The Development of IARV's for the Hybrid III Neck Modified for Dynamic Rollover Crash Testing

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Abstract - In the U.S., more than 27,000 catastrophic and fatal injuries occur annually in rollover crashes. This study is part of an ongoing research program aimed at mitigating these injuries. Recent papers introduced a prototype "soft" low-durometer Hybrid III neck design and presented results of matched-pair tests, comparing production and prototype Hybrid III neck responses. This paper

- discusses neck injury criteria, and
- proposes preliminary rollover injury criteria for the prototype "soft" low-durometer neck.

Then, peak neck injury measures are used to calibrate the dynamic relationship for flexion bending between the tensed production Hybrid III neck IARV and:

- the prototype "soft" low-durometer neck, which is representative of 1/3 tensed, and
- the untensed cadaveric logistic regression curves for major flexion injury risk by Pintar et al.

Keywords: Dummy, Neck, Biomechanics, Flexion

INTRODUCTION

This study is part of an ongoing research project aimed at mitigating catastrophic human neck injuries in rollovers. High-speed videotapes of dynamic rollover tests show that excessive neck bending is the predominant mechanism of these injuries, and that head injury potential is influenced by the neck's capacity to replicate real-world pre-trip occupant positioning. Recent papers introduced a prototype "soft" low-durometer Hybrid III neck design [2] and presented results of matched-pair tests, comparing prototype and production Hybrid III neck responses [6].

Production Hybrid III Neck Design

At present, the Hybrid III dummy is considered to be the best available human surrogate for dynamic rollover tests. The production Hybrid III neck is fabricated with 67-durometer butyl rubber discs and nodding blocks. Loading is measured and specified only at the occipital condyles, not at the lower neck, where bending injuries typically occur in rollovers. The Hybrid III neck bending stiffness, peak injury measures, and onset criteria are based on tensed volunteer musculature. Unlike the preflexed, untensed, and neither aligned nor oriented human neck, the production Hybrid III dummy neck is axially stiff, angularly tensed, aligned, and oriented to the roof intrusion force vector in rollover tests. The result is that the dummy is predisposed to predicting axial compression injury in rollover tests, not the flexion injury predominantly sustained by humans in real-world crashes.

Experiments [7,10,11] show that, relative to the human neck, the production Hybrid III neck is:

- 10 times stiffer in untensed bending, and
- equally stiff in tensed bending.

Prototype Neck Design

The Hybrid III neck was modified by replacing the stiff 67-durometer with softer 35-durometer butyl rubber discs and nodding blocks [2]. Static bending tests [6] have shown that, relative to the soft prototype neck, the production neck was:

- 3 to 4 times stiffer in extension,
- 3 times stiffer in flexion, and
- 5 to 6 times stiffer in lateral bending.

Dynamic bending tests [6] of the both production and prototype necks showed:

- The onset-to-peak neck loading is very fast, about 5 ms, and independent of impact speed and stroke (roof crush).
- Peak neck loading is dependent on speed, but independent of stroke (roof crush). Peak neck load occurs quickly, before injury and before the roof crushes significantly. *Peak neck load is not equal to bending injury*.
- Lower neck moment duration is dependent on stroke (roof crush), not speed.

Neck Flexion Injury Criteria

Most injury metrics are based on peak forces or moments. This leads to a misconception that all injury metrics, like HIC, are described by short-duration (16 or 32 ms) high-acceleration impulses. Research by these authors disagrees with the use of peak values, particularly with the non-biofidelic production neck and the fully-tensed musculature it represents. Dummy neck data and time-coordinated videos of rollover impacts show that the neck bending occurs after the peak axial force and is not isolated to the midsagittal plane, which suggests that neck flexion injury is a function of momentum and both neck Mx and neck My, not peak force or peak moment. The bending that produces injury is the result of the torque duration and/or angle through which the torque is applied. Neck bending injury only occurs with an extended stroke of 4 or more inches.

