Chapter 3 – INJURY CRITERIA AND TOLERANCE LEVELS

This chapter starts with a general introduction on injury biomechanics (see Section 3.1). Information on vehicle mine incidents and their injuries can be found in Annex B and C. In the research area Vehicle Mine Protection (VMP) specific body regions have been identified as the most critical ones for injuries caused by local and global effects (see Chapter 2). As mentioned in the previous chapters, it is assumed that the fragments and detonation products are not a threat to the occupants since the preservation of the integrity of the vehicle is a mandatory requirement. However, the overpressure still can affect primarily the auditory and non-auditory internal organs. Therefore these organs are discussed in this chapter as well.

The Sections 3.2 to 3.6 provide information on the different injury mechanisms and injury criteria for these specific body regions:

- Lower leg (see Section 3.2);
- Thoraco-lumbar spine (see Section 3.3);
- Neck (see Section 3.4);
- Head (see Section 3.5); and
- Internal organs (see Section 3.6).

This chapter concludes with a summary in Section 3.7.

3.1 INJURY BIOMECHANICS

Physical injury will take place if the biomechanical response is of such a nature that the biological system deforms beyond a tolerable limit resulting in damage to anatomical structures and/or alteration in normal function [Wismans, 1994]. The mechanism involved is called injury mechanism.

The following principal **injury mechanisms** are usually distinguished in impact (see Figure 3.1):

- **Elastic injury mechanisms**: Compression and tension of the body causing injury if elastic tolerances are exceeded. Injury can occur in case of slow deformation of the body (crushing) or in the case of high velocity impacts.
- **Viscous injury mechanisms**: Impulsive type of loading causing mechanical waves in the body, which results in internal injuries if so-called viscous tolerances are exceeded.
- **Inertial injury mechanisms**: Acceleration type of loading causing tearing of internal structures due to inertia effects.

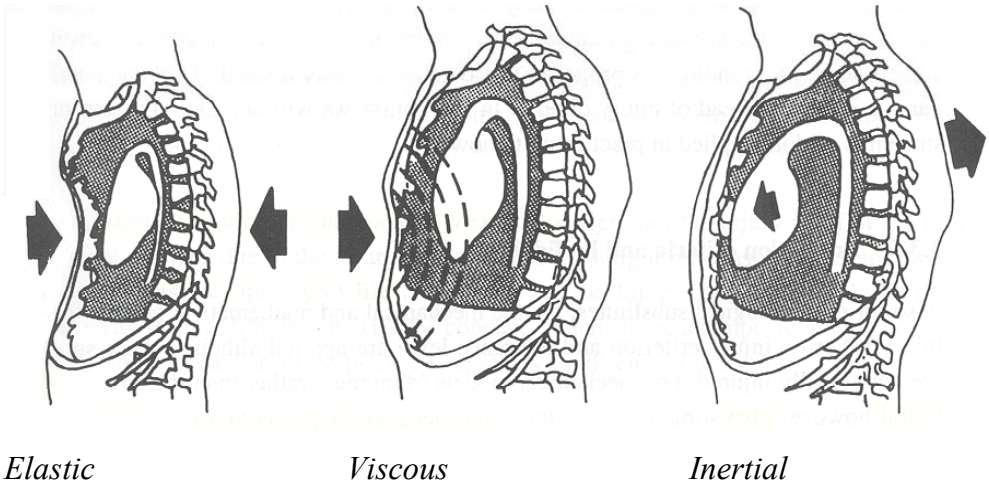


Figure 3.1: Three Principle Mechanisms [Wismans, 1994].

An **injury criterion** is defined as a physical parameter or a function of several physical parameters which correlates well with the injury severity of the body region under consideration. Frequently used parameters are those quantities that relatively easy can be determined in tests with human substitutes like the linear acceleration experienced by a body part, global forces or moments acting on the body or deflection of structures [Wismans, 1994].

Injury risk curves are used to define the injury risk for a given human body response. Examples of risk curves are shown in Figure 3.2. Along the vertical axis the injury risk is depicted, while along the horizontal axis the injury criterion is presented. The boundary conditions for which the injury criteria or models were developed are very important and will give more information about the validity of the risk curves. Sometimes, anthropomorphic factors such as age and gender, are included in the risk curves, however, the data set is usually too limited to differentiate for these factors. It should be noted that depending on the shape (which is related to the mathematical method used) of the risk curve, a small difference for the injury criterion value could result in large differences for the injury risk.

