



Insurance Institute for  
Highway Safety



# **Moderate Overlap Frontal Crashworthiness Evaluation Guidelines for Rating Injury Measures**

**September 2014**

988 Dairy Road  
Ruckersville, VA 22968  
+1 434 985 4600

[iihs.org](http://iihs.org)

## **Moderate Overlap Frontal Crashworthiness Evaluation Guidelines for Rating Injury Measures**

### **Document Revision History**

The June 2009 revision of this document clarifies the femur and neck force duration corridors. The May 2014 revision includes an explanation of the tibia moment adjustment, which previously had been mentioned in the frontal offset crash test protocol. The September 2014 revision reinstated a flowchart that describes the head hard contact downgrading procedure (Figure 1).

### **Injury Rating**

Injury measures obtained from an instrumented 50th percentile male Hybrid III dummy in the driver seat are used to determine the likelihood that an occupant would have sustained significant injury to various body regions. Thirty-two different measures are recorded in each of the small overlap crash tests:

- Head acceleration and angular rate (three directions from head's center of gravity)
- Axial force, anterior-posterior force, lateral-medial force, and anterior-posterior bending moment acting at the connection between the dummy's head and neck
- Thoracic spine acceleration (three directions)
- Sternum compression
- Femur axial force (each leg)
- Tibia-femur displacement (each leg)
- Tibia transverse bending moments (upper and lower, each leg)
- Tibia axial force (each leg)
- Foot acceleration (two directions, each foot)

The 32 measures are grouped into four body regions: head and neck, chest, left leg and foot, and right leg and foot. Four injury parameters are used to evaluate protection for the head and neck, three parameters for the chest, and six parameters for each leg and foot.

Each body region receives an injury protection rating of good, acceptable, marginal, or poor based on the injury parameters for that region. For any body region to receive a good rating, the scores for all injury parameters in that region must indicate good results. If any parameter indicates an acceptable result, then the rating for that body region is acceptable. If any parameter has a marginal result, then the rating for that body region is marginal. Thus, the overall injury rating for any body region is the lowest rating scored for an injury parameter within that region.

Table 1 shows the injury parameter ranges associated with the possible ratings: good, acceptable, marginal, and poor. Injury results that round to the values shown in Table 1 will receive the better of the two ratings they separate. With some exceptions (e.g., chest acceleration), the borders between acceptable and marginal ratings for a given injury parameter correspond to published injury assessment reference values (IARV) for significant injury related to that parameter. Acceptable ratings correspond to measures somewhat below (better than) the IARVs, and good ratings correspond to measures well below the IARVs. Similarly, marginal ratings correspond to measures just above (worse than) the IARVs, and poor ratings correspond to measures well above the IARVs. It is important to remember in reading Table 1 that crashes are complex events. It is not possible for an injury-coding scheme to foresee all the

possible combinations of outcomes that could suggest injury risk in a crash. Therefore, the information in Table 1 should be interpreted as providing guidelines for evaluating dummy injury measures, which are always subject to modification based on the circumstances of the particular crash test and on new biomechanical information about injury tolerance. IARVs are based on current biomechanical information about human injury mechanisms and are described in more detail below.

**Table 1**  
**Injury Parameter Cutoff Values Associated with Possible Injury Protection Ratings**

Body Region	Parameter	IARV	Good – Acceptable	Acceptable – Marginal	Marginal – Poor
Head and neck	HIC-15	700	560	700	840
	N <sub>ij</sub>	1.00	0.80	1.00	1.20
	Neck axial tension (kN)*	3.3	2.6	3.3	4.0
	Neck compression (kN)*	4.0	3.2	4.0	4.8
Chest	Thoracic spine acceleration (3 ms clip, g)	60	60	75	90
	Sternum deflection (mm)	-50	-50	-60	-75
	Sternum deflection rate (m/s)	-8.2	-6.6	-8.2	-9.8
	Viscous criterion (m/s)	1.0	0.8	1.0	1.2
Leg and foot, left and right	Femur axial force (kN)**	-9.1	-7.3	-9.1	-10.9
	Tibia-femur displacement (mm)	-15	-12	-15	-18
	Tibia index (upper, lower)	1.00	0.80	1.00	1.20
	Tibia axial force (kN)	-8.0	-4.0	-6.0	-8.0
	Foot acceleration (g)	150	150	200	260

\* Neck axial force duration corridors are shown in Figures 2 and 3.

