

Development of Lower Neck Injury Assessment Reference Values Based on Comparison of ATD and PMHS Tests

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ABSTRACT

Previous studies have suggested injury assessment reference values (IARVs) for lower neck injury based on scaled upper neck values. This study developed independent flexion and extension IARVs for the lower neck by matching Anthropomorphic Test Device (ATD) data to impact-tested post-mortem human subjects (PMHSs) with mid- to low-cervical spine injuries. Pendulum and sled tests with Hybrid III midsize male and small female ATDs were run under conditions mimicking those of published PMHS torso drop-sled tests and other PMHS studies. Measurements included upper and lower neck forces and moments, head acceleration, head rotation rate, and head/neck angles for the pendulum tests. Rear impacts corresponding to rigid seatback tests without a head restraint produced lower neck extension moments that increased dramatically with test severity, as measured by increasing delta-V and/or decreasing pulse duration. In contrast, upper neck extension moments increased only modestly with test severity, and all recorded peak values of upper neck force and moment, as well as calculated N_{ij} , were below currently accepted IARVs. Our proposed lower neck IARVs are well below those previously published, which are based on simple scaling of upper neck IARVs. Frontal impact tests provided an indication that injuries may occur at lower neck flexion moment values that are below suggested tolerances; however, since the available PMHS dataset only included injured subjects for which the observations were left censored, no IARV could be determined. Relationships between peak moments at the upper and lower neck, as well as between male and female ATDs in tests of matching severity, were examined and compared to standard scaling used for previously proposed IARVs. Peak moment versus tension and compression forces were also evaluated, yielding additional data points for

estimating critical intercepts for lower neck injury criteria (N_{ij}).

INTRODUCTION

There has been considerable research interest in human tolerance to forces and moments at the upper cervical spine, due to the occurrence of upper cervical injuries in airbag deployments. However, several trauma center studies indicate that while C2 is the most commonly injured cervical level, injuries to the C5, C6, or C7 levels are nearly as frequent [e.g., 1,2,3]. Studies of pediatric trauma admissions indicate that injuries at the C5-C7 level account for as many as half of the cervical spine injuries observed [4-5], though the likelihood of upper cervical spine injury appears higher for young children [6]. Cervical spine injury occurs most commonly as a result of motor vehicle accidents (including car, motorcycle, and pedestrian), falls, and sports injuries. Although lower neck Injury Assessment Reference Values (IARVs) are not included in current regulatory compliance limits, such IARVs have been proposed for the Hybrid III family of Anthropomorphic Test Devices (ATDs). However, the development of an accurate IARV requires access to relevant biomechanical injury data from controlled tests to compare ATD response data with human injury outcomes. While the injury basis for the upper neck IARVs and regulatory limits is well documented [e.g., 7,8,9,10; see below], the proposed lower neck IARVs are less supported. Additional post-mortem human subject (PMHS) data do, however, exist that should allow for the development of more accurate lower neck IARVs.

Mertz and Patrick [7] investigated neck injury tolerance in volunteer and PMHS testing. Two male PMHSs were subjected to rear impact in sled tests classified as non-severe (16 kph delta-V, 25.4 cm stopping distance, duration

approximately 130 to 150 ms) and severe (24 kph delta-V, 44.5 cm stopping distance, duration approximately 150 ms). In the severe test, the smaller PMHS sustained minor ligamentous damage to the mid-cervical spine, between C3 and C4. The peak upper neck moment value calculated from data measured in this test was scaled by head weight to obtain a threshold of 57 Nm for upper neck extension injury for a mid-sized male. Based on the peak moment for the uninjured PMHS, the authors also proposed a non-injury threshold of 47 Nm, or twice the volunteer tolerance for static loads. Additional PMHS and volunteer testing reported by Mertz and Patrick [8] provided the basis for a proposed upper neck flexion injury threshold of 190 Nm. This value corresponds to the highest moment sustained in all of the tests in which the PMHS was not injured.