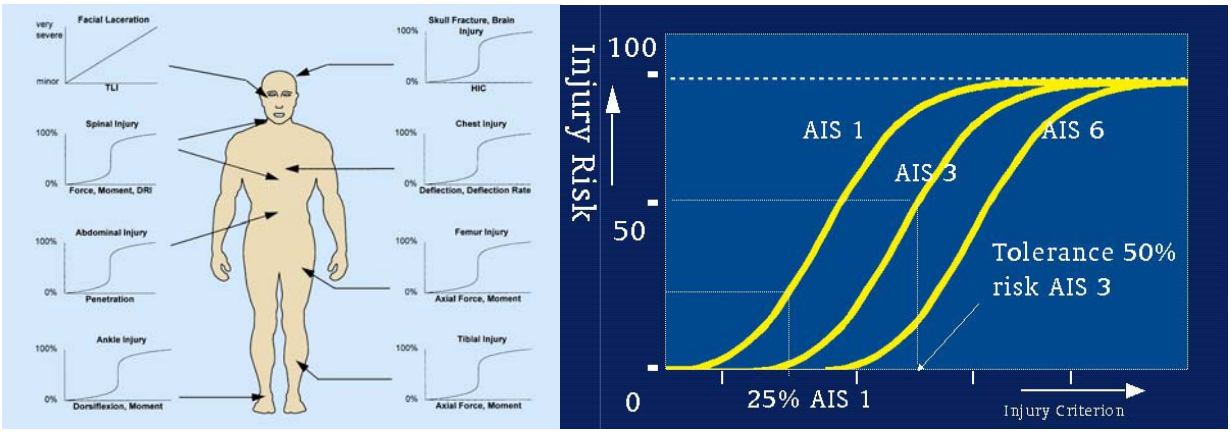


Figure 3.2: Examples of Injury Risk Curves for Different Body Parts (left) and Different Severities (right).

The term **tolerance level (**or **injury criterion level)** is defined as the magnitude of loading indicated by the threshold of the injury criterion, which produces a specific type of injury severity and risk.

Injury risk curves are mostly based on data from real case studies and experiments using animals, Post Mortem Human Surrogates (PMHS) or human volunteers. The risk curves are usually defined per body region and for specific injury severities (see Figure 3.2).

The injury severity can be defined using **injury scaling** which is defined as the numerical classification of the type and severity of an injury. The most well known anatomical scale, which is accepted world wide, is the Abbreviated Injury Scale (AIS) (see Table 3.1 and Annex C).

Table 3.1: The Abbreviate Injury Scale [AIS, 1990]

AIS Code	Injury Description
1	Minor
2	Moderate
3	Serious
4	Severe
5	Critical
6	Maximum (currently untreatable)
9	Unknown

The relation between threat-load-injury is presented in the TLI model by van der Horst [Horst, 2005] (see Figure 3.3). In case an injury risk curve is available, it can be used for injury assessment. This means that with information of the human body response it can be determined what the injury risk is for a specific load on that body part (Figure 3.3). Furthermore, an injury risk curve can be used to define the tolerance levels for a specific criterion (see Figure 3.4). The HFM-090/TG-25 used this method to define the tolerance levels needed for the pass/fail limits to qualify the protection level according to STANAG 4569. Based on discussions with the members of STANAG 4569 it has been decided that a 10% risk of AIS 2+ (AIS 2 or more) injuries will be accepted as pass/fail criterion for the AV mine strike tests. It should be noted that when accepting a 10% risk of AIS 2+, still a lower risk of higher severity (AIS 3+) injuries can be expected. On the other hand, this level implies a risk of more than 10% of AIS 1 injuries.