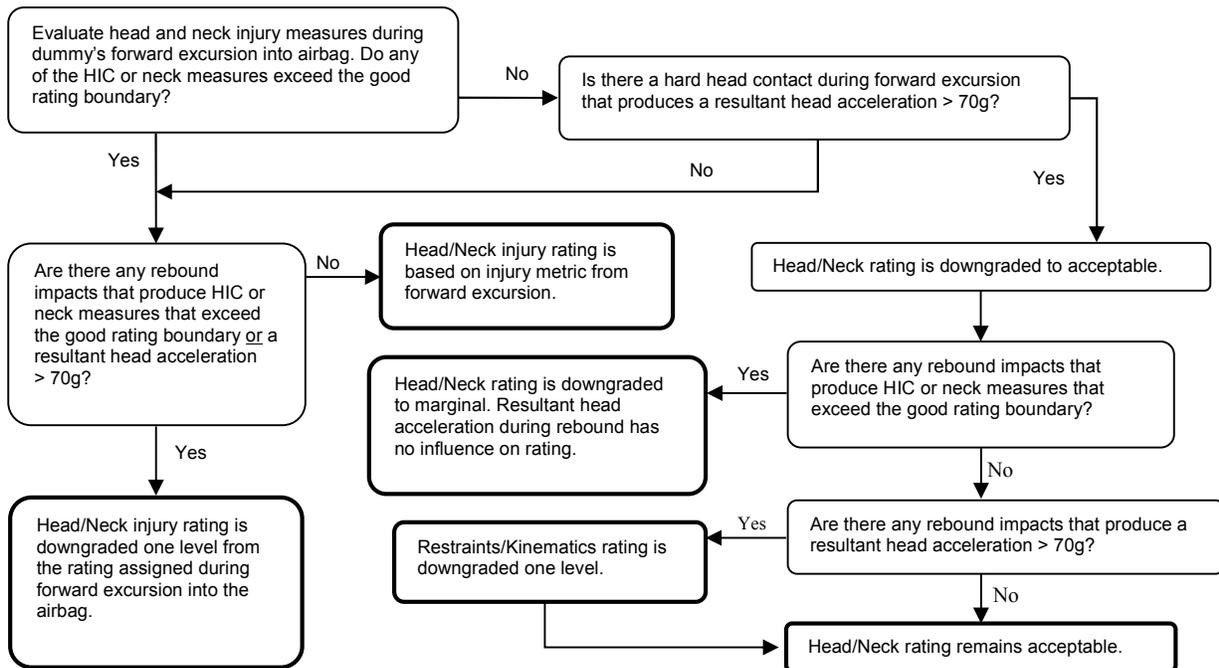
\*\* Femur axial force duration corridors are shown in Figure 5.

## Head and Neck

Head injury risk is evaluated mainly on the basis of head injury criterion (HIC) with a 15 ms limit on the period over which it is calculated. A value of 700, which is the maximum allowed under the provisions of the U.S. advanced airbag regulation (National Highway Traffic Safety Administration (NHTSA), 2000), marks the border between an Institute rating of acceptable and marginal. A HIC-15 of 700 is estimated to represent a 5 percent risk of a severe injury (Mertz et al., 1997). A “severe” injury is one with a score of 4+ on the Abbreviated Injury Scale (AIS) (Association for the Advancement of Automotive Medicine, 1990).