Prasad et al. [11] ran a series of rear-impact rigid bench sled tests using midsize male ATDs. The tests were designed to replicate the characteristics from the earlier PMHS testing of Mertz & Patrick [7], with the most severe tests having a delta-V of 24 kph (6.7 m/s). Average extension moments for the 24-kph tests were 34.5 Nm for upper neck and 112.1 Nm for lower neck. The ratio between these peak moments, reported as 3.28, was then multiplied by the upper neck non-injury threshold of 47 Nm and by the extension IARV of 57 Nm to produce lower and upper bounds of 154 Nm and 187 Nm for a suggested lower neck extension moment IARV. More recently, Mertz et al. [10] utilized data from studies of animal and 3-year-old ATD airbag interaction [12,13,14] to develop revised IARVs for upper neck extension moment (78 Nm for Out-of-Position (OOP) testing, corresponding to relaxed neck muscles, and 96 Nm for In-Position (IP), where the neck muscles are assumed to be tensed at 80% of maximum muscle tone). The value for the additional moment generated by neck muscles was derived from the maximum static volunteer tolerance of 23.7 Nm [7]. Mertz et al. [10] also derived values for lower neck IARVs by dividing the previously suggested lower neck extension limit value of 154 Nm by the upper neck extension moment IARV of 78 Nm, which yielded an approximate ratio of 2.0. This multiplier was then applied to other flexion/extension IARVs, resulting in a suggested IP peak extension moment for the midsize male of 192 Nm, or twice the current upper neck IARV of 96, and a peak flexion moment of 380 Nm. Tension and compression IARVs were assumed to be identical to the corresponding values for the upper neck. In addition, flexion and extension critical intercept values for the combined axial force and moment Neck Injury Criteria, Nij, were estimated using the same scaling procedure.

Clemens & Burow [15] performed an extensive series of sled tests involving both frontal and rear impacts to PMHS torsos. The PMHSs were described as half males, half females, aged 50-90 years, but height and body weight information was not provided. Specimens were mounted to a mini-sled, which was dropped vertically from various heights and decelerated by

deforming steel sheets over a range of sled velocities and deceleration distances. In the rear impact or “retroflexion” tests without head restraints, the torsos were strapped to a rigid baseplate and supported by a steel backrest with foam upholstery. The torsos were reportedly “pressed slightly” into the backrests. The first series, at impact speeds ranging from 22 to 25.6 kph (6.1-7.1 m/s) and deceleration distances of 9.5 to 20 cm, produced lower cervical spine injuries concentrated in the C5-C6 area. These included injuries ranging from isolated intervertebral disc injury to compound injuries involving various combinations of injuries to the disc, joint capsule, anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), and ligamentum flavum, and fractures of the vertebrae. Additional tests run at lower speeds, 19.1 to 20.3 kph (5.3-5.6 m/s), resulted in no injury to the PMHSs. Tests with head restraints were also conducted, at an impact speed of 25 kph (6.9 m/s) and deceleration distances of 10 to 20 cm, with no injuries observed. The study also included a series of 21 frontal impact or “anteflexion” tests. These flexion tests were performed at a much higher severity than the extension tests, with speeds of 28.2 to 29.2 kph (7.8-8.1 m/s) and deceleration distances of 7.2 to 15 cm. As a result, all the specimens showed evidence of injury which were concentrated in the lower cervical spine, and included disc and joint capsule injuries, damage to the ALL, PLL, and ligamentum flavum, as well as the fracture to the vertebra. However, there were a number of cervical spine injuries reported at the C3-C4 level and above, possibly due to interactions with the foam block used to arrest forward motion of the head.

Hu et al. [16] performed sled tests at a nominal impact speed of 7 m/s with PMHSs seat-belted in vehicle seats with both rigid and deflecting seatbacks. Five males (age 57-64 years, average height 178±3 cm, and weight 71±12 kg) and one female (age 64, height 163 cm, and weight 60 kg) were tested. Although the seats were equipped with head restraints, tests were conducted with these in the lowest position, which reportedly prevented contact between the head of the test subject and head restraint. Two of the PMHSs in rigid seats and the three in deflecting seats were found to have lower cervical spine injuries, while one PMHS in a rigid seat was uninjured. The injuries sustained in the rigid seat tests were described as C6 compression fractures and C5-C6 vertebral arch fractures for one PMHS, and C6-C7 disk injury for the other.

The primary purpose of this study was to evaluate and refine suggested IARVs for lower cervical spine injury by recreating the test conditions of Clemens and Burow [15] as closely as possible in tests with Hybrid-III ATDs. Our secondary goal was to determine the relationship between ATD upper and lower neck flexion/extension moments, as well as the scaling between midsize male and small female data, by performing tests over a range of severities, as defined by delta-V and duration. Finally, the current Nij critical

intercepts for the lower neck were evaluated by comparing the injurious and non-injurious loading regions defined by our test series to the boundaries defined by previously proposed IARVs.

