5 Injury Risk Assessments Based on Dummy Responses

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Injury assessment reference values (IARVs) were developed by Mertz[1,2] in 1978 to assess the efficacy of General Motors (GM) restraint system designs under a variety of simulated frontal accident conditions using the Hybrid III midsize adult male dummy as the vehicle occupant. These design limit values were chosen such that if an IARV was not exceeded in the prescribed test, then the risk of the associated injury would be unlikely. "Unlikely" was defined as risk levels less than 5%. Since injury risk curves for the various dummy responses did not exist, the limit values that were chosen for the IARVs were simply conservative estimates of response values that would be consistent with the unlikely definition. These IARVs were published in 1984 as part of the GM petition of the National Highway Traffic Safety Administration (NHTSA) to allow the use of the Hybrid III midsize adult male dummy in Federal Motor Vehicle Safety Standard (FMVSS) 208 testing. In 1993, IARVs were published for response measurements of the Hybrid III small female and large male adult dummies.[3,4] These values were obtained by applying constant failure stress scaling to the Hybrid III midsize male IARVs, taking into account size differences.[5,6] IARVs have also been developed for the response measurements of the child restraint air-bag interaction (CRABI) and Hybrid III child dummies.[7–10] For these IARVs, variations in tissue strength with age as well as size were considered in the scaling relationships. A summary of all the IARVs defined for the CRABI and Hybrid III family of dummies as well as the midsize male side impact dummies (SID, EUROSID-1, and BIOSID) are given in Chapter 4, Tables 4.6, 4.7, and 4.8.

In 1998, the American Automobile Manufacturers Association (AAMA), which is now the Alliance of Automobile Manufacturers (Alliance), proposed to NHTSA that the FMVSS 208 certification limits should be set at a 5% injury risk level.[11] This would put all FMVSS 208 limits on an equal injury risk level that is consistent with the "injury unlikely" intent of the IARVs while allowing for a reasonable compliance margin. The injury risk curves used by the Alliance were based on those proposed by Mertz et al[12] for responses of the CRABI and Hybrid III family of dummies. This chapter summarizes the injury risk curves proposed by the Alliance.[11]

Head Injury Risk Curves

Prasad and Mertz[13] and Mertz et al[14,15] have published injury risk curves for skull fracture (Fig. 5.1) and for Abbreviated Injury Scale (AIS) ≥4 brain injury due to forehead impacts based on the 15-ms HIC criterion (Fig. 5.2), and for skull fracture based on peak resultant head center of gravity acceleration (Fig. 5.3). These curves represent estimates of the injury risks for the adult population since adult cadavers were used to obtain the biomechanical data and these data were not normalized for size, mass, and tissue strength effects.

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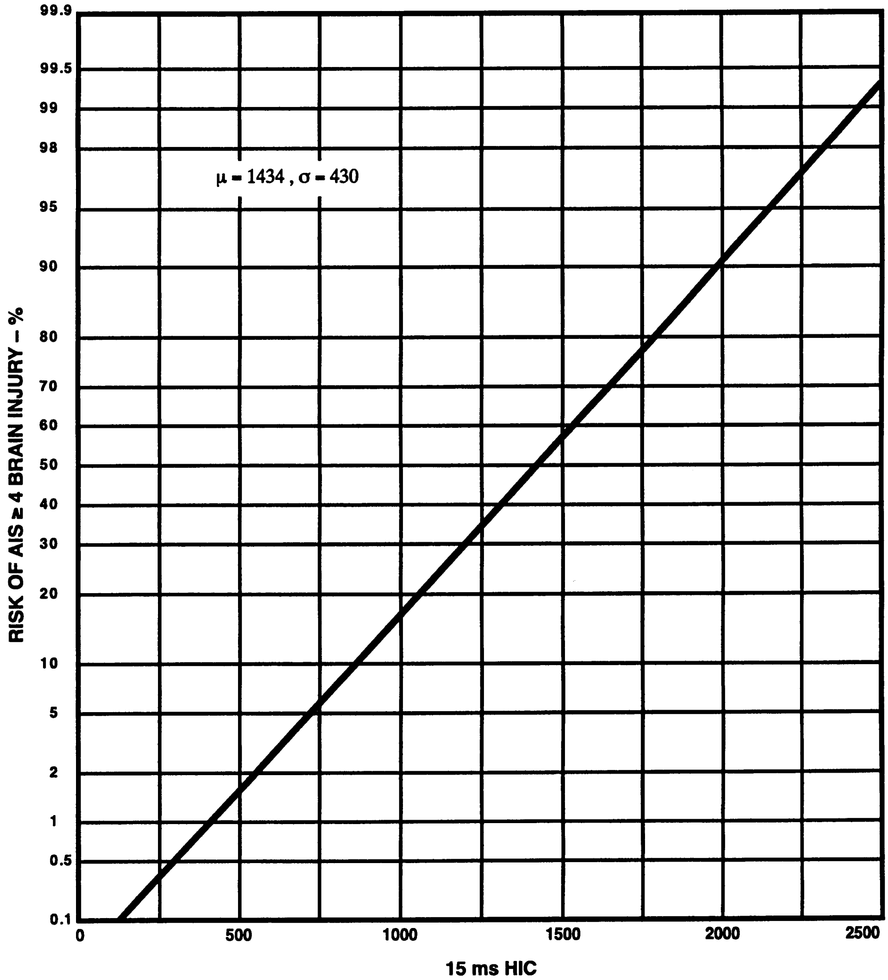


FIGURE 5.1. Risk of skull fracture as a function of 15-ms HIC for forehead impacts.

The Alliance proposed to use the 5% risk of AIS ≥4 brain injury, which is 15-ms HIC = 700, as the FMVSS 208 compliance limit for the midsize adult male dummy. Note this is the limit value imposed by the Canadian regulations, Canadian Motor Vehicle Safety Standard (CMVSS) 208. Since the brain injury risk curve of Fig. 5.2 is for the adult population, the 15-ms HIC value of 700 is a conservative estimate of the 5% risk level for the midsize male. The Alliance[11] proposed to obtain estimates of 5% risk levels of AIS ≥4 brain injury for other sizes of dummies by scaling this estimate using the following relationships:

$$\text{HIC Ratio: } \lambda_{HIC} = \lambda_\sigma^{2.5}\,\lambda_L^{-1.5}$$

where

acceleration ratio $\lambda_a = \lambda_\sigma\,\lambda_L^{-1}$
time ratio $\lambda_t = \lambda_L$
head size ratio is λ_L
failure stress ratio is λ_σ.