In addition to HIC-15, the maximum vector resultant acceleration of the head is considered. A maximum head acceleration that exceeds 70 g and is caused by contact between the head and a hard surface of the vehicle interior can result in lowering the head injury rating one level (details are provided in Figure 1). This additional criterion for evaluating protection of the head is based on concern about moderate and serious injuries (AIS 2-3) to the face and weaker parts of the skull than are represented by the biomechanical data underlying the establishment of HIC. This is a particular concern when the high accelerations are caused by head impacts against hard structures with relatively small areas of contact, such as may occur when the back of a dummy’s head hits the B-pillar or the face of the dummy contacts the steering wheel through the deflating airbag. Minimum force levels associated with face and skull fractures caused by blows with a small impactor (6.45 cm<sup>2</sup> area) range from 670 to 4000 N (Schneider and Nahum, 1972). These forces correspond to head accelerations of 15-90 g for the 4.5 kg head of the Hybrid III dummy used in Institute tests. This additional head evaluation criterion is similar to requirements of the Canadian Motor Vehicle Safety Regulations Standard 208, which limits test dummy head accelerations to less than 80 g in cars not equipped with airbags.

**Figure 1**  
**Flow Chart: Influence of Rebound Impacts on Head/Neck Injury Rating**



Neck injury risk is evaluated primarily on the basis of  $N_{ij}$ , neck tension force, and neck compression force, which are the same parameters used to evaluate compliance with the U.S. advanced airbag rule (NHTSA, 2000). The  $N_{ij}$  is a linear combination of neck axial force ( $F_z$ ) and the bending moment about a lateral axis passing through the dummy's occipital condyle ( $M_y$ ), as shown in Equation 1. The critical values against which the force and moment are compared depend on the direction of the applied loads, as shown in Table 2. An  $N_{ij}$  value of 1.0 marks the border between an Institute rating of acceptable and marginal. According to a risk analysis published by NHTSA (Eppinger et al., 1999), an  $N_{TE}$  (neck tension-extension) of 1.0 was estimated to represent a 30 percent risk of AIS 2 neck injury, a 22 percent risk of AIS 3 injury, an 18 percent risk of AIS 4 injury, and a 7 percent risk of an AIS 5 injury. Risk assessments for the other load configurations ( $N_{TF}$ ,  $N_{CE}$ ,  $N_{CF}$ ) are not available because the data set on which  $N_{ij}$  is based does not include them. Another analysis of the neck injury risk associated with an  $N_{TE}$  value of 1.0 suggests a much lower risk of AIS 3+ neck injuries for subjects tensing their neck muscles in anticipation of a crash. According to Mertz and Prasad (2000), this risk would be less than 1 percent. The actual risk of AIS 3+ neck injuries for a human the same shape and size as the Hybrid III dummy used in Institute tests and subjected to similar loads is probably between these estimates.

$$N_{ij} = (F_z / F_{critical}) + (M_y / M_{critical}) \quad (1)$$

**Table 2**  
**Critical Values for  $N_{ij}$  Calculation**

	Tension (+ $F_z$ ) / Flexion (+ $M_y$ )	Compression (- $F_z$ ) / Extension (- $M_y$ )
Neck axial force ( $F_z$ )	6806	6160
Front/back bending moment ( $M_y$ )	310	135

In addition to  $N_{ij}$ , neck axial force by itself is compared with the cutoff values shown in Table 1. Axial forces of 4.0 and 3.3 kN mark the borders between Institute ratings of acceptable and marginal for compression and tension, respectively. The compression limit for an acceptable rating is the same as the highest compression forces allowed under the provisions of the U.S. advanced airbag regulation (NHTSA, 2000), but the tension limit for an acceptable rating is somewhat lower than the highest forces allowed by Federal Motor Vehicle Safety Standard (FMVSS) 208. The lower limit for acceptable tension loading was chosen because it is consistent with earlier versions of the injury rating protocol, which were based on evaluation of neck loads by Mertz (1984). The neck tension reference value in that report is based on tests with the Hybrid III dummy that simulated real crashes (Nyquist et al., 1980). This lower limit for an acceptable tension force rating also is consistent with the 3 percent AIS 3+ injury risk for an unaware subject, according to Mertz and Prasad (2000). In contrast, the FMVSS 208 limit assumes the subject is resisting the crash forces by tensing the neck muscles. Although tensing of the neck muscles can effectively increase a subject's tolerance to neck injuries, the amount of tensing present in real-world crash-involved occupants is not known, so the more conservative limit is used here.

Finally, a head/neck rating that is otherwise good will be lowered to acceptable if the neck tension, compression, or shear (X-direction) forces fall outside the force duration corridors specified by Mertz (1984). The force duration corridor limits are shown in Figures 2-4.