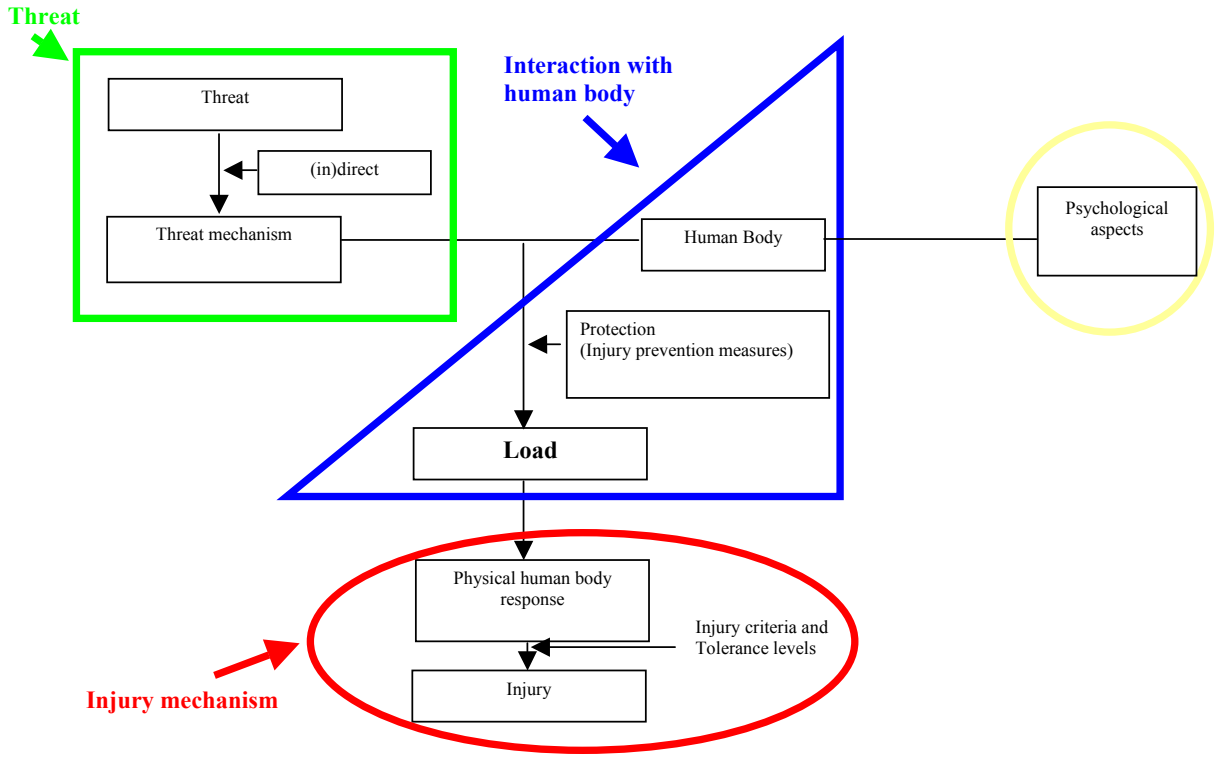


Figure 3.3: The Threat Load Injury (TLI) Model [Horst, 2005].

