Assessing the Neck Injury Index (NII) Using Experimental Cadaver Tests

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ABSTRACT

The neck injury index, NII, was developed in ISO 13232-5 (2005) as a testing and evaluation procedure for assessing the risk of injury to the AO/C1/C2 region of the cervical spine in motorcycle riders. A recent series of 36 head/neck component tests was used to examine the risk of neck injury in frontal impacts and to assess the predictive capability of NII for impacts of various orientations. Neck injuries produced in the testing ranged from AIS 1 to AIS 5. Using force and moment load cell cadaver experimental data, injury risk was assessed using NII evaluated with the ISO 13232-5 algorithms. The injury risk predictions are compared with the injury outcomes from the head/neck cadaver. The average predicted risk of injuries for the experimental injury tests based on NII at the corresponding MAIS level observed in testing was 0.7% though there were 11 AIS 3+ injuries in the testing. Using the experimental injury outcomes and the experimental force and moment time histories, the normalizing coefficients from NII are reevaluated to minimize the difference between NII risk assessment and the experimental injury outcome in the L_2 basis. This reanalysis is compared with existing human and cadaver neck injury criteria.

Limitations of this study are primarily the limitations of the source data including lack of dynamic muscle response, anthropometric and age differences between experimental and epidemiological populations, and limitations on the principal direction of impact, generally flexion/compression and flexion/tension. Further, the injuries seen in the source data include both upper neck (OC-C2) and lower neck (C3-T1) injuries; the NII injury criterion is intended to assess the potential for upper neck injuries.

Keywords: NECK, FRONTAL IMPACTS, INJURY CRITERIA, STANDARDS

The neck injury index, NII, was developed in ISO 13232-5 (ISO 13232, 2005; Van Auken, 2005) as the neck injury evaluation portion of a comprehensive testing and evaluation procedure for assessing the risk of injury in motorcycle riders. NII predicts the risk and severity of neck injuries based on computational and experimental crash tests using the MATD dummy neck (Withnall, 2003; Van Auken, 2001) and two epidemiological databases of motorcycle crashes. In this study, a recent series of cadaver tests (Bass, 2006) is used to examine the risk of neck injury in frontal impacts and to assess the predictive capability of NII for impacts of various orientations.

The neck injury assessment in ISO 13232-5 was based on datasets from two epidemiological databases: the combined Los Angeles (LA)/Hanover database (injury data in ISO 13232-2:2005 Annex C), and a study by Thom et al. (1995). The LA/Hanover database included 498 motorcycle crashes with a closing velocity less than 121 km/hr (13 fatalities). There were, however, only 15 AIS 2+ neck injury cases in the nonfatal dataset of which 12 were AIS 3+. Without detailed postmortem neck necropsies, it was recognized that additional neck injuries may have occurred in the fatalities (ISO 13232, 2005). Further, with the limited number of AIS 1 neck injuries (4) and AIS 2 neck injuries (3) in the dataset, there was potential for occult neck injuries among the survivors. Kebschull et al. (1998) used a computational model to simulate the epidemiological database and found that approximately 30% of the impacts in which one or more of the NII values exceeded 1 ("interpreted as likely neck fracture or dislocation; with significant likelihood of spinal cord damage"). However, only 2 percent of nonfatal cases had neck ruptures, dislocation or fracture. The disparity between the large number of calculated injuries and the limited number of injuries identified in the field was the impetus for the first NII injury risk assessment from a previous version in ISO 13232.

The second epidemiological database, the study by Thom (1995), investigated 304 fatally injured motorcycle riders in the Los Angeles area. A subset of 71 fatal cases with a motorcycle to vehicle closing velocity of 121 km/hr was further analyzed by Smith (2000); 77.5% of the riders did not use a helmet. Smith found that 92.5% of fatalities had non-fatal neck injuries and 7.5% had fatal neck injuries. A compression injury mechanism was attributed to 25% of the neck injury cases, a flexion injury mechanism was attributed to 34% of cases, and a tensile injury mechanism was attributed to 14% of cases. The neck injuries were mostly AIS=3 cases (69%). An extensive series of computational simulations was used to provide dynamic neck forces and moments in multiple axes. Calibrated simulations using NII as an injury assessment tool matched the number of injuries in the LA/Hanover database.

ISO-13232 strictly focuses only on OC-C2 neck injuries for motorcycle riders. Further, an underlying assumption of the use of the epidemiological databases is that helmet use does not affect severity of neck injury (cf. Hurt, 1981). However, under inertial situations with helmeted occupants, the proportion of lower cervical spinal injuries (C5-T1) may increase (cf. Bass, 2006, McEntire, 1997). Such loading may occur with primary thoracic contact in motorcycle riders.