Since there are no data on the variation of failure stress of brain tissue with age, the Alliance made the assumption that its variation is the same as the variation of the calcaneal tendon noted by Melvin.[10] The ratios of failure stress and head size for the various ages and sizes of dummies compared to the midsize adult males are given in Table 5.1, along with the estimates of the HIC values corresponding to 5%

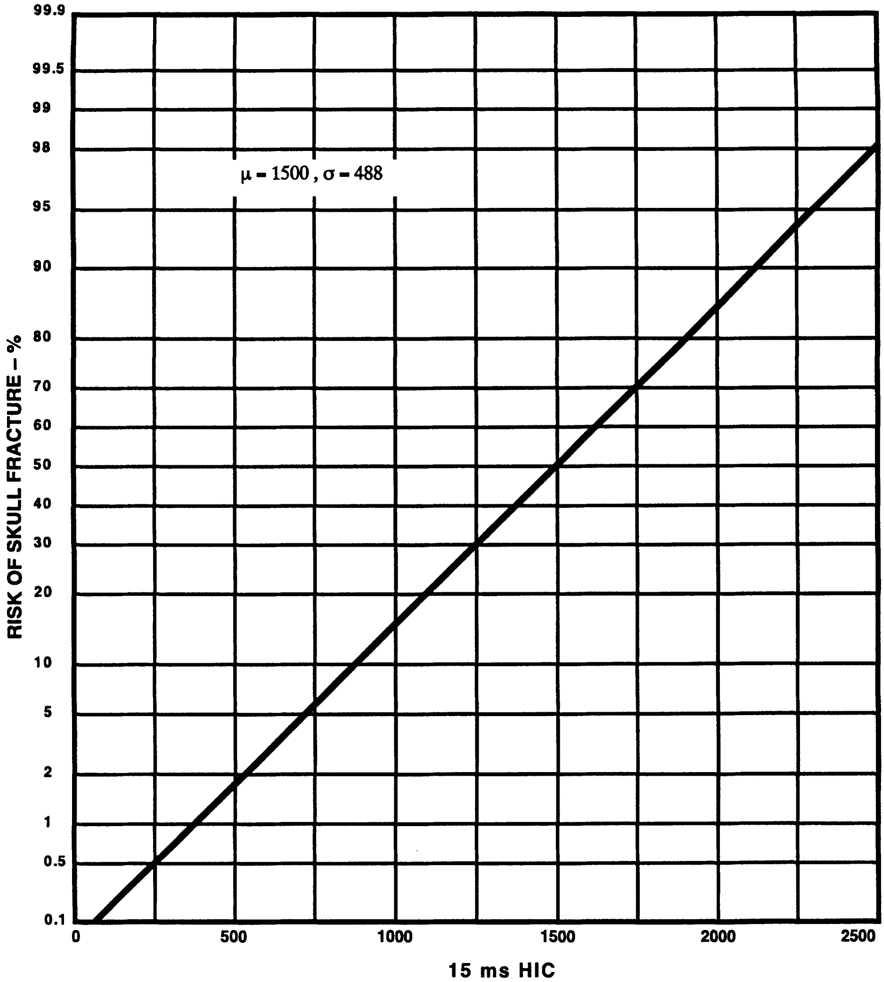


FIGURE 5.2. Risk of Abbreviated Injury Scale (AIS) ≥4 brain injury as a function of 15-ms HIC for forehead impacts.

TABLE 5.1. Scale factors and corresponding 15-ms HIC limits for 5% risk of AIS ≥ 4 brain injury for CRABI and Hybrid III dummy families.

	Scale factors				5% risk of AIS ≥ 4	
Dummy	λ_L	λ_{σ_f}	λ_t	λ_{HIC}	HIC τ (ms)	HIC
CRABI 6mo	0.775	0.67	0.775	0.539	11.6	377
CRABI 12mo	0.817	0.70	0.817	0.555	12.3	390
CRABI 18mo	0.844	0.75	0.844	0.628	12.7	440
HIII—3yr	0.876	0.85	0.876	0.812	13.1	570
HIII—6yr	0.914	0.96	0.914	1.033	13.7	723
HIII—Small female	0.931	1.00	0.931	1.113	14.0	779
HIII—Midsize male	1.000	1.00	1.000	1.000	15.0	700
HIII—Large male	1.030	1.00	1.030	0.957	15.5	670