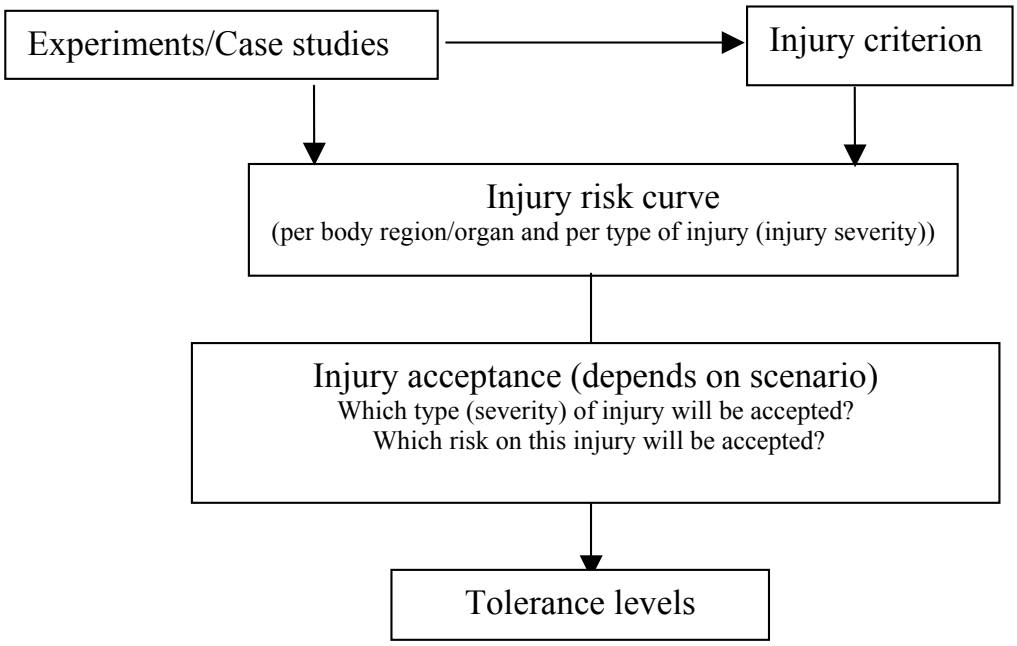


Figure 3.4: Flow Scheme to Determine Injury Tolerance Levels for Body Regions.

Standard injury criteria for the military field of application are usually not available. That is why STANAG 4569 asked HFM-090/TG-25 to define injury criteria and tolerance levels for the vehicle mine protection field. In general, the loading conditions for the occupant during a mine strike under a vehicle

are quite similar to vertical loads of pilots in an ejection seat in the aerospace, and to the loading of the human leg due to foot well intrusion in case of a frontal car impact, although higher amplitudes and shorter durations have to be expected. Criteria available from automotive and aerospace industry research areas were used by HFM-090/TG-25 as starting point to define injury criteria for the vehicle mine protection field.

3.2 LOWER LEG

This section provides information on lower leg anatomy and injuries, and on the injury risk model to be used when performing AV mine protection testing. Additional information related to lower leg injury assessment is provided in Annex E.

3.2.1 Anatomy, Loading Mechanisms and Injuries

The lower limb (or lower extremity) is divided in four regions: the thigh, the knee, the leg and the foot/ankle complex. The femur is the thigh-bone, the patella is the knee bone, the tibia and the fibula are the leg bones, and the phalanges, the metatarsals and the tarsals are the bones of the foot/ankle complex (see Figure 3.5). The tibia is the larger of the two leg bones and bears most of the body weight. In this report, the term lower leg is used to designate the leg and the foot/ankle complex.

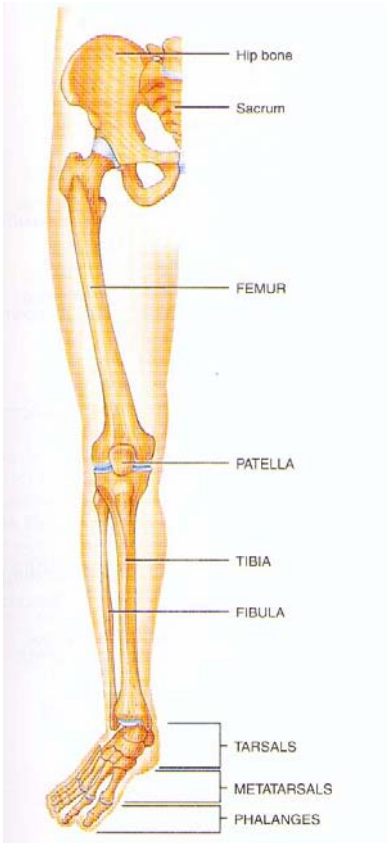


Figure 3.5: The Lower Limb [Tortora, 2003].

Figure 3.6 shows the medial view of the right foot. The tarsal bones (in the ankle) are the calcaneus, the talus, the cuboid, the navicular and the three cuneiforms. The talus, which is the only bone to articulate with the tibia and the fibula, transmits all the forces from the foot to the leg.

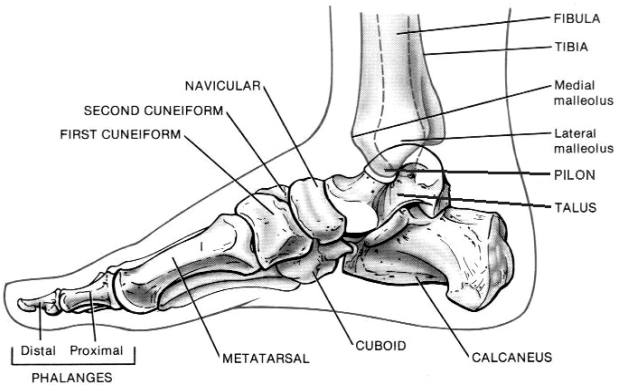


Figure 3.6: Foot/Ankle Complex Anatomy (Modified from [Tortora, 1984]).