Kasantikul (2002) studied 955 motorcycle crashes with 1082 riders and 399 passengers in Thailand. Detailed dissection based on the technique of Rehman and Noguchi (1983) was performed in 73 fatalities of helmeted and unhelmeted passengers and drivers in 67 crashes. In Figure 1, the injury distribution for structural cervical spinal injuries from Kasantikul et al. and Smith are compared. Anterior injury includes fracture at all cervical spinal levels and disc fracture and separation; posterior injury types include atlanto-occipital separations (attributed to C0 in the figure) and atlanto-axial separations (attributed to C1 in the figure). The two datasets exhibit different patterns of cervical spinal injury. Kasantikul et al. showed similar numbers of injuries for the upper and lower cervical spine while the injuries seen by Smith were predominantly in the upper cervical spine; in the study by Smith, the upper neck (OC-C2) had 3.5 times the number injuries as the lower neck (C3-T1). Though the differences between the two assessments may be attributed to numerous factors, including differences in force and moment magnitudes and directions resulting from different vehicle sizes, speeds, impact configurations, and helmet use, the results from Kasantikul et al. emphasize the potential for injuries to lower cervical spinal structures from motorcycle crashes.

a) b)

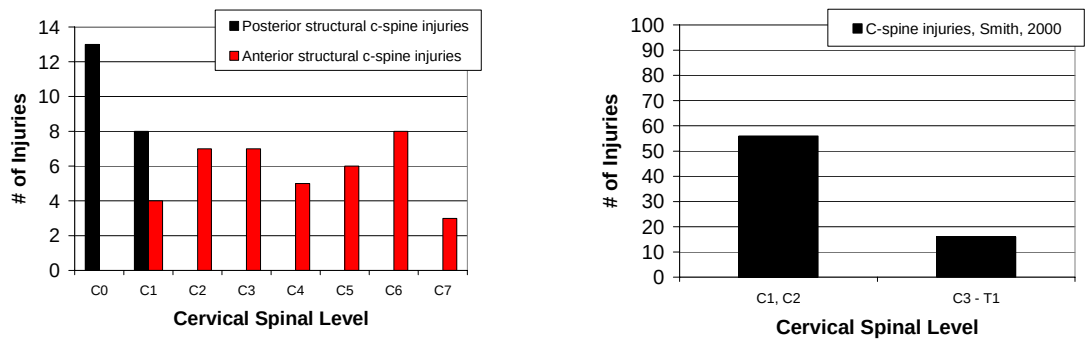

Figure 1. Cervical Spinal Level Injury Data, a) Kasantikul, 2002, b) Smith, 2000

Based on a generalized stress ratio for the estimation of the strength of materials, NII is formulated as

$$NII = \max\left(\left(\left(\frac{F_C}{F_{CC}} + \frac{F_T}{F_{TC}} + \left(\left(\frac{M_X}{M_{XC}}\right)^2 + \left(\frac{M_{Ext}}{M_{EC}} + \frac{M_{Flx}}{M_{FC}}\right)^2\right)^{1/2}\right)^2 + \left(\frac{M_Z}{M_{ZC}}\right)^2\right)^{1/2}, 3.1\left(\frac{F_C}{F_{CC}} + \frac{F_T}{F_{TC}}\right)\right) \quad (1)$$

where F_C is the neck axial compression force, F_T is the axial tension value, M_{ext} is the extension moment, M_{flx} is the flexion moment, M_X is the lateral bending moment, M_Z is the torsion moment (ISO-13232). Values with the last subscript *C (e.g.* F_{CC} and M_{CC}) represent critical values that define the dummy injury predictor thresholds for isolated modes of loading (Table 2). The critical values are strictly valid only for the MATD dummy, with the dummy and the injury criterion forming a 'matched system'; however, the injury values are based on computational simulations of the human

epidemiological databases. So, the NII formulation should also be valid for human datasets and be approximately valid for cadaveric dataset. A crucial assumption used to derive the limiting axial force assessment as the right-side function under the maximum is that the probability of an AIS 3+ injury is 0.03 with a 4.17 kN tension only force (ISO 13232, 2005). In subsequent discussion, the left function is termed NII-left and the right function is termed NII-right.

NII injury assessment is based on the function

$$P_{AIS} = 1 - \exp\left(-\left(\frac{NII - a}{b} \right)^{3.5} \right) \tag{2}$$

with limits on NII for each AIS level (the probability of injury is assumed to be zero if NII is less than these minimum NII values). As shown in Table 1, the *b* coefficient is the same for each severity level, and the *a* coefficient is the same for AIS=4 and AIS=5 injuries since there were no AIS=4 injuries in the dataset. NII values less than 1.06 have a low risk of AIS = 1 injury.

Table 1. NII Injury Assessment Values and Coefficients

Severity Level	**Min NII**	***a***	***b***
AIS≥1	1.06	1.06 (0.39, 3.02)[a)]	4.38
AIS≥2	1.86	1.86 (0.52, 3.48)	4.38
AIS≥3	2.29	2.29 (0.76, 3.90)	4.38
AIS≥4	4.73	4.73 (2.68, -)	4.38
AIS≥5	4.73	4.73 (2.68, -)	4.38
AIS≥6	6.13	6.13 (3.65, -)	4.38

[a)] 95% confidence interval, based on 95% confidence interval for μ in ISO 13232-5 Table J.8 and assuming b=4.38.