METHODOLOGY

A total of ten sled tests and 67 pendulum tests were performed, the latter including 36 tests in extension and the remainder in flexion. Several of these tests were set up to replicate as closely as possible the conditions of the PMHS testing performed by Clemens and Burow [15]. The initial criteria for matching were the reported values of impact speed and deceleration distance, as well as the general characteristics of the reported sled decelerations, such as pulse shape and duration. The rest of the tests were performed over a broad range of speeds and durations, shown in Figure 1.

A Hybrid III midsize male ATD was used for 48 pendulum tests (31 extension, 17 flexion) and all of the sled tests; a Hybrid III small female ATD was used for the remaining pendulum tests. ATD instrumentation included a lower neck load cell, upper neck load cell, head linear accelerometers, and head angular rate sensors. For pendulum tests, a linkage of two potentiometers, measuring head and neck angles, was attached to a pin passing through the occipital condyle point of the head-neck interface. Summing the two potentiometer outputs provided a measurement of head angle relative to the initial position.

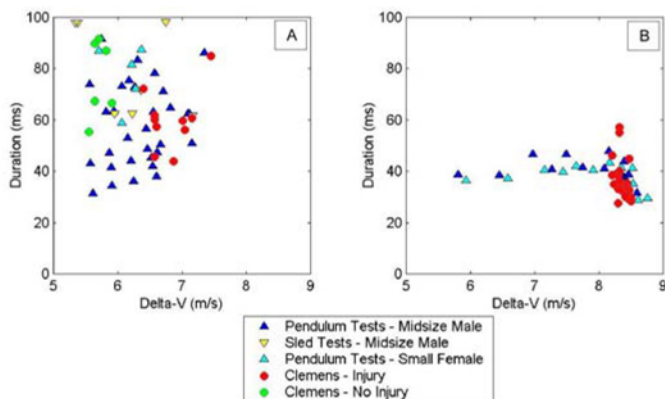


Figure 1. Scatter plot illustrating the range of impact severities obtained in the sled and pendulum extension (A) and flexion (B) tests, as indicated by delta-V and duration. Three sled tests at delta-Vs of 2-4 m/s (durations 95-101 ms) are not shown. Shown for comparison are the injurious and non-injurious tests of Clemens and Burow [15].

SLED TESTS

Sled testing was performed using a Seattle Safety sled with a wire-bending decelerator. Sled deceleration and high-speed and real-time video were recorded for all tests.

For the first four sled tests, the ATD was seated on a rigid bench mounted to the sled top plate, as shown in Figure 2A. The seat back of the rigid seat was angled at 9 degrees rearward of vertical. A 1-inch foam pad was attached to the seat back. The ATD torso was strapped to the seat with two crossing straps over the shoulders and a single horizontal strap across the chest to couple the ATD firmly to the sled and approximately replicate the positioning described and depicted by Clemens and Burow [15]. This resulted in the ATD head and neck being angled 8 degrees forward of vertical. This appeared to approximate the slight forward tilt of the head and neck seen in the photos shown in Clemens and Burow [15], although the authors describe some variation in initial inclination angle, with a plot of head inclination for three tests showing a range of initial head angle from approximately -1 to -10 degrees. The seating surface was horizontal, and unpadded except for a thin cloth mat. The ATD's arms were placed at its side and tied to the thighs. The ATD knees were flexed so that the lower legs were positioned underneath the seat to wedge the feet against the floor and minimize motion.

In order to best match the rapid rise times, short pulse durations, and relatively low deceleration distances utilized in previous test series, paired sled tests were performed. The first test ("wire bending") in each pair, conducted at sled velocities ranging from 7.2 to 7.9 m/s and involving deceleration distances of 26.7 to 27.7 cm, was performed to pre-bend the decelerator wires for the subsequent test. The second test ("pre-bent") was conducted at velocities ranging from 5.4 to 6.5 m/s and involved deceleration distances of 11.6 to 14.0 cm. The impactor struck the pre-bent wires (from the initial test), causing a rapid rise in deceleration and substantially shorter stopping distance. Data were collected in both wire-bending and pre-bent tests, but only the data from the four pre-bent tests were used for further analysis, as it was determined that the extreme test severities of the wire-bending tests caused the back of the ATD head to interact with the lower neck load cell, rendering the neck load cell data questionable.

An additional six extension sled tests were run at lower severities (delta-Vs of 2 to 7 m/s) to expand the range of data collected. In these tests (part of a separate study with other ATDs), the Hybrid III midsize male ATD head-neck complex was mounted directly to the sled with the head neck tilted forward 7 degrees, comparable to that in the full ATD sled tests. Although a complete ATD could not be utilized for these tests, it was felt the head-neck complex should produce results similar to the pendulum tests described below.