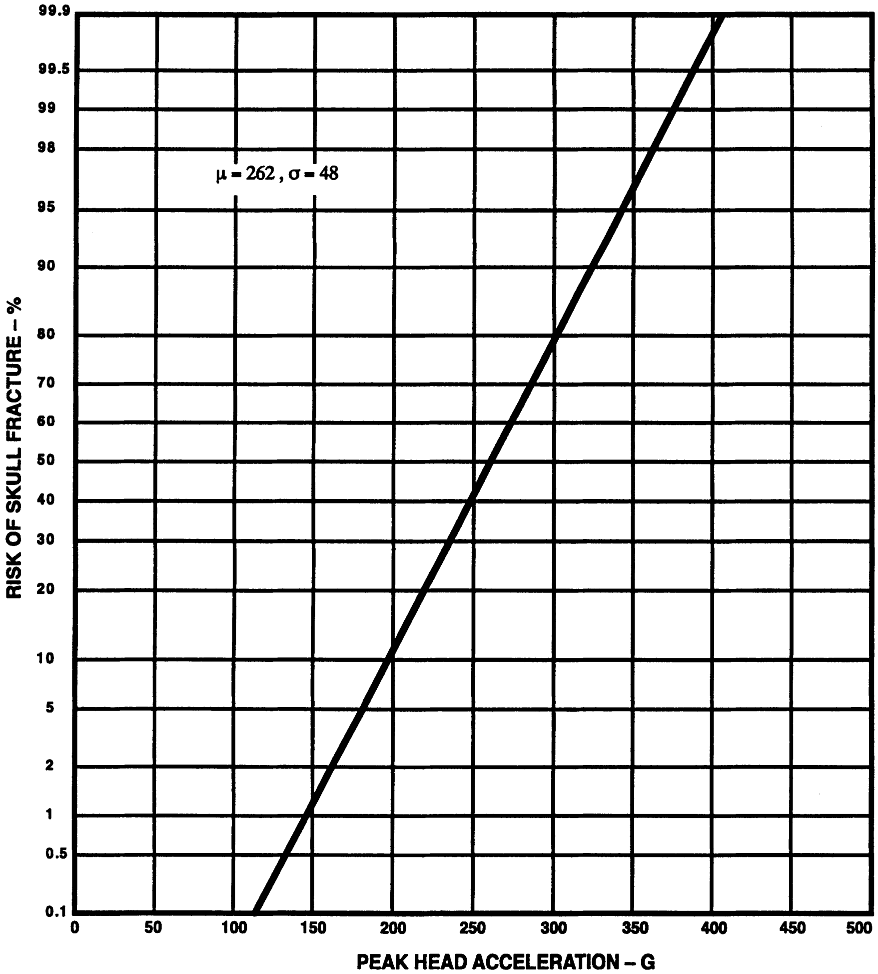


FIGURE 5.3. Risk of skull fracture as a function of peak resultant acceleration of the center of gravity of the head for forehead impacts.

risk of AIS ≥4 brain injury. Also given in Table 5.1 are the time periods that should be used for searching for maximum HIC values. Note that the time ratio is proportional to the head size ratio and independent of the ratio of skull bone elastic moduli. This is because the high bulk modulus of the brain controls the stiffness of the cranium.[7] To simplify the HIC calculation, the Alliance recommended using 15 ms as the limit for all HIC calculations.

Neck Injury Risk Curves

Neck Tension and Extension Moment

Mertz et al[16] and Prasad and Daniel[9] have conducted tests to assess the effects of deploying passenger air-bag interactions with animals (10-week-old pigs) that were chosen to represent the size, weight, and state of tissue development of a 3-year-old child. In their studies, a series of matched tests was conducted where for every pig test a similar test was conducted using the 3-year-old "air-bag" dummy.[17] This allowed the various injury severities experienced by the pig to be correlated with corresponding dummy response measurements. The neck injury observed in both studies was initiated by the rupturing of small blood vessels of the membrane that encase the occipital condylar joints. As the impact severity increased, the damage progressed to rupture of the alar ligament, damage to the spinal cord and brainstem, and finally to death. Blood in the synovial fluid of the occipital condylar joint capsules was rated as an AIS = 3 neck injury and was defined as the threshold of undesirable neck trauma. Both studies showed that neck tension was the best indicator of the onset of AIS ≥3 neck injury with no AIS ≥3 neck occurring below a neck tension load of 1,160 N.

Mertz and Weber[8] analyzed the Mertz et al[16] data and provided an injury risk curve for AIS ≥3 neck injury based on neck tension measured with the 3-year-old airbag dummy. This injury risk curve was update by Mertz et al[12] by combining the data of Mertz et al[16] and Prasad and Daniel.[9] Mertz et al[12] also provided an injury risk curve for AIS ≥3 neck injury based on peak neck extension moment measured at the occipital condylar joint. They normalized the data for size, which allowed them to generate risk curves for use with the CRABI and Hybrid III family of dummies. The Alliance[11] extended the analysis by normalizing for differences in tissue failure stress levels due to age. The scaling relationships used for tension and extension moment are:

Neck Tension Failure Ratio: $\lambda_F = \lambda_\sigma \lambda_C^2$

Neck Extension Moment Failure Ratio: $\lambda_M = \lambda_\sigma \lambda_C^3$

where

neck circumference ratio is λ_C
failure stress ratio is λ_σ.

Again, since there are no data for the variation failure stress of neck ligaments as a function of age, the calcaneal tendon data noted by Melvin[10] were used.

Figures 5.4 and 5.5 are the injury risk curves for neck tension and neck extension moment for the various size dummies. Table 5.2 provides a summary of the scale factors and the limit values for 3% and 5% risk of AIS ≥3 neck injury due to neck tension and extension moment, respectively. The Alliance chose to use a 3% risk level for the neck tension since the biomechanical database contained a fatal neck lesion at the 5% risk level. These values, as well as the injury risk curves shown in Figs. 5.4 and 5.5, are for minimum muscle tone; that is, the neck muscles are tensed only to the degree required to keep the head upright.

TABLE 5.2. Scale factors and values of peak neck tension and peak neck extension moment for 3% and 5% risks of AIS ≥ 3 neck injury, respectively, for CRABI and Hybrid III dummy families.

					Peak neck loads			
			Maximum muscle effect		Tension (N) for 3% risk AIS ≥ 3 injury		Extension Moment (Nm) for 5% risk AIS ≥ 3 injury	
	Scale factors							
Dummy	λ_C	$\lambda_{\sigma f}$	Tension (N)	Extension moment (Nm)	Minimum muscle tone	80% muscle tone	Minimum muscle tone	80% muscle tone
CRABI 6 mo	0.906	0.79	*	*	730	730	10	10
CRABI 12 mo	0.918	0.82	*	*	780	780	11	11
CRABI 18 mo	0.926	0.88	*	*	850	850	12	12
HIII—3-yr	1.000	1.00	446	6.1	1,130	1,480	17	22
HIII—6-yr	1.082	1.13	522	7.8	1,490	1,910	24	30
HIII—Small female	1.246	1.18	693	11.9	2,070	2,620	39	49
HIII—Midsize male	1.570	1.18	1,100	23.7	3,290	4,170	77	96
HIII—Large male	1.725	1.18	1,330	31.5	3,970	5,030	102	127

*No allowance for muscle effect for infants.