The head and neck injury rating is based primarily on injury measures recorded during the dummy's forward excursion into the airbag or other contacts during forward excursion (e.g., with the instrument panel, A-pillar or window sill); however, high head or neck injury measures recorded during a rebound impact can downgrade the overall head/neck injury or the restraints and kinematics rating. The flow chart in Figure 1 describes the downgrading procedure applied to hard head contacts.

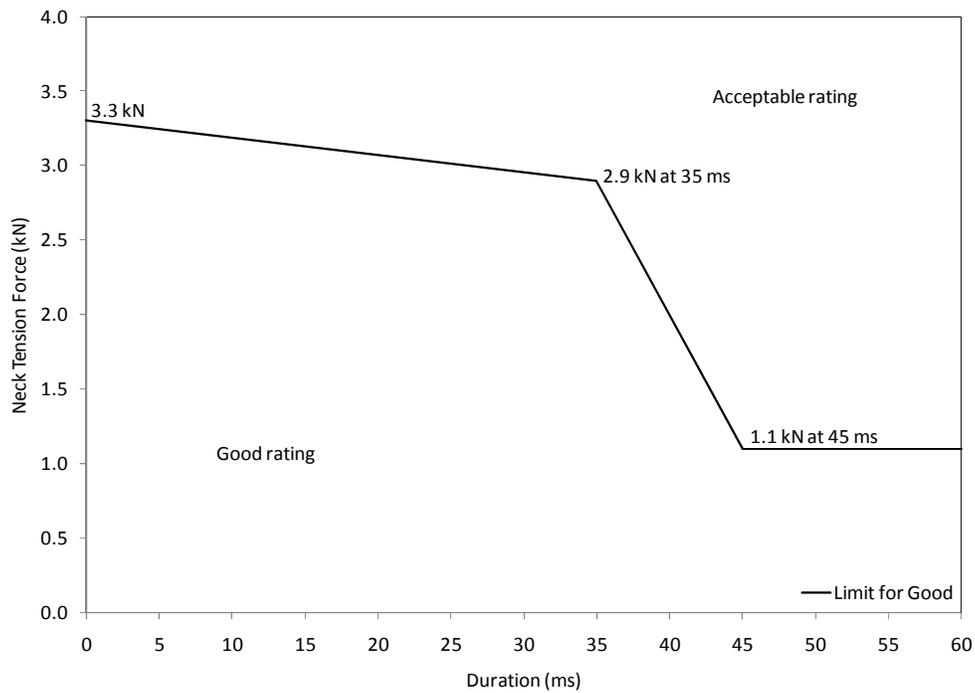
## **Chest**

Chest injury risk is evaluated on the basis of sternum deflection, sternum deflection rate, viscous criterion, and thoracic spine acceleration. A sternum deflection of 60 mm marks the border between an Institute rating of acceptable and marginal. This is near the same limit used to evaluate compliance with the U.S. advanced airbag rule (NHTSA, 2000). According to Mertz et al. (1997), it represents either a 45 or 70 percent risk of an AIS 3+ chest injury depending on whether the airbag or seat belt causes the chest deformation. The higher injury risk for seat belt loading is in approximate agreement with earlier risk estimates based on re-creations of real-world crashes using a Hybrid III dummy (Mertz et al., 1991).