Because of the difficulty in recreating the extreme severity (high speed and relatively short stopping distance) in the Clemens and Burow [15] flexion tests, full-scale sled tests were run for neck extension only.

PENDULUM TESTS

The pendulum apparatus designed for calibration of Hybrid II and Hybrid III necks was used for further testing. The small mass of the pendulum allowed for substantially shorter stopping distances than could be obtained with a sled, allowing impacts of greater severity for a given initial velocity. Because the inclusion of a fixed-angle lower neck load cell precluded use of the adjustable lower neck bracket, the head and neck were tilted 13 degrees forward relative to vertical at the point of impact.

Pendulum deceleration was provided by aluminum honeycomb (Hexcel®, cell size 3/4", foil thickness 0.0026", density 1.8 lb/ft³), which was crushed by an impactor surface on the pendulum. The impact speed of the pendulum was varied by changing the drop height, and the stopping distance or duration was altered by varying the number of cells (range 10-41) and/or thickness (6" or 8") of honeycomb. The pendulum accelerometer and speed trap data and measurements of pendulum geometry were used to calculate the acceleration and velocity profiles of the base of the pendulum where the lower neck load cell was mounted.

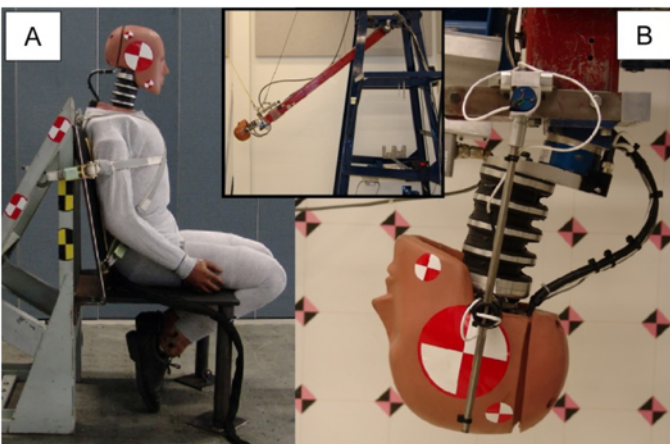


Figure 2. A. Sled test setup with midsize male ATD strapped to rigid seat (extension tests only). B. Pendulum setup with ATD head-neck complex (used as is for extension tests, reversed position for flexion tests). Small inset picture shows overall view of pendulum fixture.

DATA ANALYSIS

All test data were collected at 10 kHz and processed and reported according to SAE J211. Nij for the upper neck was computed using the critical intercepts specified in FMVSS 208, as well as the OOP intercepts tabulated by Mertz et al.

[10]. Nij for the lower neck was computed based on the IP and OOP critical intercepts suggested by Mertz et al. [10].

In order to account for differences in restitution between the pendulum tests, sled tests, and the tests of Clemens and Burow [15], the deceleration traces for each sled and pendulum test were integrated to determine delta-V, and pulse duration was determined from the first zero crossing of the deceleration trace after impact. Multiple regression was then utilized to relate delta-V, duration, and maximum lower neck extension moment. The resulting relationship was visualized in two dimensions by generating a contour map relating delta-V, duration, and isolines of neck extension moment maxima. In order to compare these results to the PMHS tests, the retroflexion acceleration trace reported by Clemens and Burow [15] for their Test 31 was digitized and integrated in order to examine velocity and displacement characteristics. It was found that the reported deceleration distance was less than the double integral of the deceleration pulse, given the reported speed at impact; therefore, it was concluded that this most likely represented measured residual crush rather than a measure of total sled displacement during the deceleration pulse. From this analysis, three ratios were computed: (1) delta-V divided by reported impact speed, (2) dynamic stopping distance over reported deceleration distance (crush), and (3) actual pulse duration over duration predicted for a square pulse. The delta-V/speed ratio (1.049) was found to be near the low end of those measured in our tests (range 1.027-1.172), while the stopping distance over crush ratio (1.263) was at the high end of those seen in our tests (range 1.009-1.286). These ratios were then used to scale the impact speeds and deceleration distances for the remainder of the test series of Clemens and Burow for the purpose of estimating delta-V and duration.

