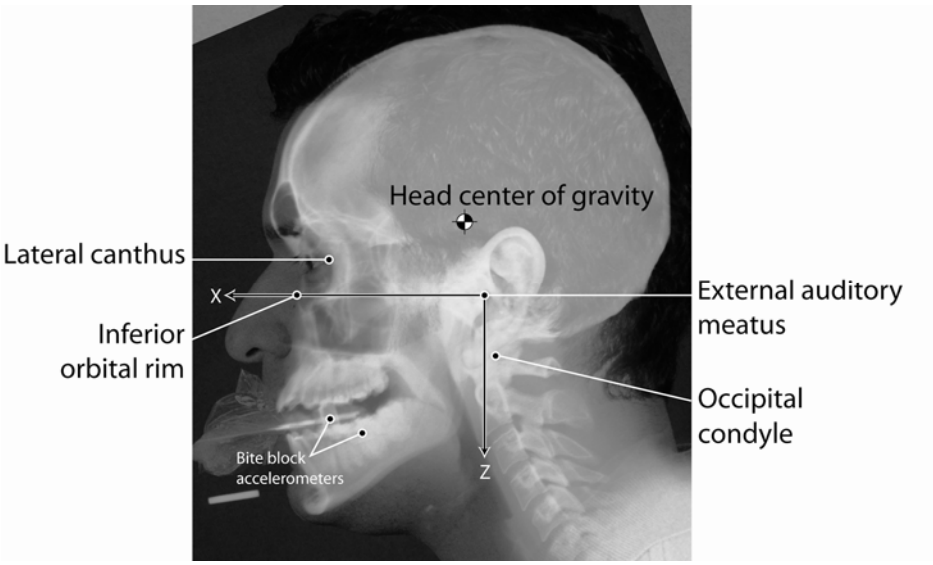

Fig. 7. Example of x-ray and photograph overlay used to locate anatomical landmarks in the head coordinate frame.

Sagittal plane loads and moments at the occipital condyles were calculated using standand inverse dynamics equations. The head mass of each volunteer was estimated using the regression equations of Clauser et al. (1969), and ranged from 4.3 – 5.5 kg (mean 5.0 kg). The sagittal plane moment of inertia for the head was estimated from the head mass using the data from Beier et al. (1980), and ranged from 217 – 327 kg-cm^2 (mean 279 kg-cm^2). In the case of soccer ball impacts, the inverse dynamics equations require knowledge of the force applied to the head by the ball. This information was obtained from video analysis. The displacement of the center of the ball at each time point was first measured in the earth-based video reference frame. The displacement data were differentiated and filtered to 100 Hz to calculate ball velocity, then differentiated again to calculate ball acceleration. The acceleration of the ball was multiplied by its mass (434 g) to obtain the force applied to the ball in an earth-based reference frame. The resultant ball force was assumed to be directed through the center of the contact patch between the ball and the forehead, also determined from video analysis. The ball force vector was then transformed to the moving head coordinate system at each time point and input into the inverse dynamics equations. The subtraction of relatively large inertial terms from the relatively large ball force term resulted in high-frequency noise that had a high amplitude relative to the small value of the calculated bending moment about the occipital condyles. This noise was removed by digitally filtering the occipital condyle forces and moments calculated from inverse dynamics to 50 Hz, which is considerably higher than the expected frequency content of forces and moments in the human neck. This approach was validated by performing matched soccer ball impact tests on the Hybrid III dummy and comparing the occipital condyle loads and moments measured by the upper neck load cell to the corresponding values calculated using inverse dynamics. No attempt was made to calculate the force applied to the head by the hand in the voluntary hand strike tests, although the approach velocity of the hand was calculated from video.

Where possible, N_{ij}, N_{km}, and NIC were calculated for each test. N_{ij} was given by:

$$N_{ij} = \frac{F_z}{F_{zc}} + \frac{M_{ocy}}{M_{yc}} \tag{1}$$

where F_{zc} is the critical axial force value equal to 6806 N in tension or 6160 N in compression, and M_{yc} is the critical moment value equal to 310 Nm in flexion and 135 Nm in extension, in accordance with the current U.S. standard for a Hybrid III 50th percentile male dummy (NHTSA, 2004). N_{km} was calculated according to the formula:

$$N_{km} = \frac{F_x}{F_{\text{int}}} + \frac{M_y}{M_{\text{int}}} \tag{2}$$

where F_{int} is the intercept shear force value equal to 845 N, and M_{int} is the intercept moment value equal to 88.1 Nm in flexion and 47.5 Nm in extension (Schmitt et al., 2001). NIC was calculated based on the relative acceleration and velocity between the occipital condyles and T1 in the A-P direction (x-axis):

$$NIC = 0.2 \cdot a_{rel} + v_{rel}^2 \tag{3}$$

NIC may also be calculated based on the relative acceleration and velocity between the head center of gravity and T1 using the same formula (Boström et al., 2000). In the present study, NIC was calculated both ways and was designated NIC_{oc} or NIC_{cg}.