Bass et al (2006) developed an injury criterion, termed Beam, using cadaveric component head/neck complexes. This criterion is based on a survival analysis with an assumed underlying logistic distribution for dynamic variables at the lower neck and a criterion for the failure stress in a simple beam as:

$$Beam = \frac{F_Z}{F_{ZC}} + \frac{M_Y}{M_{YC}} \tag{3}$$

where F_Z is either the tension (F_T) or compression (F_C) axial load in the lower neck, F_{ZC} is the critical axial load, M_Y is the moment in the sagittal plane in either flexion (M_{flx}) or extension (M_{ext}), and M_{YC} is the critical moment. Critical forces and moments for Beam are shown in Table 2. The injury spectrum seen in the Bass et al. experimental data was imposed on the datasets by the range of conditions selected for the study, largely compression-flexion and tension-flexion conditions. Many of the injuries were posterior fractures and ligamentous separations. Of the AIS 3+ injuries, two occurred in the upper cervical spine (C1, C2), and ten occurred in the lower cervical spine (C3-T1). Bass et al. also found that anterioposterior shear had limited effect for the experimental conditions investigated.

Kleinberger et al. (1998) developed an injury criterion for the upper neck of the Hybrid III dummy that combines the effects of axial loading and bending moments as

$$Nij = \frac{F_Z}{F_{ZC}} + \frac{M_Y}{M_{YC}} \tag{4}$$

When little or no moment is present, additional peak tension and compression limits serve as the critical values producing a 'kite'-shaped injury assessment boundary (Eppinger et al., 2000). Values of these critical values are shown in Table 2. These values provide a demarcation for which Nij greater than 1.0 corresponds to a 22% risk of AIS 3+ neck injury. The functional form of Beam and the Nij criteria are similar; however, beam uses moments and forces measured in the lower neck and Nij uses moments and forces measured in the upper neck. The functional form of NII, however, is only similar to Beam and Nij for pure tension or compression forces and/or pure flexion or extension moments. The critical values for compression are similar among the three neck injury assessment techniques as shown in Table 2. Critical values for tensile force are similar between Beam and Nij; however, the limited biofidelity of the Hybrid III in bending for large shearing motion with head supported mass (c.f. Bass et al (2005)) lowers the critical value for Beam to less than half the value

for Nij. For NII, the critical value in axial tension is substantially lower than either Beam or Nij. The critical value for flexion is larger in Nij than in Beam or NII, presumably the result of differences in response between the human/MATD dummy and the Hybrid III dummy neck. Finally, the critical value in extension for NII is less than half that for Nij, again presumably the result of the relatively stiff Hybrid III neck (cf. Baughn, 1993, McGill, 1994).

Table 2. Critical Values for Neck Injury Criteria (* Nij axial limits altered for small moment values)

Analysis	Author	F_C (N)	F_T (N)	M_{flx} (Nm)	M_{ext} (Nm)	M_x (Nm)	M_z (Nm)
NII	ISO-13232 (2005)	-6530	3340	204.2	-58	62.66	47.1
Beam	Bass et al (2006)	-5430	5660	141	NA	NA	NA
Nij	Kleinberger et al (1998), FMVSS-208	-6160*	6806*	310	-135	NA	NA

The objective of this study is to evaluate the NII neck injury predictions using an existing set of cadaveric impact tests to assess the correlation between the predicted injuries and injuries seen in the experimental dataset. Further, using the injury data from the existing dataset, new coefficients will be derived within the framework of the existing NII methodology.

METHODOLOGY

A recent series of tests was performed by the University of Virginia with added head-supported mass (helmet) for 36 head/neck component cadaver tests (Bass, 2006). The sled tests were instrumented with accelerometers and angular rate sensors, while the component tests added load cells at the distal thoracic spinal mount. Injuries produced in the testing ranged from AIS 1 to AIS 5. The experimental test matrix is shown in

Table 3. The neck angle is relative to the T1/T2 spinal unit, and the center of added mass was located in the midsagittal plane with an posterioanterior X direction and a inferior-superior Z direction.

Cadaveric specimens were procured in accordance with state and federal regulations and are subject to the oversight of the University of Virginia Cadaver Use Committee. All cadaveric specimens were fresh frozen with no evidence of wasting disease, Hepatitis B, Hepatitis C or HIV. The bone quality of each specimen was assessed using a histogram technique (QBMAP, The IRIS) on pretest CT scans. No specimen had evidence of osteoporosis (T-Score > 2.5 for UCSF 25 year old dataset), preexisting fractures, or significant cervical spinal disease that might compromise skeletal strength. The average mass and height of the specimens used for the component tests was 79±19 kg and 1761±79 mm, respectively. These averages compared well with the 50th percentile male mass and height of 79 kg and 1780 mm, respectively. The average age of the specimens in all component tests was 59 years with no cadavers older than 74 years.