When a landmine detonates under a vehicle, the movement of the structure on which the foot rests (floor, driver pedal, etc.) may cause an impact on the foot resulting in injuries to the lower leg. For example, when the foot is in direct contact with the bottom plate, its dynamic local deformation induces an important axial load to the lower leg, which may result in fractures in the foot/ankle complex. Tibia and fibula fractures and especially calcaneus fracture may also occur when the lower leg is subjected to an axial impact [Funk, 2002], but foot/ankle fractures will be predominant. Lower leg fractures have a score of AIS 2 or 3 and are not life-threatening injuries. The tendency of bone to fracture depends on their mineral density, which is strongly correlated with age, gender and body mass [Funk, 2002]. Bone density decreases with age and increases with body mass. Also, males have higher bone mineral density than females, which increases their tolerance to fracture. Soft tissue injuries (ligaments/muscles/vascular), such as sprain and contusion (AIS 1), may also occur with or without hard tissue (bones) fractures. It should be noted that complex fractures can also cause serious vascular injuries which can be of course life threatening, especially in military applications (i.e. long time till medical treatment).

Lower extremity injuries are associated with important disability and impairment. For example, fracture of the calcaneus, which is common to AV mine strikes [Medin, 1997, 1998; Radonić, 2004], often results in long-term complications such as infection and osteoarthritis [Funk, 2002]. Because the healing of calcaneus fracture is poor and is usually associated with intense pain, such injury may result in a surgical amputation of the leg.

3.2.2 Injury Risk Model

The loading mechanism acting on the lower leg during an AV mine strike is comparable to the one observed during frontal car crashes during which axial loading is applied via the pedal or the toepan. Therefore, literature from car crash safety research has been used here. Since seat belts and airbags significantly reduce the risk of life-threatening injuries to the head, neck, chest and abdomen, the automotive community is presently concentrating an effort towards the study of other body regions such as the foot/ankle complex. Studies on PMHS lower leg tolerance to fracture under dynamic axial impacts, such as the ones of [Yoganandan, 1996; Griffin, 2001; Seipel, 2001; Kuppa, 2001; Funk, 2002], are of strong interest in the field of mine protection research. This section focuses on the Yoganandan model and its (dis) advantages, since the TG-25 has selected this model among the available models found in the literature. Analysis of the other injury risk models is presented in Annex E.

3.2.2.1 Description of Yoganandan Model

Dynamic axial impact tests were conducted by [Yoganandan, 1996], at the Medical College of Wisconsin (MCW). As shown in Figure 3.7, twenty-six intact adult lower legs were tested under dynamic loading using a mini-sled pendulum device.

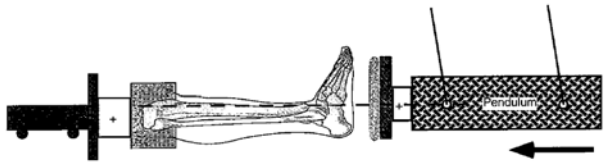


Figure 3.7: Experimental Set-up for MCW Dynamic Axial Impact Tests on PMHS Lower Legs [Yoganandan, 1996].

The forces were recorded on the pendulum impactor and at the upper end of the tibia. Statistical analysis, using the Weibull technique [Devore, 1999] was performed to derive a risk function. The probability of foot/ankle fracture is described as a function of specimen age and dynamic axial force measured at the distal end of the tibia. The pendulum maximal velocity ranged from 2.2 to 7.6 m/s and fracture tibia forces ranged from 4.3 to 11.4 kN. Results of MCW tests were combined to the ones obtained from similar studies at Wayne State University and Calpan Corporation to derive the foot/ankle injury risk equation. The model was developed using a sample of 52 specimens with an age range of 27 to 85 years old. Figure 3.8 shows the probability distribution for foot/ankle injuries as a function of age and tibia force (peak value). Plus/minus one standard deviation limits are shown in dotted lines and solid circles represent the fracture and non-fracture data points. Fractures, including extra/intra-articular (outside or inside the joint) fractures of the foot/ankle and distal tibia complex, were reported.