Injury Assessment Reference Values (IARV):

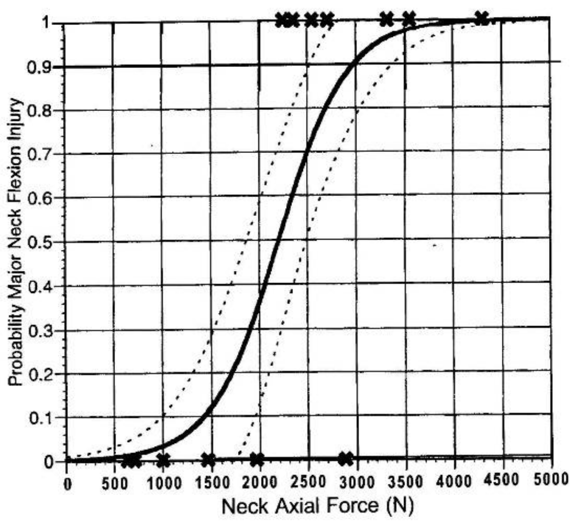
In 1984, Mertz [8] published the internal GM design criteria for the Hybrid III dummy neck (see Table 1). The metric, Injury Assessment Value (IARV), was defined as a lower bound of its Injury Threshold Level, below which a specified injury does not occur and above which the specified injury will occur for a given individual. The IARV limits are based on testing of volunteer tensed necks for frontal impact and then interpreted for side impact.

Table 1. IARV chart (from Mertz, et al., 2003), 50th Percentile Male Upper Neck (C7-T1), Out-Of-Position (OOP) v. In-Position (IP)

Production Hybrid III	**Axial Fz (N)**	**Moment My (Nm)**	**Moment Mx (Nm)**
Upper Neck (C7-T1) IP and OOP Flexion Moment	4,000	380	
Upper Neck (C7-T1) OOP Extension My / Lateral Mx	~2,000	~90	~90
Upper Neck (C7-T1) IP Extension My / Lateral Mx	~1,640	~59	~59

Human Cadaver Neck Flexion Injury Tolerance

In 1998, Pintar et al. [9] generated the regression curves in Figure 1, which show a 10% risk of major neck injury at a peak My of 58 Nm and peak Fz of 1,500 N. They also noted that flexion injury (i.e., bilateral locked facets) does not occur until after an anterior musculature reaction time of about 140 ms.



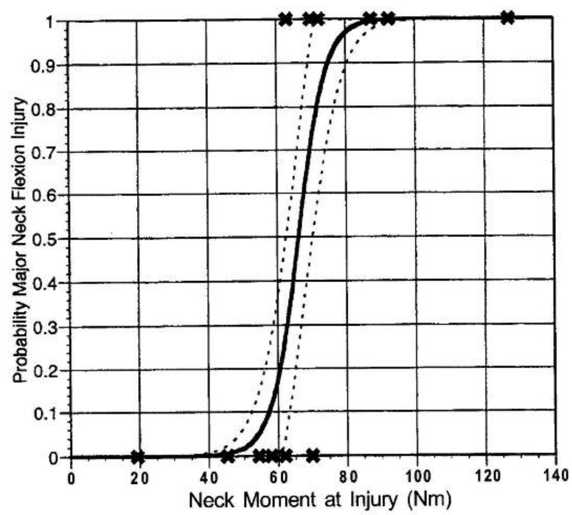


Figure 1: Logistic Regression Curves for the Probability of Major Neck Injury v. Peak Human Neck Axial Force (left) and Peak Human Neck Bending Moment (right) at the Level of Injury (abstracted from Pintar, et al.[9]).

Integrated Bending Moment (IBM)

In 2008, the Integrated Bending Moment (IBM) was proposed by Paver, et al. [5] as a neck flexion injury metric. It accounts for muscle contraction, which may take 100 ms or more, and combines measured lower neck Mx and My, integrated over the time duration of neck loading. For tests with only upper neck load cell data, the IBM was calculated from the upper Fz, integrated over the duration of neck loading, and multiplied by 0.048 plus 42 times that duration. The IBM can also be calculated from a combination of the measured lower neck bending moment and the angular bending.

The IBM is a departure from peak values, which have been found to be poor injury predictors by these authors. Instead, the IBM is the result of key findings by these authors that neck bending injury is primarily the result of forcible angular displacement in a momentum exchange approximated by the area under a moment-time plot.

METHODS

Testing was conducted to calibrate the dynamic relationship for flexion bending between the tensed production Hybrid III neck IARV and:

- the prototype "soft" low-durometer neck, which is representative of 1/3 tensed, and
- Pintar et al.'s untensed cadaveric logistic regression curves for major flexion injury risk.