The probability of long-term AIS 1 neck injury was estimated for each test based on the median NIC risk curve proposed by Eriksson and Kullgren (2006). Explicit values defining the risk curve were not provided in the paper, so a piecewise linear approximation was derived by interpolating between points obtained from digitizing Figure 3 in the paper. Using this method, it was estimated that a NIC of 7.5 corresponded to a 0% risk of injury, a NIC of 22.5 corresponded to a 36% risk of injury, and a NIC of 35 corresponded to a 100% risk of injury. For each test scenario, the percentage of *n* subjects expected to sustain long-term (> 1 month) AIS 1 neck injury was calculated by summing the probability of injury *p* associated with the highest NIC value recorded for each subject:

$$E_{inj} = \sum_{i=1}^{n} p_i \tag{4}$$

The probability that no subject in the cohort would sustain a long-term (> 1 month) AIS 1 neck injury was also calculated:

$$p(no\ injuries) = \prod_{i=1}^{n} (1 - p_i) \tag{5}$$

RESULTS

A total of 240 human subject tests were conducted under the five scenarios studied (complete results are in the Appendix). The methodology for obtaining biomechanical data was demonstrated to be accurate even in the most demanding test scenarios, which were the higher speed soccer ball impacts. In those tests, accelerations measured at the bite block were relatively low in amplitude and biphasic in shape. However, when the acceleration was transformed to the location of the head center of gravity, the magnitude of the acceleration increased and the shape became more sinusoidal (Figure 8). The shape of the head center of gravity acceleration trace then closely matched the shape of the ball force trace, as was expected (Figure 9). The inverse dynamics methodology for calculating the forces and moment at the occipital condyles gave results that closely matched the measurements of the upper neck load in the Hybrid III dummy (Figure 10). Although some of the axial force signal in the dummy was not captured by the inverse dynamics method due to filtering, it was felt that the high frequency ringing (~ 125 Hz) in the dummy axial force measurement was not a biofidelic response that needed to be captured in the human subjects.

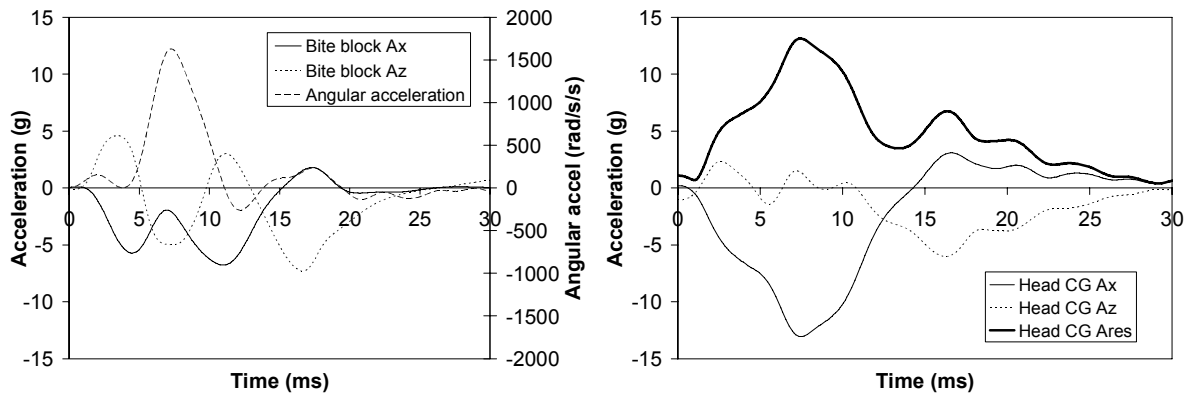

Fig. 8. Accelerations measured at the bite block (left) and transformed to the head center of gravity (right) for a 8.5 m/s soccer ball impact on a human subject.