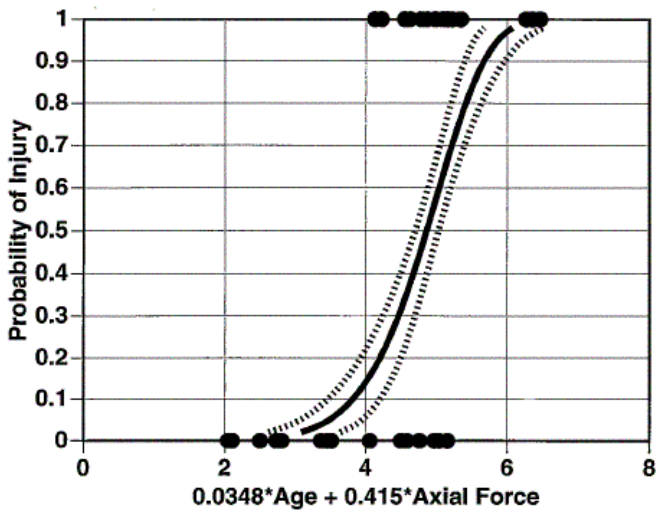


Figure 3.8: Risk of Foot/Ankle Injury as a Function of Age and Tibia Axial Force [Yoganandan, 1996].

Based on the relation given on Figure 3.8, the probability of foot/ankle fracture can be expressed as follows:

$$p(fracture) = 1 - \left[\exp\left\{ -\left(\frac{0.0348 * age + 0.415 * force}{5.13076} \right)^{7.42582} \right\} \right]$$

where:

- p(fracture) is the probability of foot/ankle fracture;

- Age is in years; and
- Force is the maximum tibia axial force value (in kN).

Using this Weibull probability equation presented above, the risk of foot/ankle fracture as a function of tibia force can be computed for any age. Figure 3.9 presents foot/ankle injury risk curves for 25, 45 and 65 years old.

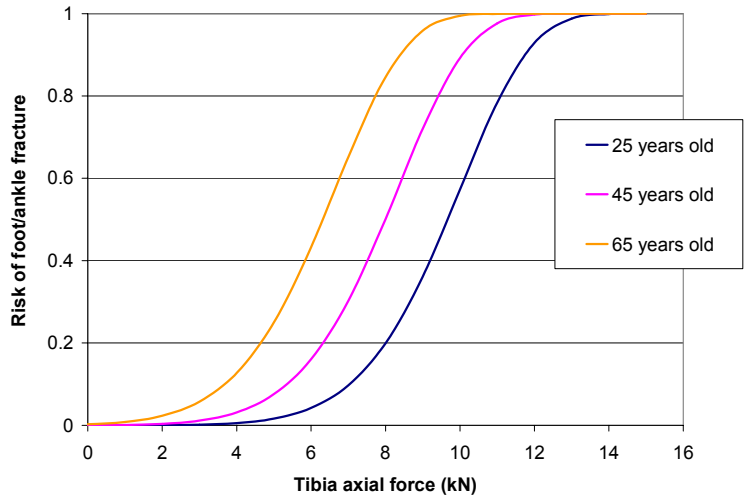


Figure 3.9: Foot/Ankle Injury Risk Curves for 25, 45 and 65 Years Old (Based on Data from [Yoganandan 1996]).

Based on the curves shown in Figure 3.9, the tolerance value for 25, 45 and 65 years old are respectively 7.0, 5.4 and 3.8 kN, representing 10% risk of foot/ankle fracture (AIS 2+). To protect most of the population in military vehicles (having an estimated age range of 20 to 45 years old), the tolerance value of 5.4 kN (for 45 years old) was chosen by the TG-25. Because most of the specimens used by Yoganandan were male (39 on 52) and the average specimen's weight was 75 kg, this tolerance value is applicable for mid-size males or large females [see Chapter 4, Figure 4.1 and Figure 4.2].