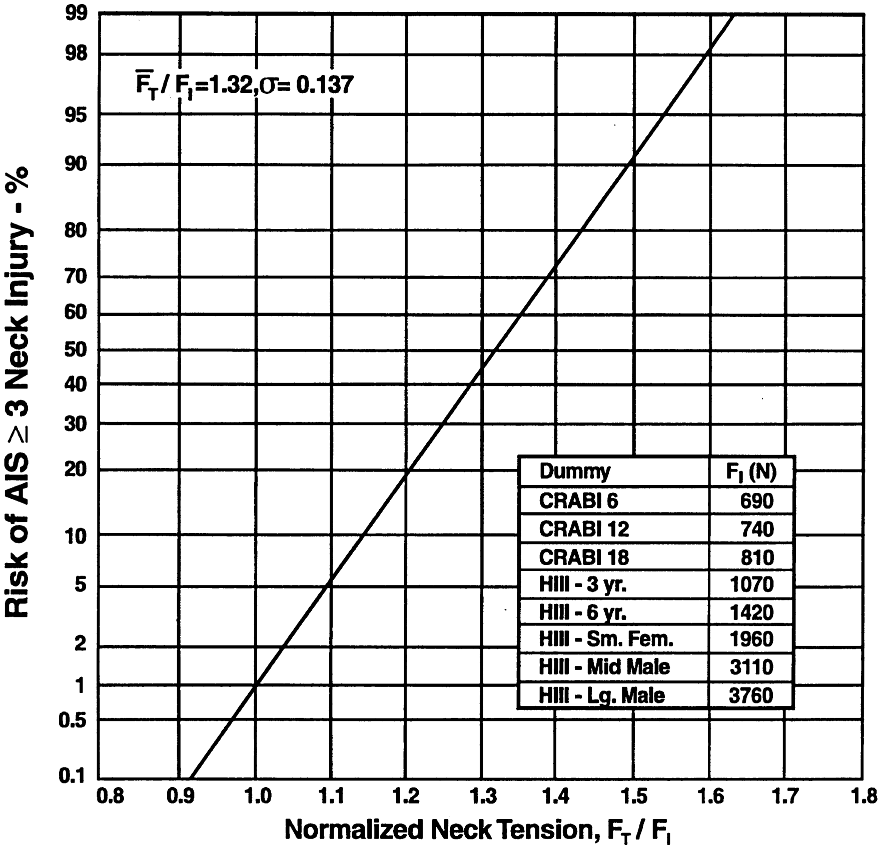


FIGURE 5.4. Risk of AIS ≥3 neck injury as a function of normalized neck tension for the child restraint airbag interaction (CRABI) and Hybrid III dummy familes. Minimum muscle tone.

Muscle tensing will increase the load-carrying capacity of the neck compared to its relaxed state since the applied load must exceed the passive muscle load before load will be transmitted by the ligaments. Based on the static neck muscle strength data of Mertz and Patrick,[18] estimates of the maximum passive neck loads for various dummy sizes were calculated[11,12] and are listed in Table 5.2. To use the neck injury risk curves shown in Figs. 5.4 and 5.5 when muscle tone is present, one must prescribe a degree of muscle tone. For example, if 80% muscle tone is assumed, then the muscle load is 0.8 of the maximum value given in Table 5.2 for the dummy size being used. The ligament load is calculated as the measured neck load minus the muscle load. The resulting ligament load is used to determine the risk level based on the injury risk curves of Figs. 5.4 and 5.5.

Combined Tension and Extension Moment

Prasad and Daniel[9] suggested that combining the neck bending moment and axial load at each instant in time to give time-varying functions of the loading of neck tissues that are likely to fail under the loading condition being simulated would be desirable. While they noted that certain straight lines on a graph of axial tension vs. extension moment would be constant stress lines, they provided no equations for such lines. Mertz et al[12] proposed the following linear combination of neck extension moment and neck tension, K, to estimate the time-varying stress level in the anterior-longitudinal ligament at the level of the head-neck interface.

$$K = [M_E + \lambda_C D_3 F_T/2] = \sigma \lambda_C{}^3 A_3 D_3$$

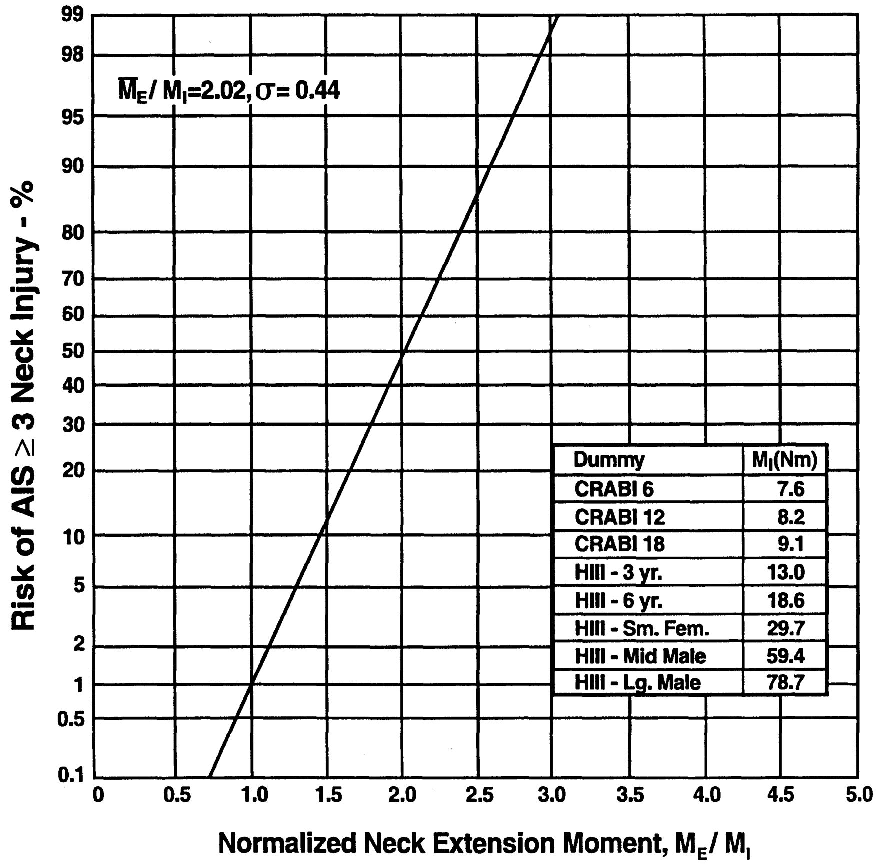


FIGURE 5.5. Risk of AIS ≥3 neck injury as a function of normalized neck extension moment for the CRABI and Hybrid III dummy families. Minimum muscle tone.

where

σ is the stress in the anterior-longitudinal ligament of the 3-year-old child,

M_E is the absolute value of the extension bending moment about the occipital condylar axis at time t_i,

λ_C is the ratio of neck circumferences of the dummy size being evaluated compared to the 3-year-old child,

D_3 is the anterior-posterior dimension of the atlas of the 3-year-old child (25.2 mm used by Mertz et al[12]),

F_T is the absolute value of the measured neck tension at the occipital condyles at time t_i,

A_3 is the cross-sectional area of the anterior-longitudinal ligament of the 3-year-old child.