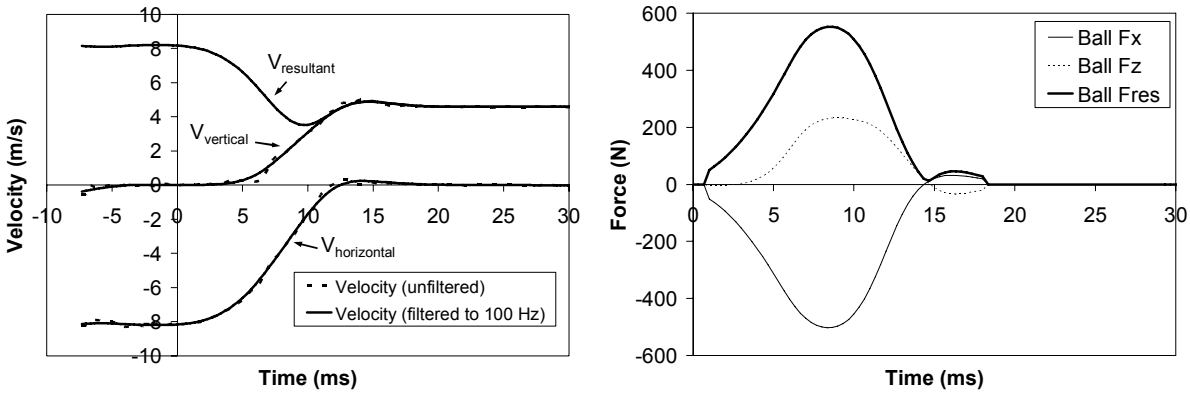

Fig. 9. Soccer ball velocity calculated from video data (left) was differentiated to determine the ball force acting in the head frame (right) (same test as shown in Figure 8).

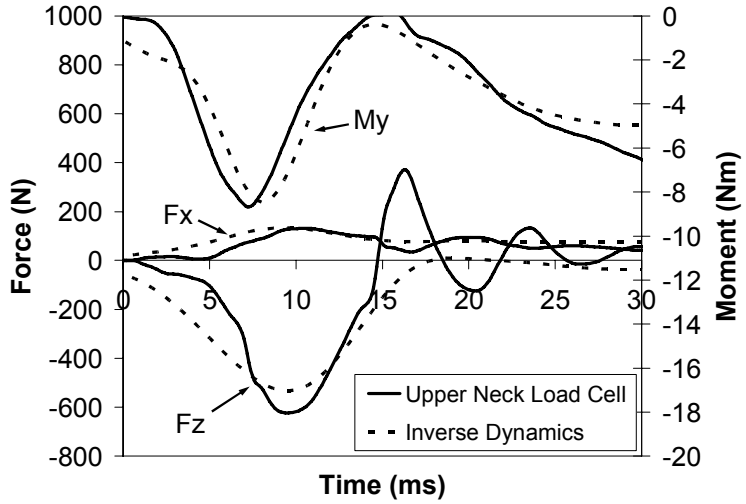

Fig. 10. Measured and calculated forces and moment at the occipital condyles in an 11.5 m/s soccer ball test with the Hybrid III dummy.

The five different test scenarios produced a wide variety of head acceleration profiles. The soccer ball impacts delivered a glancing blow that caused the head to experience relatively more angular than linear acceleration. In contrast, the voluntary head strike appeared to be directed very near the center of gravity of the head, producing almost purely linear acceleration (Figure 11). Hand velocities ranged from 2.2 – 4.5 m/s just prior to striking the head in those tests. The other scenarios that did not involve a direct head impact generated much lower head accelerations than the scenarios that did involve a direct head impact. With the exception of one outlying data point for a seat plop test, all subjects experienced a peak resultant head acceleration less than 10 g in tests not involving a head impact (Table 1). The maximum Head Injury Criterion (HIC) value calculated for any test was 10.

Although all the scenarios caused the neck to be forced generally rearward with respect to the torso, the magnitude and timing of the calculated upper neck loads was quite varied. For most scenarios, the magnitudes of the peak upper neck shear force and extension moment were in the same range as those seen in low speed (< 10 kph delta-V) rear end collisions (Kroonenberg et al., 1998; Vijayakumar et al., 2006). Although the soccer ball impacts produced the highest head accelerations, the calculated neck loads were modest. Mean values for the peak positive shear force (head forced rearward with respect to the torso), axial compression, and extension moment increased with increasing ball speeds, but never approached the N_{ij} or N_{km} critical values. Compared to the higher speed soccer ball tests, the chair tip tests resulted in similar posterior shear loads on the neck and lower axial loads. The chair tip tests required the subjects to actively tense their neck muscles prior to being dropped to hold their head up against gravity. During the drop, the active muscle tension of the neck flexors caused the neck to flex. Upon landing, the flexed neck experienced axial compression as

well as posterior shear and extension. Because the neck muscles were tensed, the neck remained flexed and did not extend enough even to return to a neutral orientation during any of the tests. Since the neck flexion during the drop increased with increasing drop height, the axial forces imparted to the neck during landing increased with increasing drop height, although the shear forces and bending moments remained relatively unchanged. Of all the scenarios, the seat plop scenario resulted in the lowest overall neck loading. The head shaking scenario produced the lowest axial neck loading of all the scenarios, but relatively high negative shear loads (head forced forward with respect to the torso) and flexion moments in the neck. Unlike the other scenarios, values were higher in negative shear and flexion in the head shaking tests. This occurred because all the neck forces were generated from internal muscle tension, rather than external forces, and the neck extensor muscles are stronger than the neck flexors.