Also included for comparison were the rigid seat tests by Hu et al. [16]. Although complete details of the sled test inputs were not provided, the authors indicate a difference in severity, as reflected by peak accelerations of 19.0 g and 19.1 g for the injured PMHSs and 14.5 g for the uninjured PMHS. The authors note stopping distance was independent of velocity due to the water braking mechanism used; therefore, given the reported differences in acceleration, it appears likely that the uninjured PMHS was subjected to a lower delta-V and longer duration than the injured PMHSs. To obtain delta-V and duration values for these tests, acceleration curves were estimated based on scaling the reported acceleration trace so that peak accelerations matched the reported peak values, with the same stopping distance (approximately 25.4 cm) as predicted by the original trace.

As noted above, independent multiple regression analyses were applied to the extension and flexion data sets to identify the significant predictors of the lower neck peak moment values ($M_{\text{ext, peak}}$, $M_{\text{flex, peak}}$) and to determine the

relationship between the identified significant predictors and these maxima. The independent variables considered as potential predictors were ΔV , $(\Delta V)^2$, duration, $\Delta V \times \text{duration}$ (the interaction between ΔV and duration), and $(\Delta V)^2 \times \text{duration}$ (the interaction between $(\Delta V)^2$ and duration). A potential predictor was considered significant (and thus included in the model) if adding it to the model resulted in a significant ($p < 0.05$) increase of the multiple correlation coefficient R^2 (the percentage of variance in the dependent variable explained collectively by all of the independent variables in the model). The overall explanatory effect of the model was determined from the adjusted multiple correlation coefficient R^2_{adj} rather than R^2 in order to account for the number of predictors included. The regression analyses were performed using MINITAB (Minitab, Inc., State College, PA, version 15). The final regression model was checked to identify any data points with large standardized residual and characterize the influence of any data points with high leverage. Additional regression diagnostics performed included evaluation of: the predicted multiple correlation coefficient $R^2_{\text{predicted}}$ (a measure of how well the model predicts responses for new observations), plots of the residuals versus the model predictors, the ratio of the root mean square (RMS) error and the standard deviation of the dependent variable (in a good model, this ratio is markedly less than 1), and the ratio of the mean of the dependent variable and the RMS error (in a good model, this ratio exceeds 1.96).

RESULTS

EXTENSION TESTS

The use of the relatively low mass pendulum apparatus and the adoption of the pre-bending procedure in the sled tests yielded test decelerations that compared favorably to the “retroflexion” acceleration trace reported by Clemens and Burow (Figure 3). Sled and pendulum tests involving matched impact severities produced similar temporal profiles and maxima in upper and lower neck extension moments (Figure 4), although some variations are seen in dynamic oscillations at the initial rise segment of the sled pulse, while a late peak due to initiation of honeycomb saturation can be seen in the pendulum pulse. Modest differences were observed in the overall time duration of the bending moments, with a slight delay observed in response in the sled tests, and the lighter pendulum apparatus and its stiffer deceleration characteristics producing comparatively larger and more rapid neck rebound loading.

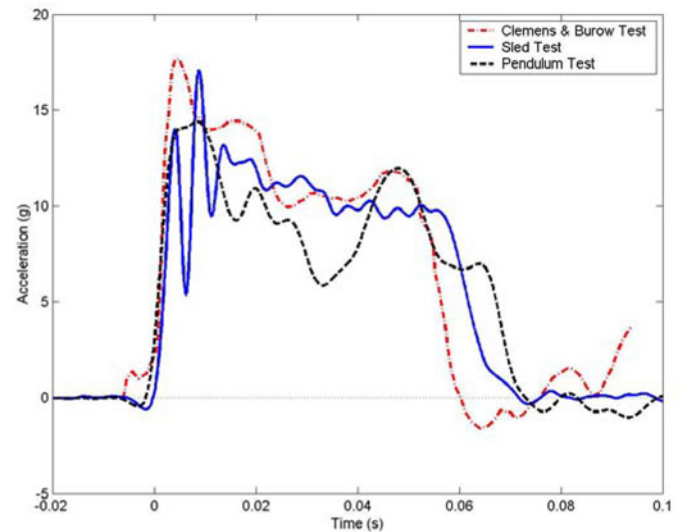


Figure 3. Representative sled and pendulum decelerations obtained in this study. The tests shown had severities similar to that of “Test 33” of Clemens and Burow (speed 6.1 m/s, deceleration distance 14.6 cm), the least severe injurious PMHS test they reported. Shown for comparison is the only extension acceleration trace reported by Clemens and Burow, “Test 31” (6.3 m/s, 12.5 cm). Of note, the pulse duration determined for Test 31 (60 ms) is shorter than that of Test 33 (72 ms) and the sled or pendulum tests (72 and 73 ms, respectively)