The dummy neck was instrumented with upper and lower neck load cells that measured axial neck force Fz and moments Mx and My. Instrumentation also included real-time and high-speed rear and lateral view cameras. The high-speed cameras were equipped with tracking software used to analyze head-neck motion and neck flexion angle. Preflexion neck angles were measured with a gravity-referenced inclinometer at the posterior neck (erect=90°). String potentiometers attached to the platen and dummy recorded platen drop height and head, upper neck, lower neck, and lumbar spine fore-and-aft motion, respectively.

Pendulum tests were conducted at Xprts, LLC in Goleta, CA, using the experimental setup illustrated in Figure 2. In these tests, a rotational pendulum-driven platen struck the head of the dummy head/neck/torso/pelvis assembly suspended approximately 2" above a production vehicle seat. Bungee cords were used to set the dummy at initial platen contact to have a torso angle of 10° relative to the vertical and the head-neck angle of 17° relative to the torso, resulting in a total angle relative to the vertical of 27°. This orientation is representative of the preflexed orientation of the human when inverted with the head in contact with the roof. The dummy was held in place by a breakable attachment that allowed the dummy to fall after initial contact with the platen. The dummy fell onto the seat with the buttocks moving to a position, where the seat cushion was compressed about 1". The rotational pendulum was set to impact the head of the dummy when horizontal and moved through an angle of 10° when depressing the head of the dummy by a maximum of 6". The fixture included an adjustable arresting stop for the pendulum arm. The pendulum was dropped to achieve head impact at 11.3 kph (7 mph) with a pendulum displacement (stroke) of 5 cm (2") to 15.2 cm (6") before being arrested. The stroke after head contact was analogous to the extent of roof crush.

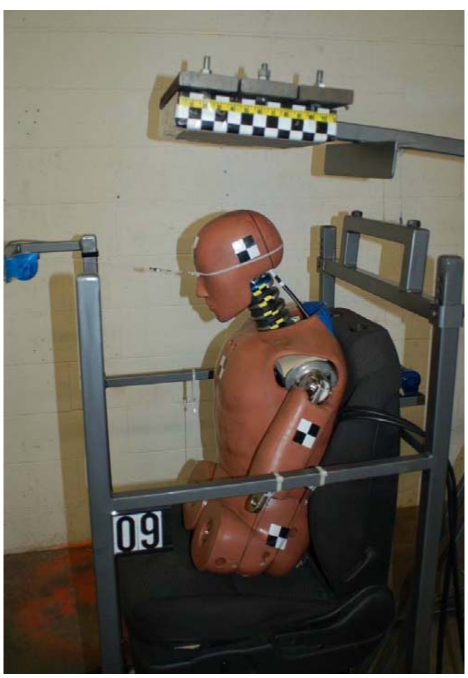


Figure 2: Test Setup

For most of the pendulum tests,

- the dummy-to-platen distance was 45.7 cm (18"),
- the dummy-to-seat distance was 5.1 cm (2"),
- the dummy-to-right side distance was 20.3 cm (8"),
- the dummy-to-front distance was 17.8 cm (7"),
- the neck angle was ~60° from horizontal, and
- the neck position was bent.

RESULTS

Table 2 shows the test matrix and peak value test results, where "P" refers to the production neck and "S" refers to the prototype “soft” low-durometer neck.

Table 2. Test Matrix and Peak Value Test Results

Test #	Platen Stroke (in/cm)	Platen Drop Height (in/cm)	Platen Weight (lb/kg)	Neck Angle (deg)	Peak Lower Neck Axial Load Fz (N)	Peak Lower Neck Moment My (Nm)
P-1	2 / 5.1	18 / 45.7	95 / 43	60	-4,276	272
P-2	6 / 15.2	18 / 45.7	95 / 43	60	-4,602	287
P-4	4 / 10.2	18 / 45.7	95 / 43	60	-4,425	282
S-3	4 / 10.2	18 / 45.7	95 / 43	60	-2,442	194
S-4	6 / 15.2	18 / 45.7	95 / 43	60	-2,299	181
S-5	2 / 5.1	18 / 45.7	95 / 43	60	-2,350	195

Dummy Neck Displacement Measurements

Figure 3 shows bending angle time-history plots of the production and prototype “soft” low-durometer necks with the same initial conditions and a 43 kg (95 lb) platen. The production neck bends about half as much as the prototype “soft” low-durometer neck. The production neck was also tested with a 70 kg (155-lb) platen because it is probably more representative of the effective mass and energy of roof intrusion; the production neck did not bend much farther. Had we tested the prototype “soft” low-durometer neck with the 70 kg (155-lb) platen, we estimate it would have bent 50% further.