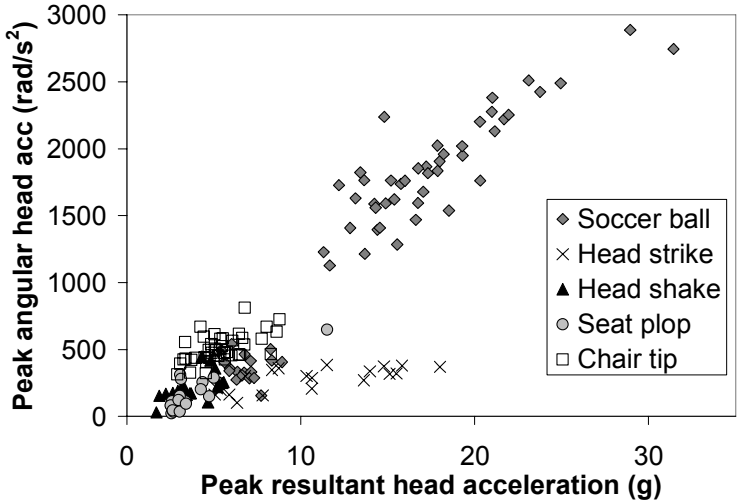

Fig. 11. Linear and angular head accelerations in each scenario.

Table 1. Mean peak values of various parameters with ranges in parentheses for all test scenarios.

Scenario	n	a_{cgres} (g)	α_y (rad/s^2)	OC F_x (N)	OC F_z (N)	OC M_y (Nm)
Soccer ball 5 m/s	20	6.8 (5.4 – 8.9)	361 (156 – 541)	95 (39 – 146)	-263 (-335 – -183)	-3 (-10– 1)
Soccer ball 8.5 m/s	18	15.1 (11.3 – 23.8)	1597 (1127 – 2425)	94 (37 – 158)	-318 (-510 – -231)	-6 (-14 – -1)
Soccer ball 10 m/s	16	17.7 (12.8 – 23.1)	1879 (1408 – 2510)	121 (28 – 169)	-354 (-440 – -272)	-8 (-17 – -3)
Soccer ball 11.5 m/s	14	21.4 (14.8 – 31.4)	2217 (1760 – 2888)	144 (99 – 187)	-414 (-526 – -279)	-9 (-18 – -2)
Head strike	20	10.5 (5.1 – 18.0)	277 (102 – 385)	n/a	n/a	n/a
Head shake	20	3.8 (1.7 – 5.6)	209 (30 – 446)	-151 (-248 – -55)	-105 (-217 – -33)	15 (6 – 23)
Seat plop	18	3.7 (2.5 – 11.5)	169 (22 – 649)	51 (9 – 218)	-167 (-481 – -114)	-4 (-18 – 0)
Chair tip 5 cm	18	4.6 (2.9 – 7.8)	505 (315 – 668)	122 (72 – 268)	-198 (-274 – -131)	-14 (-36 – -6)
Chair tip 10 cm	14	6.1 (3.6 – 8.8)	520 (322 – 727)	131 (76 – 215)	-287 (-363 – -178)	-15 (-26 – -7)
Chair tip 15 cm	10	6.0 (4.5 – 8.3)	496 (403 – 811)	139 (91 – 206)	-292 (-421 – -219)	-15 (-23 – -9)

A number of subjects reported short-term symptoms as a result of testing (Appendix). Almost all of these symptoms fell into one of two categories: head complaints related to soccer ball testing, and

neck complaints related to muscle soreness as a result of the head shaking or chair tip tests. The head shaking and chair tip tests were the only scenarios that required active neck muscle tension on the part of the study participants, and the subjects typically characterized their neck complaints after these tests as being due to muscle exertion or fatigue. No long-term AIS 1 neck injuries or any other long-term injuries of any kind were reported. Values for the various injury criteria that were calculated were not significantly higher in the subjects who reported symptoms compared to the subjects who did not report symptoms.

The load-based injury criteria predicted a low risk of neck injury in all tests (Table 2). The highest N_{ij} value calculated for any test was 0.31, and there was only one test in which the calculated N_{km} exceeded 1. N_{km} was highly correlated with Nij for all of the scenarios (Figure 12). In general, the N_{km} value was approximately four times higher the N_{ij} value. In the soccer ball tests, N_{ij} values were relatively higher for a given N_{km}, due to the greater compressive forces in the neck. In the head shaking tests, N_{km} values were relatively higher for a given N_{ij}, due to the greater neck shear forces.

Table 2. Mean values of neck injury criteria with ranges in parentheses for all test scenarios.