To develop this relationship, the force in the anterior-longitudinal ligament was assumed to be of the form,

$$F_L = M_E / D_3 + F_T / 2$$

The Alliance[11] extended the analysis of Mertz et al[12] by allowing a variation in failure stress of the ligament with age which gives,

$$K = \lambda_\sigma \, \sigma \, \lambda_C^{\,3} \, A_3 \, D_3 = [M_E + \lambda_C \, D_3 \, F_T / 2]$$

where λ_σ is the ratio of failure stresses of the anterior-longitudinal ligament of the dummy age being evaluated compared to the 3-year-old child.

Again, since there are no data on the variation of failure stress of the ligament with age, the Alliance used the data of the calcaneal

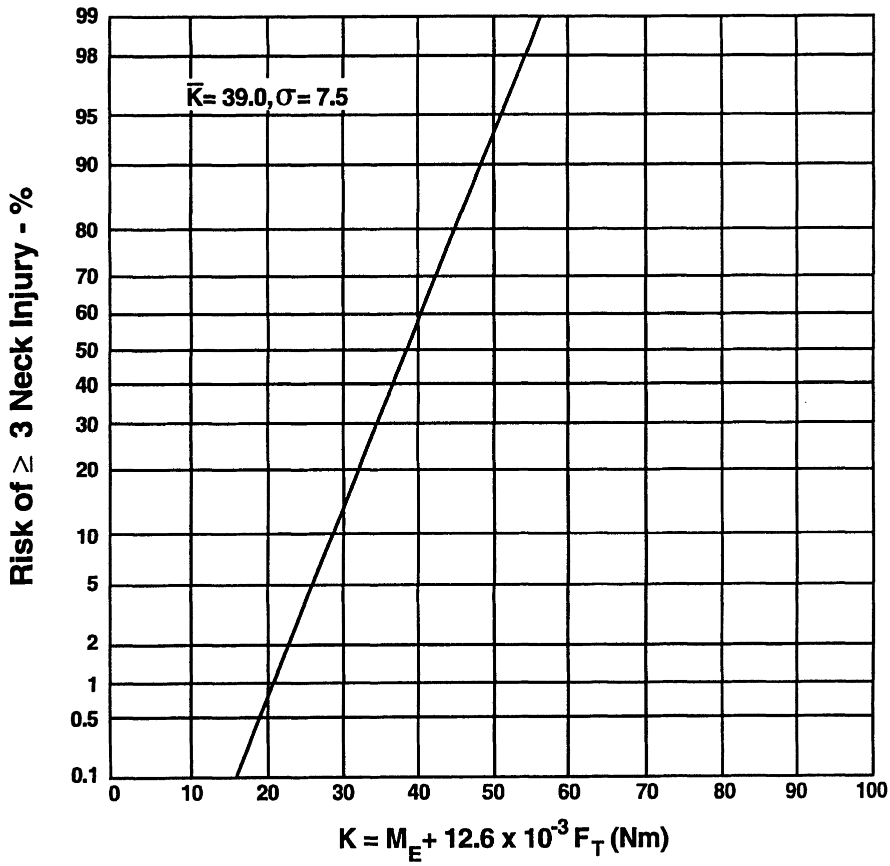


FIGURE 5.6. Risk of AIS ≥3 neck injury as a function of combined neck extension moment taken with respect to the occipital condyles and neck tension for the Hybrid III 3-year-old dummy. Minimum muscle tone.

tendon noted by Melvin.[10] An injury risk curve was developed for the kernel, K, for a 3-year-old child by using the combined data sets of Mertz et al[8,16] and Prasad and Daniel[9] and is shown in Fig. 5.6. This curve was proposed by the Alliance[11] and differs from the curve developed by Mertz et al[12] in that an outlier was deleted from the data set used by the Alliance. For a 1% risk of an AIS ≥3 neck injury for the 3-year-old, $K = 21.6$. The corresponding stress level is,

$$\sigma_1 = 21.6/A_3 D_3$$

A normalized stress ratio can be defined as,

$$N_{TE} = \sigma/\sigma_1 = [M_E + \lambda_C D_3 F_T/2]/(21.6\lambda_\sigma \lambda_C^3)$$

The normalized stress ratio, N_{TE}, can be expressed in terms of the ordinate value, M_C, and the abscissa value, F_C, of the constant stress line corresponding to 1% risk of AIS ≥3 neck injury for any size occupant which is,

$$N_{TE} = M_E/M_C + F_T/F_C$$

where

$$M_C = 21.6\ \lambda_\sigma\ \lambda_C^3 \text{ when } M_E \text{ is in Nm}$$

$$F_C = 1710\ \lambda_\sigma\ \lambda_C^2 \text{ when } F_T \text{ is in N.}$$

The resulting injury risk curve is shown in Fig. 5.7. The values for M_C and F_C for the various sizes of dummies were calculated using the values for λ_σ and λ_C given in Table 5.2 and are shown on Fig. 5.7.