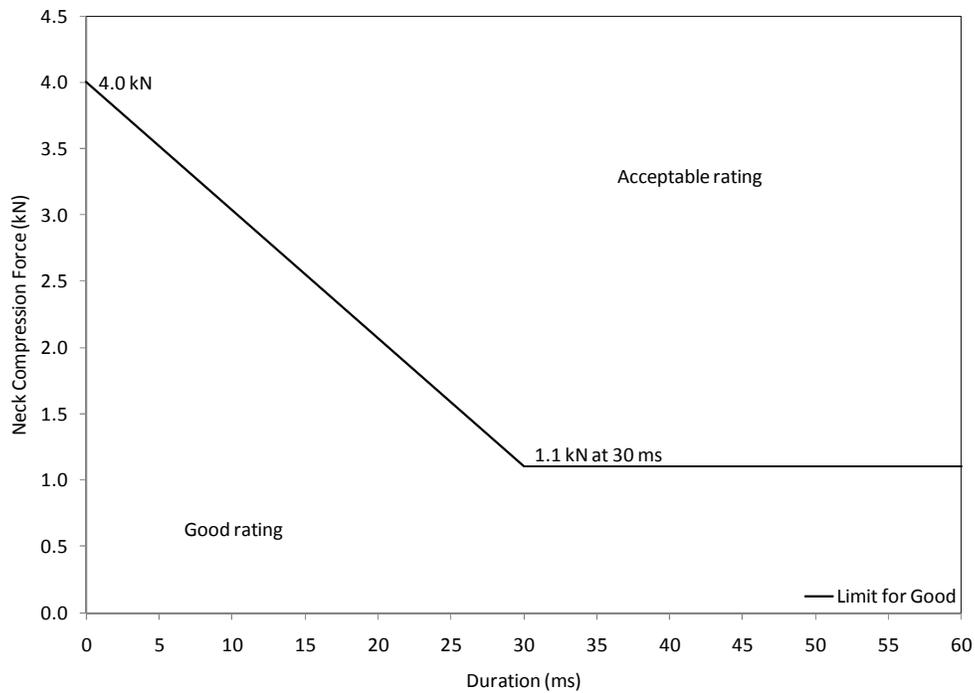
A sternum deflection rate of 8.2 m/s marks the border between an Institute rating of acceptable and marginal. The Alliance of Automobile Manufacturers (1999) recommended to NHTSA that this limit be used to evaluate compliance with the U.S. advanced airbag rule (NHTSA, 2000), but NHTSA rejected the recommendation pending the availability of more biomechanical data to justify it. According to Mertz et al. (1997), this level of sternum deflection rate is estimated to represent a 65 percent risk of AIS 3+ chest injury or about a 15 percent risk of AIS 4+ chest injury.

Another rate-dependant injury criterion, viscous criterion, also is calculated from sternum deflection measurements. Viscous criterion is the product of sternum deflection, normalized by chest depth, and the sternum deflection rate. According to Lau and Viano (1986), a viscous criterion value of 1 m/s represents a 25 percent risk of AIS 4+ thoracic injury.

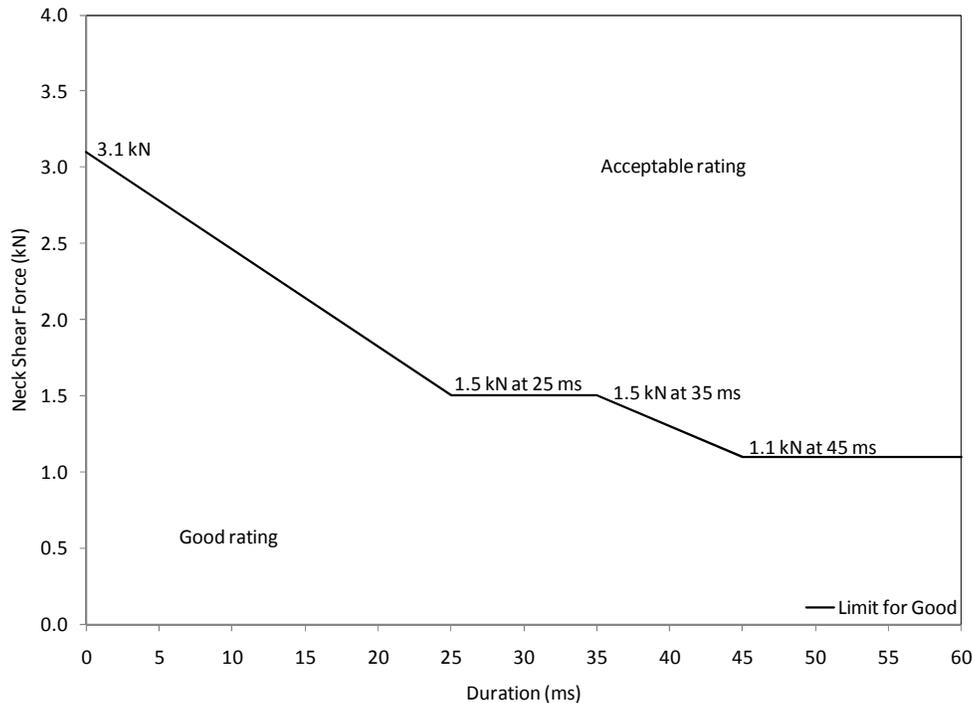
**Figure 2**  
**Force Duration Corridor for Neck Tension Force**