Scenario	n	NIC_{cg} (m^2/s^2)	NIC_{oc} (m^2/s^2)	N_{ij}	N_{km}
Soccer ball 5 m/s	20	13.2 (10.0 – 17.7)	10.5 (6.0 – 13.5)	0.06 (0.04 – 0.08)	0.18 (0.10 – 0.28)
Soccer ball 8.5 m/s	18	29.8 (22.4 – 41.5)	14.3 (10.4 – 20.5)	0.09 (0.04 – 0.15)	0.25 (0.13 – 0.48)
Soccer ball 10 m/s	16	34.8 (26.2 – 47.8)	16.3 (8.9 – 22.5)	0.10 (0.06 – 0.18)	0.31 (0.19 – 0.55)
Soccer ball 11.5 m/s	14	41.4 (29.7 – 59.0)	20.1 (10.6 – 29.8)	0.13 (0.07 – 0.19)	0.39 (0.16 – 0.54)
Head strike	20	19.7 (9.9 – 33.6)	17.7 (7.5 – 33.3)	n/a	n/a
Head shake	20	6.0 (3.0 – 9.6)	4.4 (1.4 – 6.9)	0.08 (0.02 – 0.15)	0.37 (0.14 – 0.63)
Seat plop	18	6.5 (1.2 – 18.4)	5.5 (1.2 – 13.0)	0.06 (0.03 – 0.23)	.21 (.08 – .71)
Chair tip 5 cm	18	14.0 (8.1 – 20.6)	10.5 (6.0 – 16.4)	0.12 (0.05 – 0.31)	0.44 (0.18 – 1.14)
Chair tip 10 cm	14	18.1 (10.8 – 26.1)	13.2 (5.2 – 21.6)	0.13 (0.08 – 0.22)	0.49 (0.26 – 0.87)
Chair tip 15 cm	10	16.0 (9.3 – 21.7)	10.9 (7.8 – 15.6)	0.12 (0.07 – 0.19)	.48 (0.27 – 0.73)

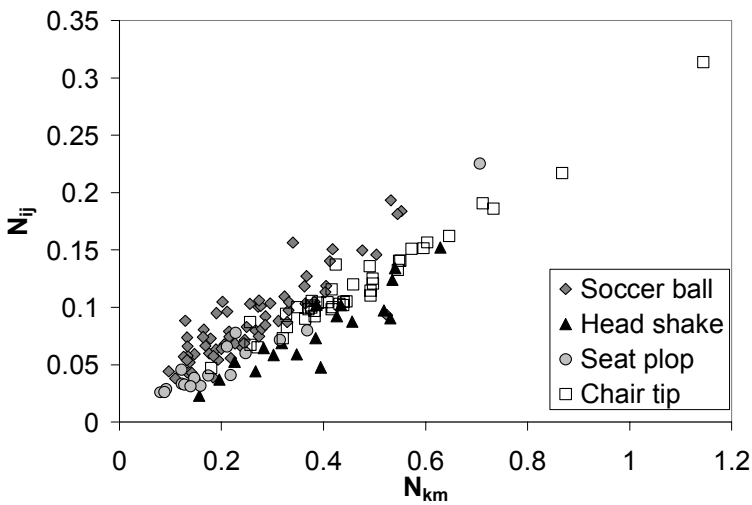

Fig. 12. Correlation between N_{ij} and N_{km}.

In contrast to the low values of N_{ij} and N_{km} that were calculated in all scenarios, high values of NIC were calculated in some scenarios. The calculation of NIC depends greatly on the relative magnitude and timing of the head and T1 acceleration pulses. In the head impact scenarios, the head acceleration pulse was so short that almost no acceleration developed at T1 and very little relative velocity was developed between the head and T1 (Figure 13). The NIC (equation 3) was therefore directly proportional to the x-axis head or occipital condyle acceleration in the head impact tests. In the chair tip tests, high NIC values developed because the x-axis T1 acceleration exceeded the lagging x-axis acceleration of the head and occipital condyles, as occurs in automotive rear impacts. In the head shaking scenario, the peak NIC values actually occurred when the neck was in a protracted position and the action of the neck extensor muscles was causing the head to be accelerated rearward with respect to T1. In the higher speed soccer ball impacts with high angular accelerations, the acceleration at the center of gravity of the head was much higher than the acceleration at the occipital condyles, which resulted in NIC_{cg} being much higher than NIC_{oc}. In the other test scenarios, the NIC_{cg} and NIC_{oc} values were similar (Figure 14). Peak NIC values generally occurred slightly earlier or at about the same time as peak N_{ij} and N_{km} values in all scenarios.