There are three concerns with the injury risk curves for N_{TE}. First, the K values used to calculate the injury risk curve for the 3-year-old (Fig. 5.6) may not have been the maximum K values for the tests since K was not calculated as a function of time for the tests. Instead K_{max}

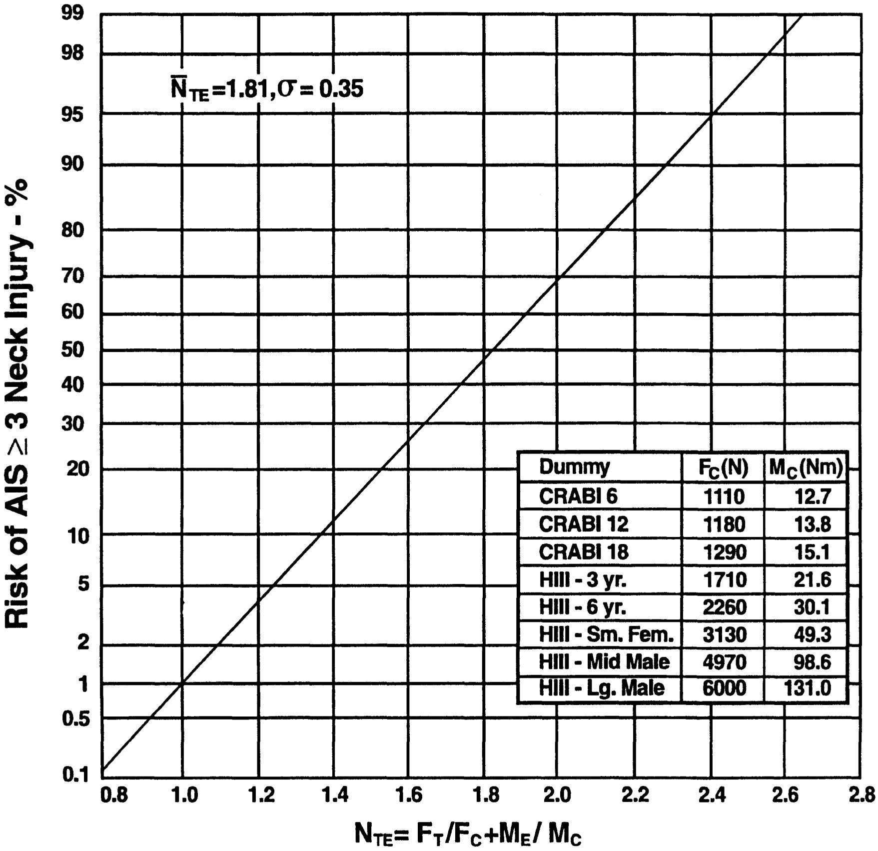


FIGURE 5.7. Risk of AIS ≥3 neck injury as a function of combined normalized neck extension moment taken with respect to the occipital condyles and neck tension for the CRABI and Hybrid III dummy families. Minimum muscle tone.

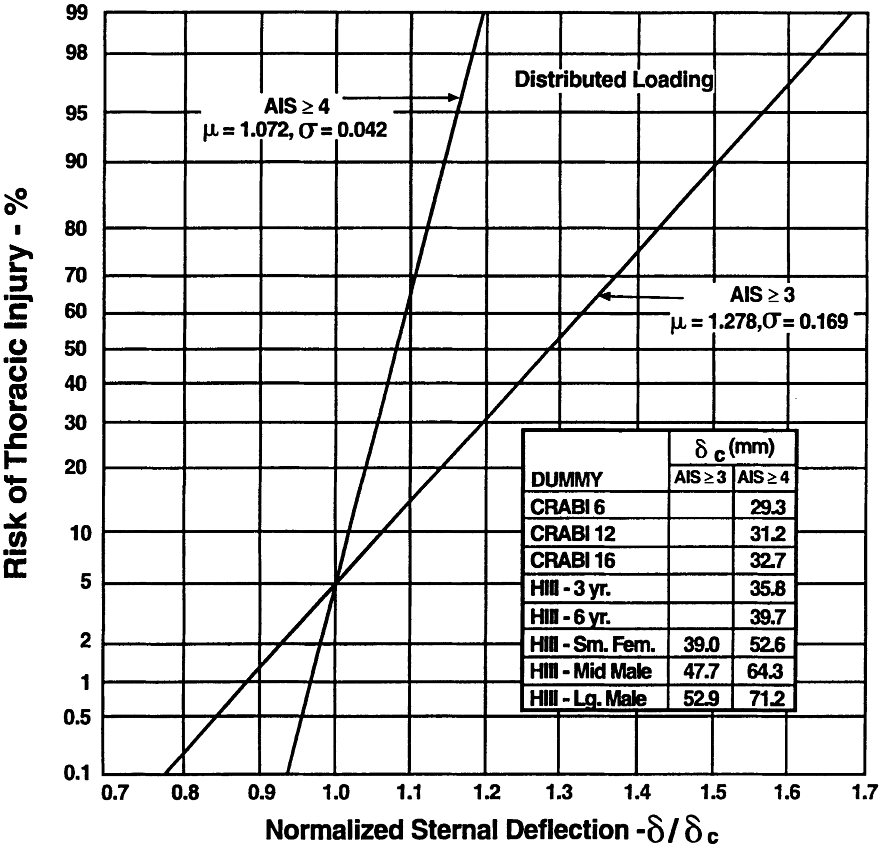
was estimated based on the maximum tensions and corresponding extension moments, or vice versa, that were recorded for each test. Second, the failure stress data are for the calcaneal tendon, not the anterior-longitudinal ligament. Third, the formula used to calculate the force in the anterior-longitudinal ligament needs to be justified. The Alliance and NHTSA have initiated a joint project to address these concerns.

Thoracic Injury Risk Curves

Thoracic injury risk curves have been proposed for peak sternal deflection, peak rate of sternal deflection, and the product of the two responses, which is called the viscous criterion.[8,11,12,19] Thoracic injuries associated with peak sternal deflection are rib and sternal fractures, which are rated as AIS ≥2, and thoracic organ damage produced by crushing forces, which is rated as AIS ≥4. Peak rate of sternal deflection is associated with AIS ≥4 thoracic organ injury, which is produced by inertial and viscous forces.