**Figure 3**  
**Force Duration Corridor for Neck Compression Force**



**Figure 4**  
**Force Duration Corridor for Neck Shear Force**

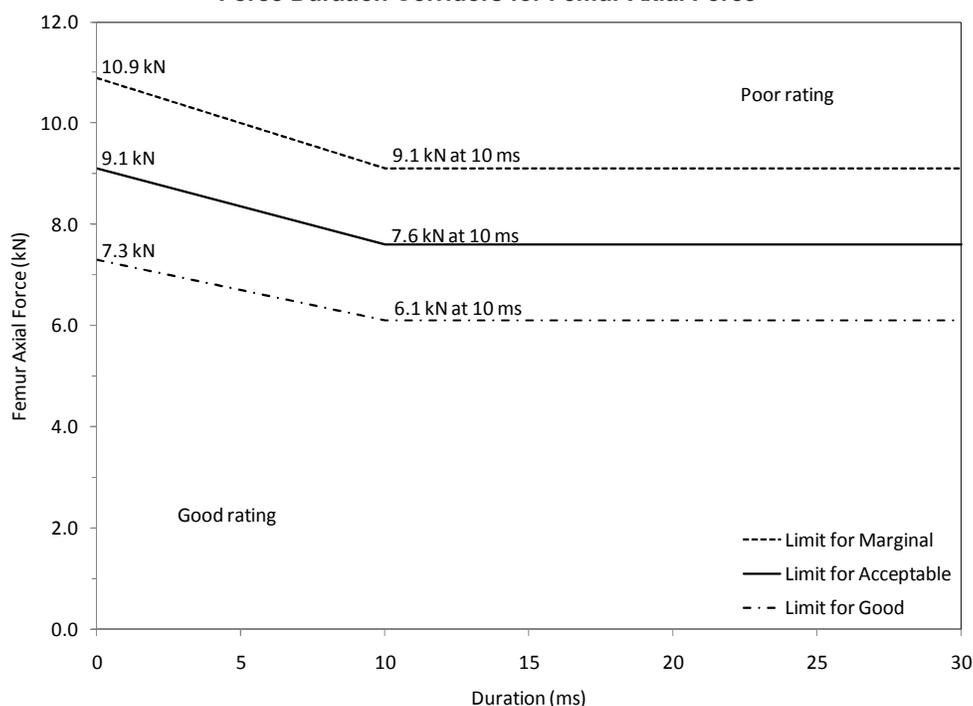


A thoracic spine acceleration of 60 g (3 ms) marks the border between an Institute rating of good and acceptable. This value also is used to evaluate compliance with the U.S. advanced airbag rule (NHTSA, 2000). According to NHTSA (2001), this represents a 20 percent risk of AIS 4+ chest injury. However, the value of whole-body thorax acceleration as a predictor of injury was highly debated through the development of the advanced airbag rule because the injuries observed among crash-involved occupants generally are associated with rib cage deformation. Mertz and Gadd (1971) reviewed the biomechanics literature and conducted experiments with a volunteer to better ascertain the survivable levels of whole-body acceleration. They noted cases of people surviving falls from as high as 30 m (100 ft) and estimated average chest accelerations of 20-200 g for such falls. One experiment involved a chest deceleration with a duration of 90 ms and a peak of 83 g, which was described as the definite limit of tolerance and caused minor injuries. Their own experiments consisted of measurements made on the body of a stunt performer who dove from heights of 8-17 m (27-57 ft) onto cushions of low-density polyurethane foam. The highest dive, which caused no discomfort to the performer, produced a chest deceleration with a duration of 100 ms and a peak of 49 g. They recommended a tolerance level of 60 g for pulse durations lasting about 100 ms on the basis of this work. These arguments suggest that the meaningfulness of chest acceleration is limited, especially given the presence of other measures that were directly related to the injury mechanism. Nevertheless, including chest acceleration in the evaluation of chest injury risk does not pose an unreasonable hurdle to achieving a good chest rating with the cutoff values shown in Table 1.