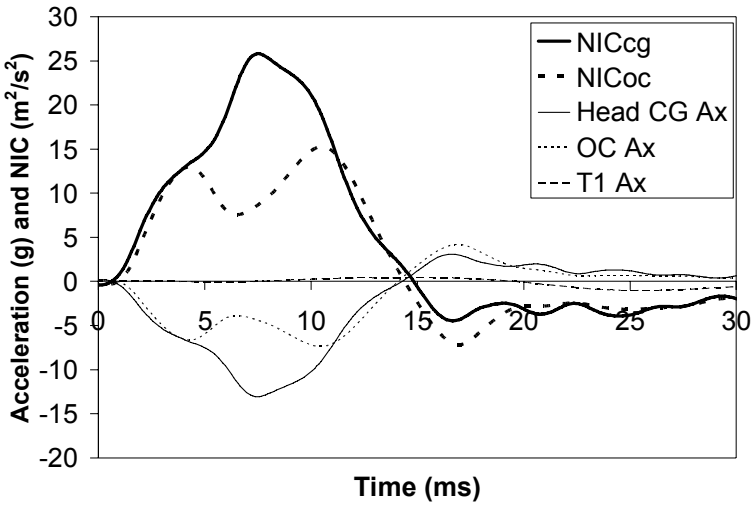

Fig. 13. X-axis accelerations and NIC values for the test shown in Figures 8 and 9.

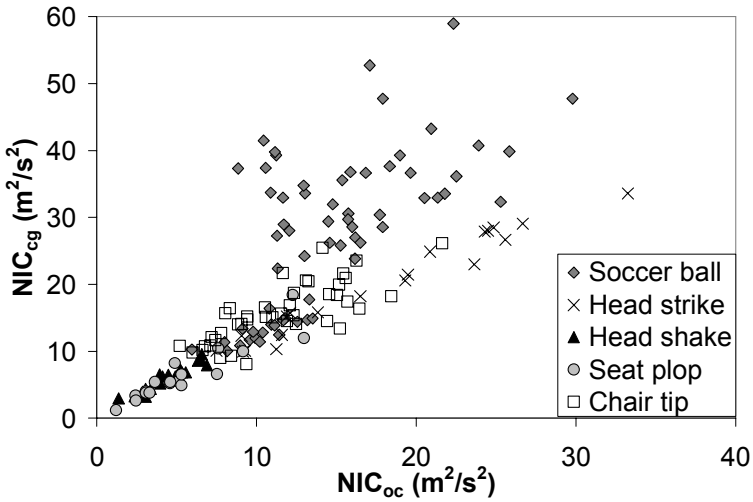

Fig. 14. Comparison NIC values calculated using the relative acceleration and velocity between T1 and the center of gravity of the head (NIC_{cg}) and the occipital condyles (NIC_{oc}).

A high proportion of subjects experienced NIC values exceeding the threshold of 15 m^2/s^2 proposed by Boström et al. (1996) in the soccer ball, voluntary head strike, and chair tip tests. Based on the NIC risk curve proposed by Eriksson and Kullgren (2006), the vast majority of the subjects

should have sustained long-term (> 1 month) AIS 1 neck injury as a result of participating in this study. In fact, no long-lasting symptoms were reported by any of the volunteers. According to the NIC risk curve, the odds of this finding are essentially zero (Table 3). There were no neck complaints at all as a result of the voluntary head strike tests in spite of the fact that highest NIC_{oc} values were recorded in that scenario.

Table 3. Analysis of the risk of long-term (> 1 month) AIS 1 neck injury for each test scenario based on the NIC risk curve proposed by Eriksson and Kullgren (2006).

Scenario	n	# subjects with a NIC > 15 m^2/s^2		Expected # of injured subjects (E_{inj})		Probability that no subject was injured	
		NIC_{cg}	NIC_{oc}	NIC_{cg}	NIC_{oc}	NIC_{cg}	NIC_{oc}
Soccer ball	20	19	13	16.4	5.3	0%	0%
Head strike	20	14	11	7.0	5.6	0%	0%
Head shake	20	0	0	0.1	0.0	89%	100%
Seat plop	18	1	0	0.5	0.3	57%	74%
Chair tip	18	15	8	4.9	2.5	0%	0%

Values for NIC, which is an acceleration-based criterion, were generally not correlated with the values of N_{ij} and N_{km}, which are load-based criteria. Furthermore, the relationship between NIC and N_{km} varied by test scenario (Figure 15). In the soccer ball scenario, very high NIC values were often associated with very low values of N_{ij} and N_{km}. The voluntary head shaking scenario generated some moderately high values of N_{km}, but only very low values for NIC. The chair tip scenario produced some high values for both NIC and N_{km}, but they were not correlated with each other.