Sternal Deflection

Neathery et al[20] have summarized the thoracic impact data of various investigators who have subjected cadavers to distributed, anterior-to-posterior thoracic impacts. Mertz et al[12] analyzed these data and developed injury risk curves for AIS ≥3 and AIS ≥4 thoracic injuries due to distributed anterior to posterior thoracic



	δc (mm)	
DUMMY	AIS ≥ 3	AIS ≥ 4
CRABI 6		29.3
CRABI 12		31.2
CRABI 16		32.7
HIII - 3 yr.		35.8
HIII - 6 yr.		39.7
HIII - Sm. Fem.	39.0	52.6
HIII - Mid Male	47.7	64.3
HIII - Lg. Male	52.9	71.2

FIGURE 5.8. Risk of AIS ≥3 and AIS ≥4 thoracic injuries for CRABI and Hybrid III dummy families as a function of normalized sternal deflection due to distributed anterior thoracic impacts.

impacts. The curves were normalized by the sternal deflections corresponding to a 5% injury risk and are shown in Fig. 5.8. Tabulated on the graph are the sternal deflections for 5% injury risks for AIS ≥3 and AIS ≥4 thoracic injury for various sizes of dummies. Note that for the child and infant dummies, no values are given for AIS ≥3 injury. This is because rib fracture is the primary indicator of AIS = 3 injury. However, children will experience AIS ≥4 thoracic organ injury without rib fracture at large sternal deflections since the low elastic modulus of their ribs produces low rib stress at large sternal deflections. The injury risk curves of Fig. 5.8 are for distributed thoracic loads such as airbag loading.

Shoulder belt loading of the thorax produces a different deflection pattern of the rib cage than does an airbag. Mertz et al[21] conducted an accident reconstruction program where the shoulder belt loading that occurred in the field was duplicated with the Hybrid III midsize male dummy. They correlated the risks of thoracic injury seen in the accidents to the sternal deflections measured in the simulations. Fig. 5.9 is the injury risk curve that they proposed for AIS ≥3 thoracic injury due to shoulder belt loading for sternal deflection measured with the Hybrid III midsize male dummy.

Rate of Sternal Deflection

Mertz and Weber[8] developed a thoracic injury risk curve for the rate of sternal deflection for a 3-year-old child. Mertz et al[12] scaled the

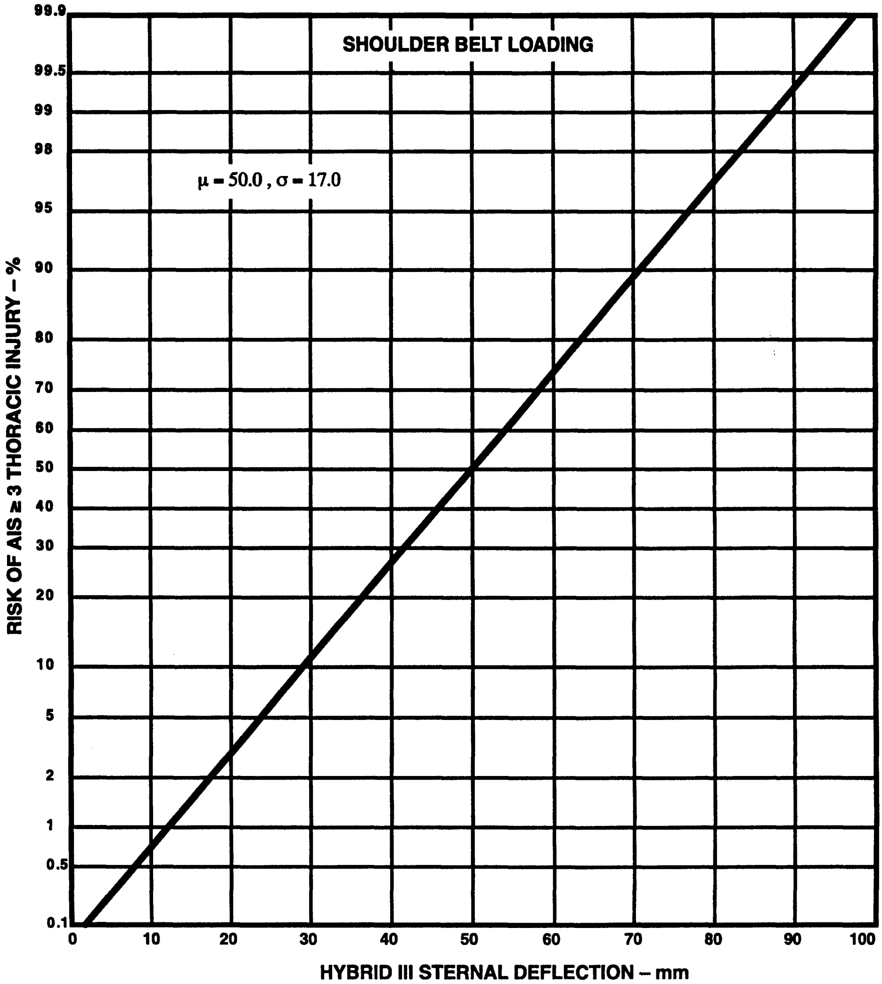


FIGURE 5.9. Risk of AIS ≥3 thoracic injury due to shoulder belt loading of the thorax as a function of sternal deflection for the Hybrid III midsize male dummy.

results to other dummy sizes, and the Alliance[11] incorporated the effect of differences in failure stress of soft organ tissue with age using the following relationship:

$$\lambda_v = \lambda_\sigma \lambda_E^{-1/2}$$

where

λ_v is the ratio of sternal deflection rates,
λ_σ is the ratio of tissue failure stresses,
λ_E is the ratio of elastic moduli.

Figure 5.10 is the thoracic injury risk curve for sternal compression rate for AIS ≥4 thoracic organ injury for various sizes of dummies. The curve has been normalized by the deflection rate for a 5% risk. The normalizing values for the various sizes of dummies are tabulated on the graph.

Viscous Criterion

The viscous criterion is the instantaneous product of the sternal deflection divided by thoracic depth times the deflection rate. Viano and Lau[19] developed an injury risk curve for the viscous criterion for AIS ≥4 thoracic injury (Fig. 5.11). A word of caution is needed about the efficacy of the viscous criterion risk curve.