3.2.2.2 Applicability of Yoganandan Model

The Yoganandan injury risk model takes into account the most important factor affecting the lower leg tolerance to fracture: age. Because the specimens available to perform this kind of tests are usually old, the injury risk models developed without considering this factor [Seipel, 2001; Griffin, 2001; Kuppa, 2001] give very severe tolerance values. The injury risk model by Funk [Funk, 2002], including age, gender and weight, was also a satisfying model. The Yoganandan model was chosen instead of the Funk model because of the larger sample size (52 specimens) used to develop the model and the large age range (27 to 85 years old) of the specimens. Funk study was done with 30 specimens with an age range of 41 to 74 years old. However, as shown in Annex E, the Funk and Yoganandan models correlate very well for age of 45 years old which gives confidence in the lower leg injury tolerance value established by the TG-25.

3.2.2.3 Limitations of Yoganandan Model

The Yoganandan injury risk model, as well as all other similar models found in the literature [Kuppa, 2001; Griffin, 2001; Seipel, 2001; Funk, 2002] was developed for a pure axial loading. As shown on Figure 3.7, the lower leg was placed in a neutral position (ankle at approximately 90 degrees) and was

axially loaded. The tibia axial force was then used to compute the risk of foot/ankle complex injuries. Depending on seating configuration and footrest systems which are used in the mine strike tests the load transfer through the tibia can differ compared to this pure axial loading conditions. In that case, the risk of foot/ankle might be underestimated. A preliminary study on the effects of lower leg positioning for AV mine strike simulations was done using the multi-body/finite element software package MADYMO [Horst, 2004]. This study (summarized in Annex E), showed how the lower leg positioning can affect the tibia axial force and therefore the injury prediction when using the Yoganandan model without necessarily representing the real risk of foot/ankle injuries.

The proposed model, as well as other lower leg injury models found in the literature, is based on peak tibia axial force value and does not consider duration of the loading application. Due to the visco-elastic properties of the human tissues, their injury tolerance is believed to be time-dependant [Yoganandan, 1989]. The currently available data on injury risk models for lower legs are generated by (simulated) car crashes, which are usually slower incidents than mine strikes. Since standard Hybrid III tibia loading durations seen in AV mine strike tests are very short (can be less than 10 ms), the proposed tolerance value could be too severe. At this point, it is not known among the loading duration, the rise-time, and the impulse, which one has the strongest influence on injury tolerance. WTD 91, Meppen, Germany has started to develop a time-dependant lower leg injury criterion based on tibia loading duration, called the Lower Leg Threshold (LLth). More information about this criterion can be found in Annex E. However, so far the validation of this criterion is not available.

Finally, the use of the anthropomorphic test device (ATD) can have some limitations for injury assessment as well. As will be discussed in Chapter 4, HFM-090/TG-25 proposes to use the Hybrid III 50th percentile male ATD. The standard Hybrid III instrumented lower leg biofidelity is known to be poor based on recent PMHS tests. [Owen, 2001] showed that for non-injurious testing, when the lower leg is axially loaded on the heel, the Hybrid III tibia peak axial force is approximately 1.2 to 1.6 times higher than the one measured on PMHS tibia. Based on these results, the proposed lower leg injury criterion could be too conservative when using the standard Hybrid III lower leg. Validation for the mine loading conditions is not possible at the moment. More discussion on the choice of the ATD leg model for injury assessment is presented in Chapter 4.

3.2.3 Conclusions

In conclusion, the proposed lower leg injury model is the best available in the open literature at this point. The Yoganandan model was chosen as opposed to other models because of the large sample size of lower leg PMHS within a wide age range. The mathematical model proposed by HFM-090/TG-25 is taking specimen age into account, which is the most important factor affecting the injury tolerance. Based on this parameter, it was possible to establish a realistic tolerance value for military vehicle occupants who are relatively young. The proposed tibia axial force tolerance value is 5.4 kN (10% risk of AIS2+). One major limitation of the model is that it was developed for a pure axial loading and that risk of injury may be underestimated when the lower leg is loaded differently. The other important limitation of the proposed lower leg injury assessment method could be the choice of the ATD lower leg model (see Chapter 4). Based on these limitations, further research on lower leg injury assessment methods was identified as a priority by the HFM-090/TG-25.