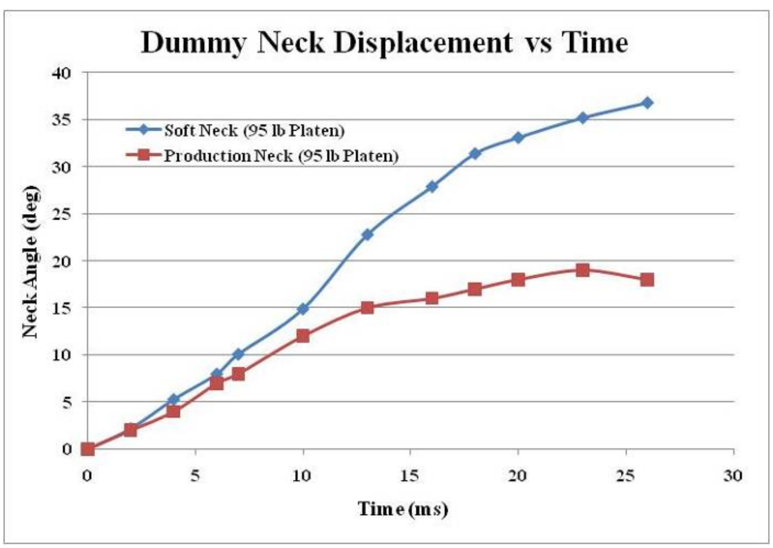


Figure 3. Pendulum Test. Bending angle as a function of time with 43 kg (95 lb) platen

Figure 4 shows time-history plots of the bending angle of the production and prototype "soft" low-durometer necks for different platen weights.

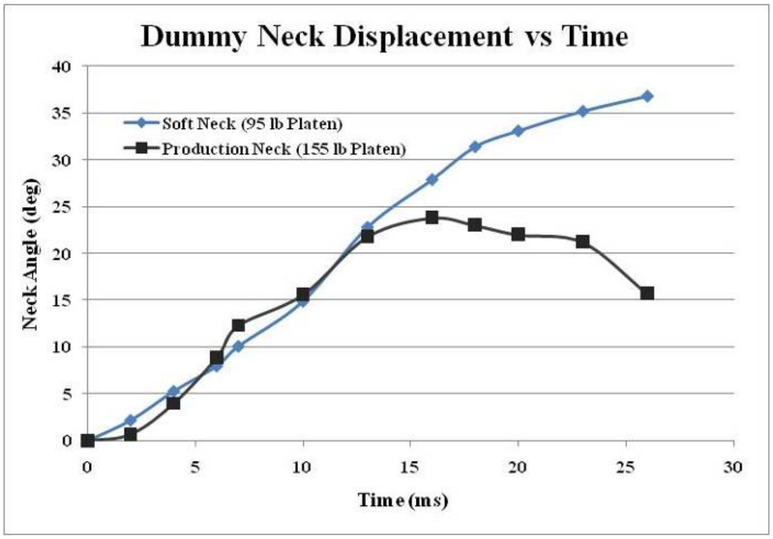


Figure 4. Pendulum Test. Bending angle as a function of time with different platen weights.

DISCUSSION:

The purpose of this work is to ultimately relate anthropomorphic dummy measures to human injury potential data that has been measured using human cadavers. We tested both the production and more realistic prototype "soft" low-durometer necks.

In 1991, Sances [10] conducted similar impactor tests of axially-aligned cadaver specimens and the Hybrid III neck. The peak F_z force measured in the cadaver specimen was roughly one-third that of the force in the Hybrid III. Comparing the peak force in the production Hybrid III neck (7,600 N) with the reduced-durometer neck (2,500 N) suggests that the response to a head impact of the prototype "soft" low-durometer neck is similar to that of a human neck. The key aspect of the pendulum impact is its speed, not its momentum.