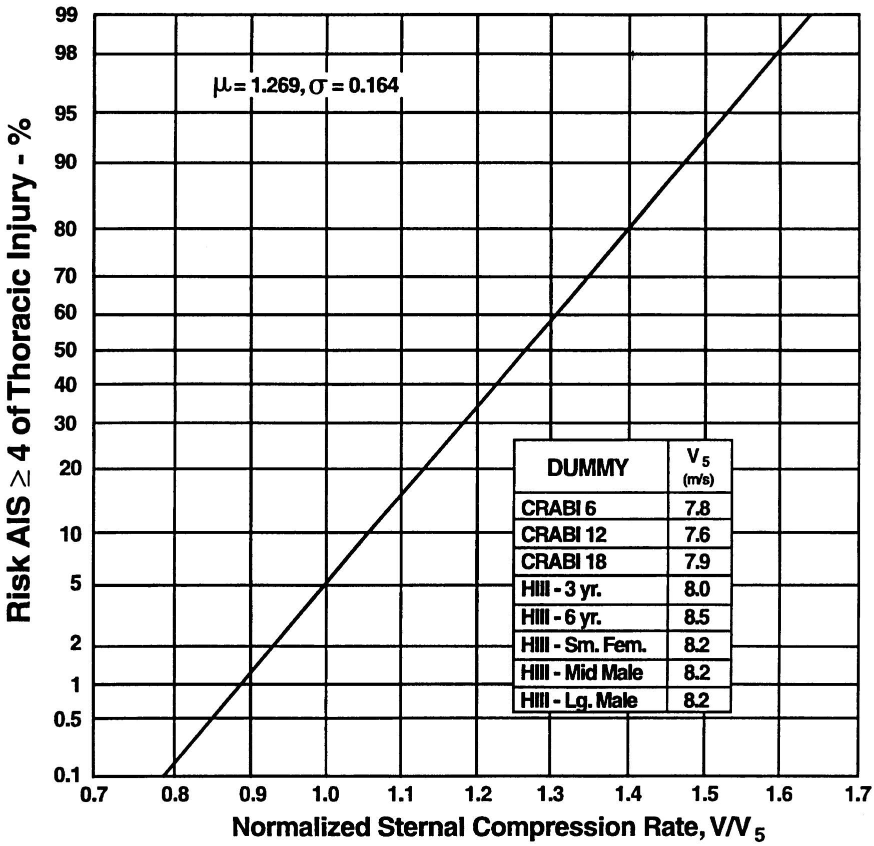


FIGURE 5.10. Risk of AIS ≥4 thoracic injury for CRABI and Hybrid III dummy families as function of normalized sternal deflection rate due to distributed anterior thoracic impacts.

The data used to generate the curve did not include high-frequency components because of instrumentation limits. However, it is the high-frequency components that produce the injuries that have been attributed to the viscous criterion.

Future Developments

In addition to the work being done to improve the efficacy of the combined neck tension and extension moment risk curve of Fig. 5.7, it would be appropriate to generate injury risk curves for other neck loading conditions as well as for lower extremity fractures. The main difficulty in generating such curves is the lack of appropriate biomechanical data. Most biomechanical experiments are conducted on a relatively small sample size of specimens that have widely different failure threshold levels. No tests are conducted at a specimen's threshold of failure level. The stimulus either exceeds or is less than the failure level by an unknown amount. Consequently, the biomechanical data are censored, which increases the difficulty of estimating the injury thresholds of the tested population and extrapolating the results to specific percentiles of the car occupant population.[8,12–15] Work needs to be done on developing testing protocols and analysis techniques that are amenable to the determination of injury threshold distributions of various sizes and ages of people based on small sample sizes of specimens that are not uniform in characteristics or representative in strength of the percentile of the car occupant population for which the injury risk curve is being developed.

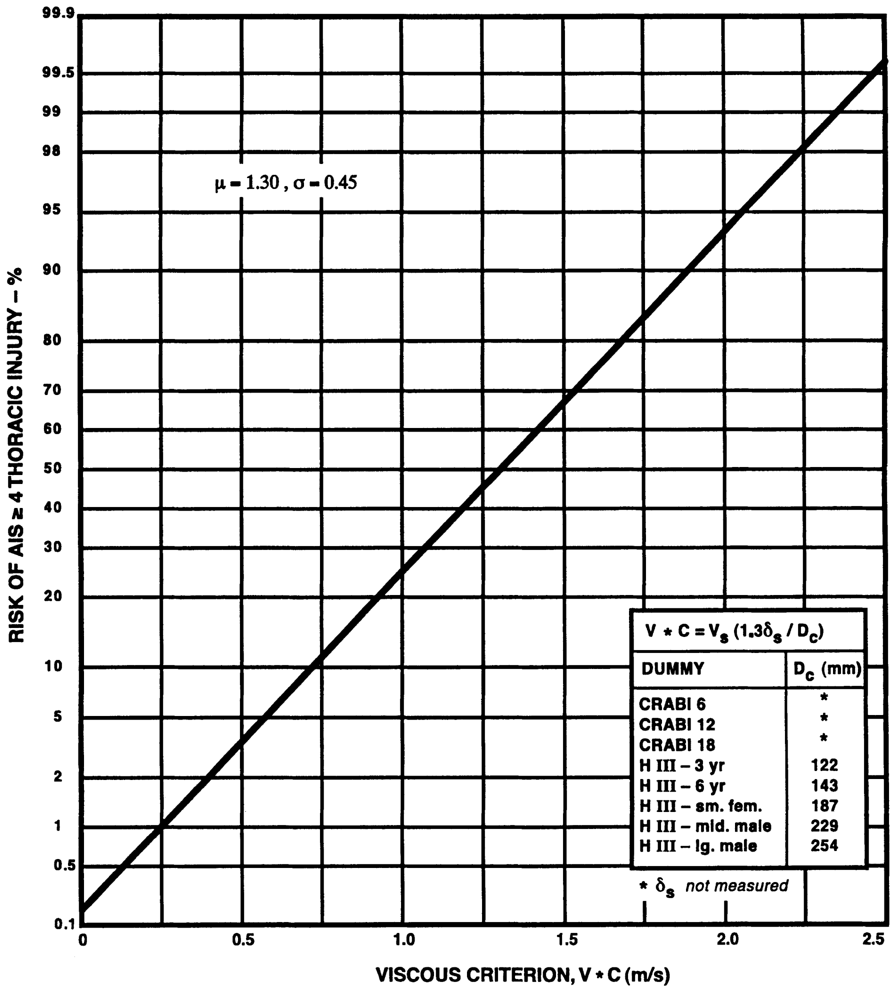


FIGURE 5.11. Risk of AIS ≥4 thoracic injury for adult dummies as a function of peak viscous criterion for distributed anterior thoracic impacts.

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