### Legs and Feet

Leg and foot injury risk is evaluated on the basis of femur axial force, tibia-knee displacement, tibia indices measured at the upper and lower portions of the tibia, tibia axial force measured at the distal end of the tibia, and foot acceleration. The femur axial force cutoff value for an acceptable rating is somewhat lower than the limit allowed by FMVSS 208 but is the same as proposed by Mertz (1984). In addition to comparing the maximum femur axial force with the cutoff values in Table 1, force duration is compared with the corridors described by Mertz. The force duration corridor limits are shown in Figure 5.

**Figure 5**  
**Force Duration Corridors for Femur Axial Force**



A tibia-femur displacement of 15 mm marks the border between an Institute rating of acceptable and marginal. This is the reference value recommended by Mertz (1984) and based on work by Viano et al. (1978). Similarly, a tibia index of 1.0 is the cutoff value between an acceptable and marginal rating. Tibia indices are calculated using adjusted bending moments.

The Hybrid III lower leg geometry allows pure compressive loads on the tibia to result in the tibia bowing forward (negative Y moment per SAE J1733 sign convention). This forward upper tibia bending in many cases can be shown to be only the product of the axial force acting on the moment arm between the upper tibia load cell and the line of force through the ankle and knee pivots, and not due to external forces bending the tibia. The upper tibia A-P moment (and to a much lesser extent, the lower tibia A-P moment) are modified mathematically to correct the problem. The mathematical adjustment is based on the tibia axial force, the known moment arms between the tibia load cells and the line of force through the ankle and knee pivots, and the known angle between the tibia shaft and the actual line of force. The adjusted upper tibia moment is calculated using Equation 2, and the adjusted lower tibia moment is calculated using Equation 3. Note that these equations are based on the SAE J1733 sign convention, with moments in Newton-meters (Nm) and axial force in Newtons (N). The lower tibia moments change very little because the line of force between the ankle and knee is very close to the center of the lower tibia load cell (Wellbourne and Shewchenko, 1998; Zuby et al., 2001).

$$M_{Y \text{ adjusted}} = M_{Y \text{ measured}} - F_{Z \text{ measured}} (0.02832) \quad (2)$$

$$M_{Y \text{ adjusted}} = M_{Y \text{ measured}} + F_{Z \text{ measured}} (0.006398) \quad (3)$$

The acceptable-marginal cutoff value for tibia axial force is somewhat lower than the reference value recommended by Mertz (1984) because Crandall et al. (1998) have shown that heel fractures (AIS 2, but associated with high degree of impairment) occur at considerably lower forces.

Zeidler (1984) suggested the conservative limit of 150 g for foot acceleration based on tests with volunteers and dummies. This level of acceleration is associated with jumps from a height beyond which injury was feared. Consequently, it marks the limit allowed for a good rating, whereas only much higher accelerations result in marginal or poor leg/foot ratings.

## References

Alliance of Automobile Manufacturers. 1999. Comments to the National Highway Traffic Safety Administration on Supplemental Notice of Proposed Rulemaking, FMVSS 208, Occupant Crash Protection – Annex 2: Dummy Response Limits for FMVSS 208 Compliance Testing. Docket No. NHTSA 99-6407, Notice 1; DMS Document No. NHTSA-1999-6407-40, Dec. 23, 1999. Washington, DC.

Association for the Advancement of Automotive Medicine. 1990. *The Abbreviated Injury Scale, 1990 Revision*. Des Plaines, IL.

Crandall, J.R.; Kuppa, S.M.; Klopp, G.S.; Hall, G.W.; Pilkey, W.D.; and Hurwitz, S.R. 1998. Injury mechanisms and criteria for the human foot and ankle under axial impacts to the foot. *International Journal of Crashworthiness* 3:147-61.