Using force and moment load cell cadaver data from the UVa tests, along with corresponding Hybrid-III and THOR response data, NII injury risk assessments were evaluated using the ISO 13232-5 algorithms. The injury risk predictions are compared with the injury outcomes from the component and whole body cadaver datasets. Using these injury outcomes and the experimental force and moment time histories, the normalizing coefficients from NII are reevaluated to minimize the difference between NII risk assessment and the experimental injury outcome in the L_2 basis (the sum mean square difference between the two curves). This reanalysis is compared with existing human and cadaver neck injury criteria.

A survival analysis (Hosmer, 2003) was performed using Minitab version 14 (Minitab, Inc, State College, PA) for injuries of maximum MAIS (maximum AIS) ≥ 2 (AAAM, 1998). A parametric analysis was performed with arbitrarily censored data using a Gaussian and a logistic curve (right censored for non-injury tests and left censored for injury tests) for the injury assessment in equation (2). Maximum likelihood estimates were used in the calculation of the survival function.

Table 3. Cadaver component test matrix.

Test	Added Mass (kg)	Mass Position (Relative to Head cg)		Neck Angle (deg)	Sled Velocity (m/s)
		X (cm)	Z (cm)		
HM2_cad1	2.0	0	0	0	3.9
HM2_cad2	2.0	0	0	0	5.4
HM2_cad3	2.0	0	0	0	5.4
HM2_cad4	2.0	0	0	0	4.0
HM2_cad5	2.0	0	0	0	4.6
HM2_cad6	2.0	0	0	0	4.0
HM2_cad7	2.0	0	0	0	3.9
HM2_cad8	2.0	0	0	0	4.9
HM2_cad9	2.0	0	0	0	4.5
HM2_cad10	2.0	0	0	30	2.9
HM2_cad11	2.0	0	0	30	3.7
HM2_cad12	2.0	0	0	30	3.7
HM2_cad13	2.0	0	0	30	3.1
HM2_cad14	2.0	0	0	30	3.7
HM2_cad15	2.0	0	0	30	3.4
HM2_cad16	2.0	0	0	45	1.7
HM2_cad17	2.0	0	0	15	4.3
HM2_cad18	2.0	0	73	-45	3.7
HM2_cad19	2.0	0	73	-45	3.2
HM2_cad20	2.0	0	73	-45	3.6
HM2_cad21	2.0	0	73	-45	3.8
HM2_cad22	2.0	0	55	-45	3.5
HM2_cad23	2.0	0	32	-45	3.6
HM2_cad24	2.0	0	32	-45	3.5
HM2_cad25	2.0	0	0	-45	3.5
HM2_cad26	2.0	0	67	-45	3.6
HM2_cad27	2.0	0	0	-45	3.3
HM2_cad28	2.0	0	55	-45	3.3
HM2_cad29	2.0	0	55	0	3.2
HM2_cad30	2.0	0	73	0	3.2
HM2_cad31	3.0	0	98	0	3.1
HM2_cad32	3.0	0	98	30	3.2
HM2_cad33	3.0	0	98	30	3.4
HM2_cad34	4.0	0	118	30	3.4
HM2_cad35	4.0	0	118	30	3.6
HM2_cad36	2.0	0	0	30	3.3

RESULTS

By applying NII to the cadaveric response data (Table 4), the risk of injury predicted by NII appears to substantially underestimate the risk and severity of injuries identified in both the component and the full cadaver head-supported mass tests. The average predicted risk of injuries based on NII at the corresponding MAIS level observed in testing was 0.7%. The 36 component tests included 11 MAIS ≥ 3 injuries and 20 MAIS ≥ 1 injuries. No correlation was found between the AIS level of injury and the NII value ($R^2 = 0.1$). For example, in one test where the component cadaver received MAIS=5 injuries with large axial force and anterioposterior moment values, the calculated maximum NII of 3.1 corresponds to a low risk of an AIS 1 injury. The AIS 1+ injury risk vs. NII is shown in Figure 2. No NII value for the Bass et al. dataset is greater than 4.4, a value at which NII would predict less than 50% risk of AIS 1+ injury. So many experiments in which AIS 3+ are obtained have a low risk of AIS 1+ injury using NII. Indeed, the L_2 norm error for AIS 3+ using the NII assessment is 10.8, an average of almost one for each AIS 3+ injury in the Bass dataset.

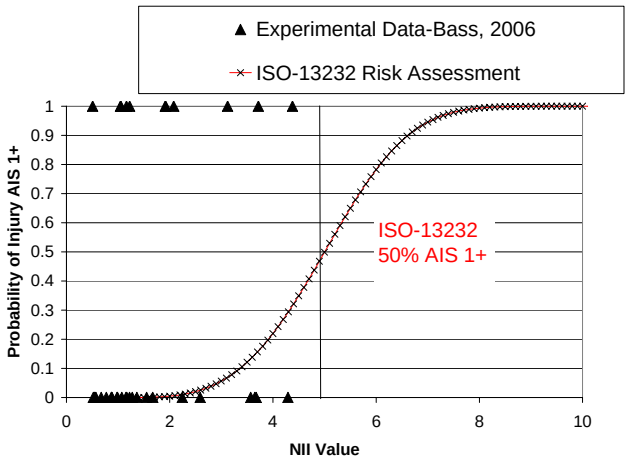

Figure 2. AIS 1+ Injury Assessment Curve and Values from Neck Injury Tests of Bass et al NII (ISO-13232)