In 1998, Pintar, et al. [9] related measured Hybrid III to human cadaver lower neck loading, and generated logistic regression curves for major (AIS≥3) neck flexion injury risk, also based on peak values. This indicates that the prototype "soft" low-durometer neck has the equivalent of 50% greater stiffness in flexion and 30% greater stiffness in compression than the cadaver neck. The difference in lateral bending is not as great.

Together, this data on the production Hybrid III and prototype "soft" low-durometer neck relative to existing cadaver data provide a direct peak value correlation to human neck bending and compression injury. This correlation can be derived from the prototype "soft" low-durometer neck peak correlation to cadaver data.

For the Hybrid III neck, Injury Assessment Reference Values (IARV's) are based on peak values (e.g., axial forces, bending moments) and/or a combination of peak values (e.g., Nij). The IARV values for the Hybrid III 50th percentile male are My = 380 Nm and Fz = 4,000 N at the lower neck C7-T1. Mertz et al. [4] opined that these values correspond to the onset of AIS ≥ 3 with an injury risk of less than 10%. [We also used the lateral bending IARV, to get an estimated range of lower neck bending IARV values.]

Table 3 below summarizes peak lower neck IARV's for a 10% probability of AIS≥3 neck bending injury.

Table 3: Peak Lower Neck IARVs for a 10% Probability of an AIS ≥ 3 Injury

Neck Type	Neck Loading Direction	Axial Fz (N)	Moment My (Nm)	Moment Mx (Nm)
Production Hybrid III	Flexion	4,000	380	
Production Hybrid III	Lateral Bending	4,000		268
"Soft" Neck	Flexion	~2,000	~90	
"Soft" Neck	Lateral Bending	~1,640		~59
Human/Cadaver	Flexion	~1,500	~58	

Note that on these charts that IARV peak values are used for injury assessment because they have been used for so long as an experimental injury measure. However, these authors do not agree that peak values are appropriate, particularly with the production Hybrid III neck with fully tensed musculature. While there may be an empirical relationship between peak force and bone fractures in axial compression, the neck is hardly ever oriented to interact axially with a collapsing roof.

In summary, the prototype "soft" low durometer neck is a reasonable preliminary working level approximation of the human neck in flexion with musculature to represent a mildly tensed neck. Since, in a rollover, bending is a composite of lateral and flexion or extension and there is substantial uncertainty that cannot be resolved as to musculature, we suggest the higher flexion IARV. We are sure that neck injury is predominantly a composite momentum exchange. We expect that, with better bending angle data, a critical IBM value will be derived and accepted.

CONCLUSIONS

Spinal cord bending injury usually requires forcible angular displacement in a momentum exchange which can be approximated by the area under a bending moment vs time plot. The IBM is such a measure. It can also be measured by a combination of the lower neck bending moment and the angular bending.

Results presented here indicate that the equivalent biofidelity of the prototype "soft" low-durometer neck is about 2/3 of the bending moment, adjusted by bending twice as much with 2/3 the pendulum weight. This equates to 2/3 x ½ x 2/3 of 380 Nm or 90 Nm for the IARV flexion moment and 59 Nm for the lateral bending moment of 268 Nm. The corresponding axial compression or tension values are half of the 4,000 N and 3,290 N IARV numbers. Pending further study, the prototype "soft" low-durometer neck would have corresponding IARV values of 59 to 90 Nm in bending and 1,640 N to 2,000 N in equivalent compression force.

In summary, the prototype "soft" low durometer neck is a reasonable preliminary working level approximation of the human neck with musculature in flexion to represent a mildly tensed neck. The equivalent biofidelity of the prototype "soft" neck is shown to be about 2/3 of the bending moment, adjusted by bending twice as much with 2/3 the pendulum weight. Since, in a rollover, bending is a composite of lateral and/or, forward flexion and/or extension and there is substantial uncertainty that cannot be resolved as to musculature, we suggest the higher flexion IARV. We are sure that neck injury is predominantly a composite momentum exchange and expect that with better bending angle data a critical IBM value will eventually be derived and accepted.

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