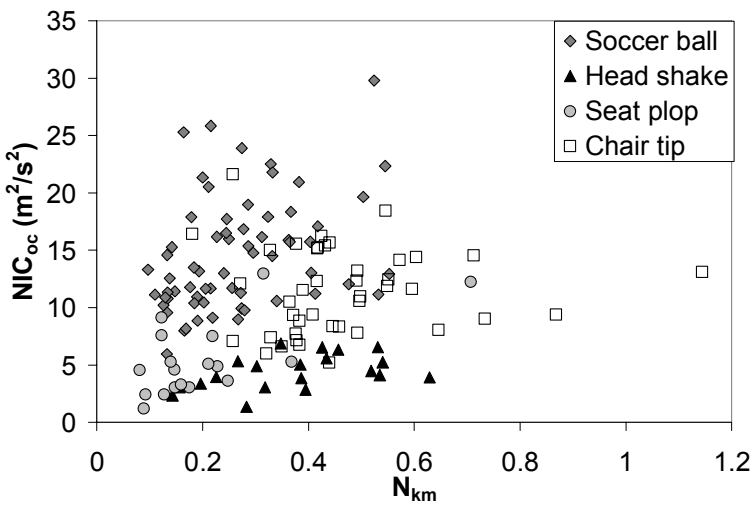

Fig. 15. Relationship between NIC_{oc} and N_{km}.

DISCUSSION

In order to better understand the injury potential associated with automotive rear end collisions, biomechanical measurements were taken from human subjects during vigorous activities in which the neck was forced rearward with respect to the neck similar to an automotive rear impact. Significant methodological difficulties were encountered in attempting to quantify the biomechanical parameters of interest, namely the acceleration at the center of gravity of the head and the loads and moments acting at the occipital condyles. These problems were most pronounced in the soccer ball head impact tests. Raw data from the bite block accelerometers did not provide a good estimate of the head acceleration at the center of gravity in the soccer ball impacts. Our bite block data exhibited a biphasic pulse shape similar to the acceleration traces published by Naunheim et al. (2000) and Shewchenko et al. (2005) in their tests of soccer ball heading. In the present study, the biphasic

acceleration traces from the bite block sensors assumed a more realistic sinusoidal shape when the accelerations were transformed to the center of gravity of the head. This transformation was not possible in our initial soccer ball testing due to significant vibration effects in the angular acceleration data. Fortunately, the vibration problem was solved by using a smaller sensor that was rigidly attached to the bite block. Moreover, a method to calculate the loads and moments at the upper neck using video analysis of the ball motion was developed and validated through Hybrid III dummy testing.

The purpose of the soccer ball scenario was not to characterize the biomechanics of heading in soccer, but to biomechanically evaluate a head impact condition that is known to be non-injurious. The ball speeds utilized in this study are on the low end of the range seen in soccer games and practices. The subjects were able to tolerate over 30 g and nearly 3000 rad/s^2 of angular head acceleration without injury. This level of impact acceleration far exceeds the value of 1800 rad/s^2 proposed as a 50% risk of cerebral concussion by Ommaya and Hirsch (1971). Interestingly, comparable levels of peak linear acceleration were generated in the voluntary head strike scenario. In that scenario, the speed of the hand just prior to striking the head ranged from 2.2 – 4.5 m/s. This hand speed is far less than the 9 m/s punch speed recorded in Olympic boxers, and is likely associated with a much lower effective mass (Walilko et al., 2005). In spite of the comparable levels of peak linear head acceleration, several subjects complained of head symptoms after the soccer ball scenario, whereas no subjects had head complaints from the voluntary head strike tests (although one subject did hurt his hand) (Appendix). It seems likely that the head complaints from the soccer ball tests were related to the higher angular head accelerations.

The present study is the first to quantify values for various neck injury criteria associated with soccer ball heading. It was found that NIC, which is an acceleration-based criterion, predicted a high risk of long-term AIS 1 neck injury in the soccer ball impacts, whereas N_{ij} and N_{km}, which are load-based criteria, predicted a low risk of neck injury. NIC also predicted a high risk of long-term AIS 1 neck injury in the voluntary hand strike tests. However, only one subject complained of short-term neck soreness after the soccer ball scenario, and no subjects had any neck complaints after the voluntary head strike. Furthermore, the soccer ball impacts and voluntary head strikes represent very minor head impacts compared to what athletes in contact sports experience. Numerous non-injurious head impacts of 80 g or greater have been recorded in American football players (Duma et al., 2005) and amateur boxers (Pincemaille et al., 1989). Based on the finding in the present study that NIC values are directly proportional to head accelerations in the case of direct head impacts, it seems plausible that athletes in some contact sports may routinely experience NIC values far exceeding the values observed in this study. In spite of this, long-term AIS 1 neck injuries due to head impacts in sports are rare. It may be hypothesized that athletes involved in contact sports have a higher tolerance to neck injury than the general population. However, the cohort in the present study also sustained high NIC values without injury, and the study population was representative of the general population in terms of the distribution of age, gender, size, athleticism, and even the degree of preexisting degeneration in the cervical spine, which ranged from none to moderate.