Table 4. Component Cadaver Injury, NII, Beam and NIJ results

Test	MAIS	Level of MAIS Spinal Injury	NII	Beam (Optimized)	NIJ
Hm2_cad1	2	C4/C5	1.07	0.60	0.48
Hm2_cad2	5	C2/C3	1.22	1.02	0.65
Hm2_cad3	5	C7/T1	1.92	1.09	0.85
Hm2_cad4	3	C5/C6	1.04	1.03	0.71
Hm2_cad5	4	C7/T1	2.08	0.99	0.78
Hm2_cad6	0	NA	1.22	1.16	0.71
Hm2_cad7	0	NA	1.36	0.85	0.62
Hm2_cad8	4	C7/T1	1.91	0.81	0.30
Hm2_cad9	5	C5/C6	1.06	0.83	0.70
Hm2_cad10	0	NA	2.24	0.76	1.16
Hm2_cad11	5	C6/C7	3.12	1.14	1.12
Hm2_cad12	4	C5/C6	3.72	0.95	0.98
Hm2_cad13	0	NA	3.57	1.09	0.97
Hm2_cad14	3	C6/C7	4.38	1.20	1.09
Hm2_cad15	2	C5/C6	3.65	1.01	0.96
Hm2_cad16	0	NA	4.29	1.05	1.27
Hm2_cad17	0	NA	3.68	1.16	1.07
Hm2_cad18	3	C3/C4	1.16	1.06	0.37
Hm2_cad19	2	C4/C5	0.67	0.82	0.22
Hm2_cad20	0	NA	0.77	0.66	0.24
Hm2_cad21	1	C4/C5	1.15	0.79	0.27
Hm2_cad22	2	C3/C4	0.88	0.92	0.28
Hm2_cad23	1	C5/C6	0.87	0.68	0.29
Hm2_cad24	0	NA	0.88	0.72	0.30
Hm2_cad25	1	C2/C3	0.98	0.88	0.32
Hm2_cad26	1	C4/C5	1.17	0.84	0.50
Hm2_cad27	0	NA	0.57	1.10	0.15
Hm2_cad28	0	NA	0.55	0.56	0.18
Hm2_cad29	3	C1	0.51	0.61	0.23
Hm2_cad30	0	NA	0.54	0.42	0.57
Hm2_cad31	2	C5/C6	0.52	0.75	0.18
Hm2_cad32	0	NA	0.99	0.26	0.20
Hm2_cad33	0	NA	1.67	0.34	0.59
Hm2_cad34	0	NA	1.27	0.61	0.62
Hm2_cad35	0	NA	1.55	0.73	0.53
Hm2_cad36	0	NA	2.59	0.59	0.94

The average NII, Nij and Beam value for the experimental AIS level is shown in Figure 3. NII values are calculated using Equation 1 and Equation 2, and using the NII-left and NII-right function. Both Beam and Nij have generally greater values as the AIS level of the specimens in the experiments increases, but NII shows smaller values for AIS=5 experiments relative to the AIS=4 experiments.

The coefficient of variation (the standard deviation of the combined AIS 2+ assessments divided by the mean for each injury assessment method), a measure of how well clustered the analysis values are for a given AIS level, are shown in Figure 4. There is a large variance in NII predictions for experimental dataset while Beam gives the lowest normalized standard deviation value, less than 20% of the mean. NII-right has a large standard deviation level that is over 80% of the mean value for AIS 2+. Owing to this variance in the right function calculations, NII shows similar standard deviation levels. The NII-left function had the lowest normalized standard deviation value. However there was a large standard deviation for the average normalized standard deviation indicating large spread between AIS levels.

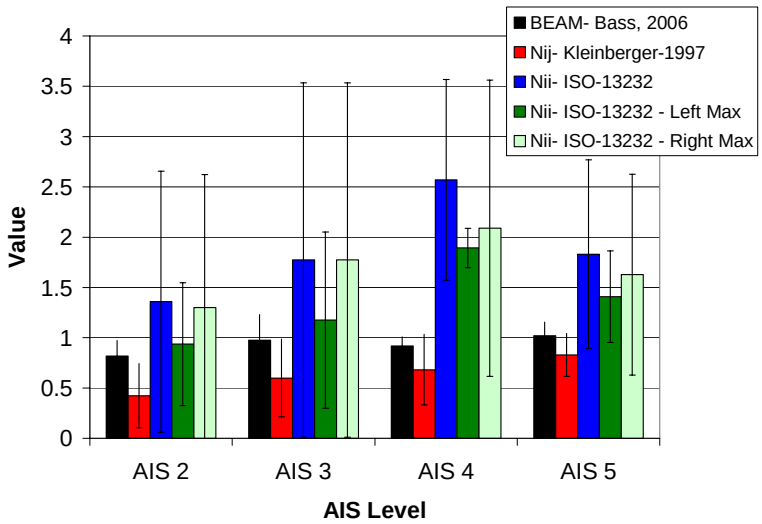

Figure 3. AIS Values from Neck Injury Tests of Bass et al Evaluated with BEAM, Nij, and NII (ISO-13232)