Eppinger, R.; Sun, E.; Bandak, F.; Haffner, M.; Khaewpong, N.; Maltese, M.; et al. 1999. Development of improved injury criteria for the assessment of advanced automotive restraint systems - II. Report from the National Highway Traffic Safety Administration concerning Federal Motor Vehicle Safety Standard 208, Occupant Crash Protection. Docket Document No. 1999-6407-5, November 1, 1999. Washington, DC: U.S. Department of Transportation.

Lau, I.V. and Viano, D.C. 1986. The viscous criterion – bases and application of an injury severity index for soft tissues (SAE 861882). *Proceedings of the 30th Stapp Car Crash Conference* (P-189), 123-42. Warrendale, PA: Society of Automotive Engineers.

Mertz, H.J. 1984. Injury assessment values used to evaluate Hybrid III response measurements. Comment to the National Highway Traffic Safety Administration concerning Federal Motor Vehicle Safety Standard 208, Occupant Crash Protection. Enclosure 2 of Attachment 1 of Part III of General Motors Submission USG 2284; Docket Document No. 74-14-N32-1666B, March 24, 1984. Washington, DC: U.S. Department of Transportation.

Mertz, H.J. and Gadd, C.W. 1971. Thoracic tolerance to whole-body deceleration (SAE 710852). *Proceedings of the 15th Stapp Car Crash Conference*, 135-55. New York, NY: Society of Automotive Engineers.

Mertz, H.J.; Horsch, J.; Horn, G.; and Lowne, R. 1991. Hybrid III sternal deflection association with thoracic injury severities of occupants restrained with force-limiting shoulder belts (SAE 910812). *Frontal Crash Safety Technologies for the 90s* (SP-812), 105-19. Warrendale, PA: Society of Automotive Engineers.

Mertz, H.J. and Prasad, P. 2000. Improved neck injury risk curves for tension and extension moment measurements of crash test dummies (SAE 2000-01-SC05). *Stapp Car Crash Journal* 44:59-76. Warrendale, PA: Society of Automotive Engineers.

Mertz, H.J.; Prasad, P.; and Irwin, N.L. 1997. Injury risk curves for children and adults in frontal and rear collisions (SAE 973318). *Proceedings of the 41st Stapp Car Crash Conference* (P-315), 13-30. Warrendale, PA: Society of Automotive Engineers.

National Highway Traffic Safety Administration. 2000. Title 49 Code of Federal Regulations (CFR) Part 571 Section 208, Occupant Crash Protection. Washington, DC: Office of the Federal Register, National Archives and Records Administration.

National Highway Traffic Safety Administration. 2001. NCAP testing results: buying a safer car, crash testing and safety features information. Washington, DC. Available at: <http://www.nhtsa.dot.gov/ncap/infopage.html#crash>.

Nyquist, G.W.; Begman, P.C.; King, A.I.; and Mertz, H.J. 1980. Correlation of field injuries and GM Hybrid III dummy response for lap-shoulder belt restraint. *Journal of Biomechanical Engineering* 102:103-09.

Schneider, D.C. and Nahum, A.M. 1972. Impact studies of facial bones and skull (SAE 720965). *Proceedings of the 16th Stapp Car Crash Conference*, 186-203. New York, NY: Society of Automotive Engineers.

Viano, D.C.; Culver, C.C.; and Haut, R.C. 1978. Bolster Impacts to the knee and tibia of human cadavers and an anthropometric dummy (SAE 780896). *Proceedings of the 22nd Stapp Car Crash Conference* (P-77), 401-28. Warrendale, PA: Society of Automotive Engineers.

Welbourne, E.R. and Schewchenko, N. 1998. Improved measures of foot ankle injury risk from the Hybrid III tibia. *Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles*, 1618-26. Washington, DC: National Highway Traffic Safety Administration.

Zeidler, F. 1984. The significance of lower limb injuries of belted drivers. *Journal of Orthopedics* [German].

Zuby, D.S.; Nolan, J.S.; and Sherwood, C.P. 2001. Effect of Hybrid III geometry on upper tibia bending moments (SAE 2001-01-0169). *Biomechanics Research and Development* (SP-1577), 1-14. Warrendale, PA: Society of Automotive Engineers.