From this analysis, it is clear that NIC overestimates the risk of long-term AIS 1 neck injury in short duration head impact events (5 – 15 ms). It may be argued that NIC is designed to predict injury in an automotive rear end collision, which occurs over a longer duration and involves a specific set of loading conditions. However, the chair tip scenario was similar to an automotive rear impact in which the occupant's neck muscles are tensed, and NIC overpredicted the incidence of long-term AIS 1 neck injury in that scenario, as well. From the standpoint of predicting injury using NIC, all of the test scenarios were biomechanically similar to an automotive rear impact because they all created relative rearward acceleration of the head with respect to the base of the neck. Since NIC purports to be biomechanically based, it should be able to predict the presence or absence of long-term AIS 1 injury in situations other than an automotive rear impact. For example, because the Head Injury Criterion (HIC) is biomechanically based, it is used not only to certify vehicles, but also to certify helmets. The finding that NIC is not a robust predictor of injury in non-automotive situations casts doubt on the biomechanical theory that long-term AIS 1 neck injury is the result of nerve damage due to a "water hammer" effect.

In spite of its apparent lack of biomechanical validity, it is not surprising that NIC has been shown to accurately predict the presence or absence of injury in real world automotive collisions (Eriksson

and Kullgren, 2006). However, this predictive capability only represents an empirical association between NIC and injury, not a causal relationship. In the setting of rear end collisions, NIC correlates with collision severity, neck loading, and a number of other parameters that can be used to predict injury. One purpose of this study was to investigate a wide variety of loading scenarios that decoupled head acceleration from neck loading so that the importance of each parameter could be isolated by controlling for the effects of the other. The scenarios chosen for this study successfully removed the correlation between NIC and N_{km} seen in automotive rear impact studies (Figure 15). Based on this analysis, it was determined that NIC, which is an acceleration-based criterion, did not perform well as an injury predictor independent of neck load. It cannot be stated that a load-based criterion such as N_{ij} or N_{km} will accurately predict neck injury based on the data from the present study. However, these criteria met the necessary requirement of accurately predicting the absence of injury. N_{ij} and N_{km} were highly correlated and did not appear to offer unique information independent of each other in the scenarios tested. The other purpose of this study was to document biomechanical parameters associated with familiar activities. In that respect, this study adds to the data set generated by Allen et al. (1994) and Vijayakumar et al. (2006).

CONCLUSIONS

The purpose of this study was to evaluate the robustness of various whiplash injury criteria in loading scenarios that are biomechanically similar, but not identical to, an automotive rear end collision. The effect of relative acceleration between the head and neck was isolated by choosing loading scenarios that resulted in relatively little displacement or loading of the neck. Low values for load-based injury criteria such as N_{ij} and N_{km} were recorded in nearly all tests. However, high values (> 15 m^2/s^2) for the Neck Injury Criterion (NIC) were recorded in numerous volunteers subjected to minor forehead impacts from a soccer ball or their own hand. Several volunteers who were dropped vertically while seated supine in a chair also experienced high NIC values (> 15 m^2/s^2) without any head impact. According to the NIC risk curve proposed by Eriksson and Kullgren (2006), long-term (> 1 month) AIS 1 neck injury would have been expected in several people and virtually inevitable in at least one person as a result of those exposures. The fact that none of the volunteer subjects experienced any significant neck symptoms suggests that NIC is not a robust injury criterion and casts doubt on the underlying biomechanical theory that long-term whiplash injury is caused by nerve damage due to relative A-P acceleration between the head and neck base.

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Appendix. Symptoms resulting from testing.

Scenario	Age	Sex	Symptoms	Length
Soccer ball	45	M	Slight ringing in right ear	2 hours
	34	M	Minor superficial stinging and was momentarily dazed after higher speed tests	seconds
	57	M	Headache	30 minutes
	45	M	Headache	15-30 minutes
	44	F	Neck tightness and slight pain	< 1 day
	28	M	Forehead hurt after 11.5 m/s test	seconds
	54	F	Minor headache	< 1 day
Head strike	54	F	Jaw discomfort from poorly fitting bite block	< 1 day
	45	M	Heel of hand stung	minutes
Head shake	48	F	Neck discomfort	< 1 day
	48	M	Neck soreness	< 1 day
	34	M	Neck soreness	< 1 day
	27	M	Neck pain/neck stiffness	2 days/4 days
	57	M	Neck soreness	< 1 day
	55	M	Neck soreness	< 1 day
Chair tip	48	F	Right lower back soreness	< 1 day
	28	M	Neck stiffness	< 6 hours
	50	F	Slight headache	< 2 hours
	45	F	Sharp back pain	< 1 hour
	34	M	Minor neck soreness the next AM	< 1 day