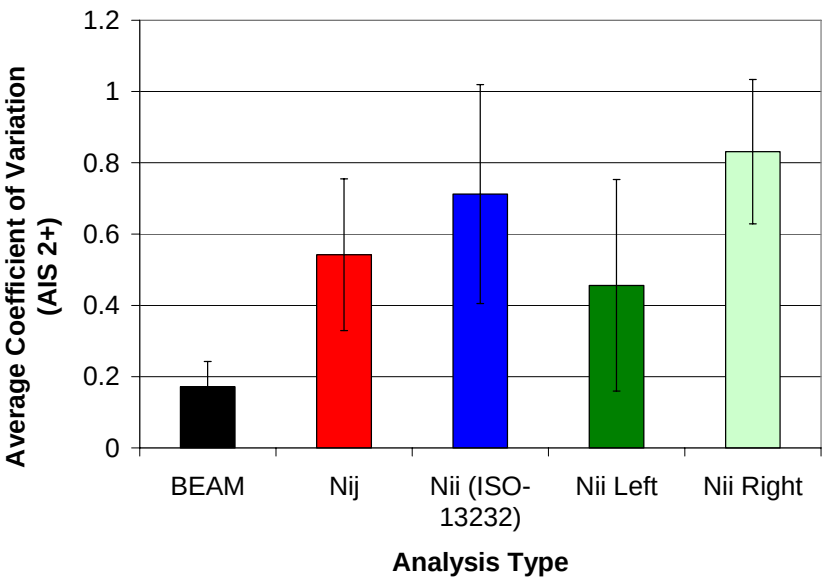

Figure 4. Coefficient of Variation for AIS 2+ Values from Neck Injury Tests of Bass et al. Evaluated with BEAM, Nij, and NII (ISO-13232)

A survival analysis for AIS 3+ injuries was performed using NII with the same functional form as in Equation 1 and Equation 2 and with the same critical values from Equation 1, but with different coefficients for the approximate normal distribution in Equation 2. Since the number of AIS 4 and AIS 5 injuries was limited in the experimental dataset, only the AIS 3+ values were calculated. The NII right side coefficient was reevaluated using the outcome of the AIS 3+ assessment of NII for pure tensile loading assuming 50% risk of injury is 6800 N (ISO-13232.5 Annex J). The value for this coefficient was determined to be 1.77 rather than 3.1 that was used in the original formulation.

Using a normal distribution, the result of the survival analysis is shown in Figure 5. The *a* and *b* coefficients for this analysis (using Equation 2) are shown in Table 5. The Anderson Darling statistic is 35.4 for the fit and the L_2 norm error of the predicted dataset is 4.0 for the AIS 3+ injuries, 40% of the value seen with the original NII analysis. This suggests the new survival curve is an improvement on the ISO-13232 curve for the experimental data used in this study. The 50% risk of injury is within the range of the experimental data while the ISO 50% risk is nearly a factor of three higher than any NII value that occurs using the existing experimental dataset.

Table 5. Coefficients for Equation 2

Analysis	**a**	**b**
NII	2.29 (0.76, 3.90)	4.38
This Study	-0.87	3.03

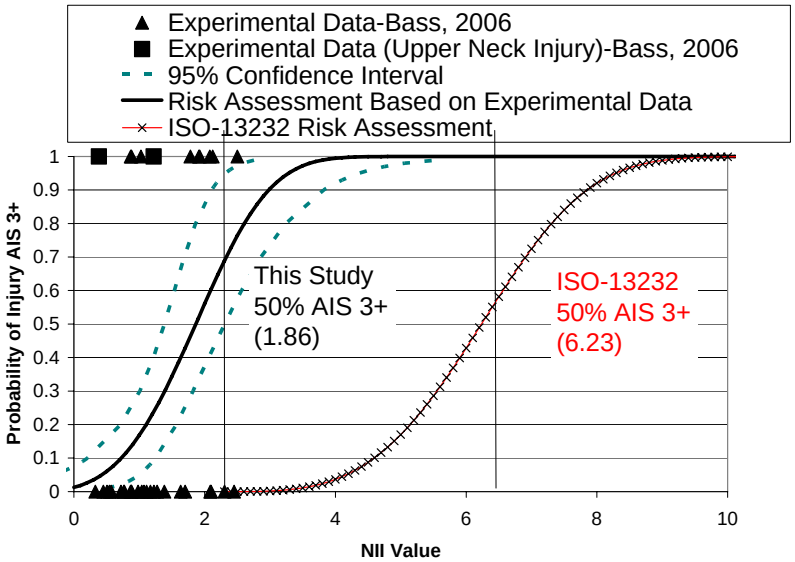

Figure 5. Survival Analysis for AIS 3+ Values from Neck Injury Tests of Bass et al Evaluated with NII (ISO-13232) and This Study with Confidence Interval (NII values for the 50% AIS 3+ injuries are indicated in parentheses)

DISCUSSION

Much of the injury data on which NII is based is AIS=3, and limited numbers of AIS=2 and AIS=1 injuries were assessed in the epidemiological studies suggesting censoring of AIS=1 and AIS=2 data. A similar distribution of injuries is seen in Smith (2000) in which the majority of injuries are AIS=3. This may affect injury assessment values and resulting risk assessments for AIS 1+ and AIS 2+ injuries.

The $NII_{cadaver}$ assessment developed in this study, if applied to MATD dummy simulations of real-world motorcycle crashes would likely substantially overpredict injuries seen in the epidemiological field data (cf. Van Auken, 2005). Assuming that the cadaveric dataset is approximately representative of the human response, the disparity between the implied mechanical forces and moments at injury for the dummy and cadaveric dataset imply limitations in the biofidelity of the MATD dummy for the loading applied in the experimental cadaver dataset. A similar situation has been seen in pediatric neck loading assessed using dummy surrogates in which critical values determined in reconstructions

are substantially higher than those derived from experimentation with biological surrogates (Sherwood, 2002).

INJURY LOCATION AND MECHANISMS: NII strictly predicts upper neck injury, though there are numerous lower neck injuries reported in the epidemiological literature. The experimental dataset used in this study has limited numbers of upper neck injuries, however, failure tolerances in the upper and lower spinal components are similar (e.g. Bass, 2007). There is evidence, however, that muscle tensing in humans tends to produce cervical spinal injuries at more superior cervical spinal levels relative to no muscle tensing for similar dynamic impacts (Nightingale, 1998).

CRITICAL VALUES: Single-axis dynamic variable values for 50% risk of AIS 3+ injury are shown in Table 6. The Hybrid III neck, which was developed for seated motor vehicle occupants, has poor biofidelity in bending, and the single-axis values have been scaled to human values using results of Eppinger et al. (Eppinger, 2000) based on air bag loading simulations. The ratio of Hybrid III extension/moment to human extension/moment is taken to be 2.4, and the Hybrid III extension values are multiplied by 0.7 to scale to human values (Eppinger, 2000). Tension and compression are scaled with the same values as flexion and extension.

For NII, the value for 50% risk of AIS 3+ injury for compression is quite large at over 13 kN, the value for 50% risk of injury in tension is 6717 N, and the value for 50% risk in flexion is over 1200 Nm. Both the compressive and flexion AIS 3+ 50% risk values appear to be much larger than values reported in the literature or sustainable in dummy or cadaveric surrogates. Cheng et al (1982) found resultant neck force (axial and shear) of 6200 N for injury; this is likely an upper limit on the force required for upper neck/basilar skull fractures. Mertz and Patrick (1971) found an axial compression limit using reconstructions of 6.75 kN using a stationary surrogate. Pintar and Yoganandan (Pintar, 1998) performed dynamic head/neck compression tests on cadavers with velocities of impact from 0.25 cm/s to 800 cm/s. They measured forces and accelerations on the specimens and correlated these with the injuries of the specimens. They found that the pure compressive tolerance level of the cadaveric specimens ranged from 2 kN to 7 kN. A regression analysis of the data suggests a compressive tolerance level of ~4500 N for a mid size 30-35 year old male under dynamic loading conditions. Regarding material properties, the NII compression tolerance appears to be large. Assuming a column load with a simply supported beam and elastic bucking failure, with trabecular bone stiffness 67 MPa (cf. Duck, 1990), and the neck length of a 50% male (119 mm), the critical buckling load implies a cylinder of bone of approximately 21 cm diameter. The 50% risk of AIS 3+ compression value of 6860 N determined in this study using the NII functional form is comparable to the Beam and Nij values and single-axis experimental values.

For flexion and extension, the NII 50% risk of AIS 3+ injury of 1272 N in flexion and 362 N in extension are also large compared with values obtained from the literature. Mertz and Patrick (1971) performed 90 static and 178 dynamic sled tests on volunteers and cadavers to determine appropriate flexion and extension injury threshold moment values. The maximum dynamic moment that the volunteer endured was 88 Nm, which caused a sharp pain that lasted for several days. When similar tests were run using cadavers under flexion, a value of 190 Nm was determined to be the maximum moment without ligamentous or bony damage, but with possible muscle damage. During flexion tests, no cadavers were found to have ligamentous or bone damage. For cadaveric extension tests, a 47 Nm torque about the occipital condyles was non-injurious, while a 57 Nm torque caused ligamentous damage. The flexion and extension critical values were scaled to a 50th percentile adult male, as 190 Nm and 57 Nm, respectively.

The NII value of over 1200 Nm for the AIS 3+ 50% risk of injury is far larger than the experimental values seen in the Bass dataset, and appears to be large on a material property basis. Assuming the ultimate strength of long bone as 130 MPa (cf. Duck, 2000) and using a simple beam theory analysis, failure would require a 4.5 cm diameter bone with a circular cross section. As many of the injuries seen in the epidemiological and experimental databases occur in the ligamentous and soft tissue, the moment tolerance value is this analysis likely underestimates the required simple failure cross section. The 50% risk of AIS 3+ flexion value of 380 Nm found in this study using the NII functional form ($NII_{cadaver}$) is larger than the Beam and Nij values and single-axis experimental values. Extension moment does not generally cause failure in the Bass dataset, but the extension value of 108 Nm is similar to the Nij value scaled to the human value.

Though this study's new $NII_{cadaver}$ assessment of the tensile loading tolerance values for 50% risk of AIS 3+ injury of 3510 N is generally smaller than those for NII and Nij scaled to the human value, it is in the range of the human experimental results.

Table 6. Single-Axis Dynamic Variable Value for 50% Risk of AIS3+ Injury, NII, Nij and Beam

Analysis	Surrogate	Value	F_C (N)	F_T (N)	M_F (Nm)	M_E (Nm)	M_x (Nm)	M_z (Nm)
NII (Upper neck) (ISO-13232, 2005)	MATD Dummy	6.235	-13,132	6717	1272	-362	390	294
$NII_{cadaver}$ (Upper and lower neck) (Current Study)	Cadaver	1.77	-6860	3510	380	-108	116	88
Nij (Upper neck) (Kleinberger, 1998)	Hybrid III Dummy	1.639	-7910	8740	211	-92	NA	NA
Beam (Upper and lower neck) (Bass, 2005)	Cadaver	1.093	-5935	6186	154	NA	NA	NA

FORM OF THE NII FUNCTION: The right side of Equation 1 has the effect of lowering the necessary dynamic peak in compression and tension to reach a given NII level. Without the right hand side, the value of compression for a 50% risk of AIS 3+ injury would be over 40 kN. In the ISO-13232 development, the coefficient multiplying this portion of the NII assessment is obtained by assuming that a 3% risk of AIS 3+ tensile-only injury occurs at 4170 N (ISO-13232.5 Annex J). There are, however, risks in using a single value from the tail of a distribution to fit a value for the entire distribution. For this assessment, the ISO-13232.5 Annex J 50% risk of AIS 3+ injuries for pure tensile loading of 6800 N was used to set the assessment coefficient on the NII-right function in Equation 1. As this is the mean value of the distribution, this will presumably increase the fidelity of the fit in the range of 50% AIS 3+ injury risk.

ANTHROPOMETRY: The detailed anthropometry of the epidemiological datasets included in ISO-13232 is unknown, but a recent anthropometric characterization of the US motorcycle population finds that riders are 91% male with an average age of 39 years (MIC, 2000). The anthropometry of the predominantly male Bass experimental dataset is approximately 50th percentile male, likely smaller than the average rider population, and the average age is over 60 years.

CONCLUSIONS

Using a relatively large experimental dataset of cadaveric head/neck anterioposterior impact tests, this study found poor correlation between the predictions of injury risk using NII and injury outcome in the cadaveric dataset. This poor correlation may be attributed to the types of injuries and data on which NII is based, the limited number of injury cases result in parameters used in the injury risk function risk assessment, and possible limitations in the biofidelity of the MATD neck/dummy system. As the force and moment critical values for NII are similar to those for upper neck injury (Nij– Kleinberger, 1998) and lower neck injury (Beam – Bass, 2006), a new injury risk assessment for NII was performed ($NII_{cadaver}$) that retains the form of the function and critical values from ISO 13232-5 but significantly improves the predictive capability for the cadaveric experimental dataset by identifying the appropriate normalizing coefficients for the risk function parameters. Substantially improved predictions for the cadaveric dataset were obtained using optimized normalizing coefficients based on the experimental data while maintaining the form of the NII risk functions. The difference between the ISO-13232 predictions and those derived in this study should be viewed as an opportunity to assess and improve the biofidelity of the MATD dummy and simulated reconstructions using the MATD dummy.

Limitations of this study are primarily the limitations of the source data. The experimental dataset has predominantly flexion/compression and flexion/tension dynamics. This limits the validity of the resulting injury criterion to conditions similar to those seen in the experimental dataset. Also, cadaveric response may be different than human response under active and passive muscle control; the cadaver is neither a tensed nor a relaxed human. So, the coefficients derived in this study are strictly applicable only to a cadaveric specimen population. However, based on discussion above, it is

likely applicable as a conservative limit in tension/flexion and tension/extension modes of untensed human anterioposterior bending. Muscle tensing Further, the injuries in the cadaveric experimental dataset are predominantly to the lower cervical spine, but NII is intended to assess upper neck injuries. However, owing to limitations in the biofidelity of other existing crash test dummy necks suitable for anterioposterior bending (e.g. Thor, Hybrid III, etc.), the results in the current study are not directly applicable without a scaling procedure. Finally, there are likely differences between the motorcycle-riding population and the cadaver specimen pool; the cadaveric pool is likely more elderly than the population of motorcycle riders, and hence may have lower tolerance values.

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