3.3 THORACO-LUMBAR SPINE

This section provides information on thoraco-lumbar spine anatomy and injuries, and on the injury risk model to be used when performing AV mine protection testing. Additional information related to thoraco-lumbar spine injury assessment is provided in Annex F. Although pelvic injuries may also occur due to the high vertical loadings, it was assumed that thoraco-lumbar spine injuries are predominant and hence, are used as an indicator for pelvic injuries [Dosquet, 2003].

3.3.1 Anatomy, Loading Mechanisms and Injuries

The major functions of the vertebral column are protection of the spinal cord and stiffening for the body, and attachment for the pectoral and pelvic girdles and many muscles. The vertebral column extends from the base of the skull to the tip of the coccyx. It contains 33 discrete vertebrae and 23 intervertebral disks. These vertebrae are held together by joints and ligaments. The joints and ligaments provide stability to the spinal column while at the same time providing limited movement. There are different types of vertebrae [see Figure 3.10], and each can be easily identified by their definite physical characteristics.

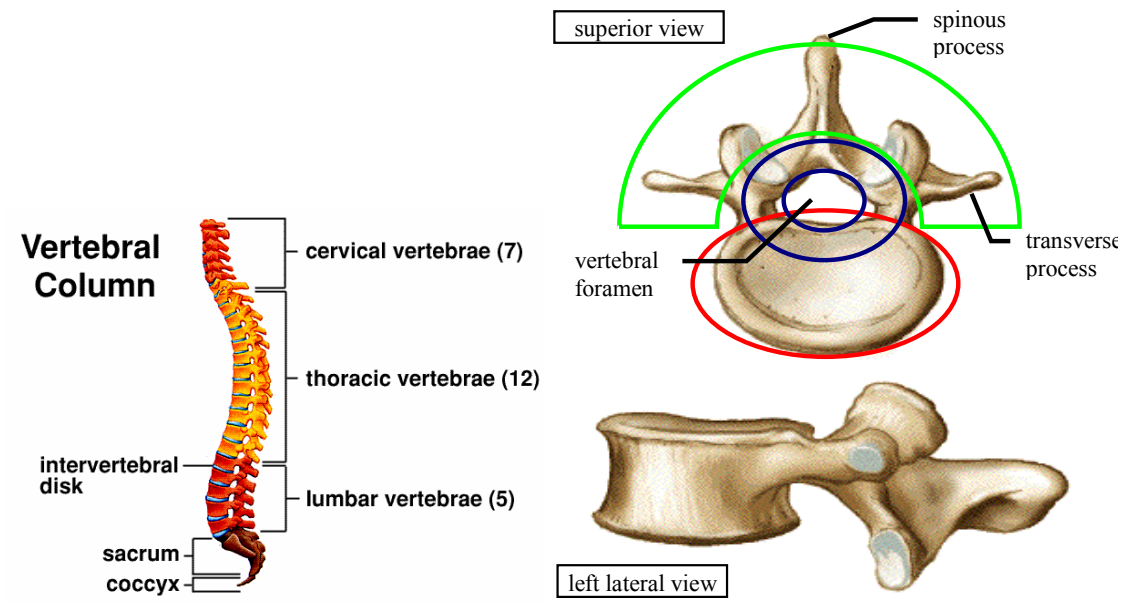


Figure 3.10: Lateral View of Vertebral Column (left) and (on the right) Lumbar Vertebra with Vertebral Body (red), Vertebral Arch (blue) and Facet Joints and Processes (green).

The seven cervical vertebrae have a dorsal spine and a large foramen in the transverse process. Following the cervical vertebrae are the thoracic vertebrae, which are medium in size compared to the cervical and lumbar vertebrae. The main function of the thoracic spine is to protect the organs of the chest by providing attachment for the rib cage. The range of motion in the thoracic spine is limited. The lumbar vertebrae are the heaviest and largest of the three types of individual vertebrae due to their weight bearing function of the body. They have a long transverse process and a thick dorsal spine (see Figure 3.10). The sacrum which consists of five fused vertebrae follows the fifth lumbar vertebra. At the end of the line three to five vertebrae are fused to the coccyx.

Each vertebra has three functional parts:

- The anterior vertebral body has to bear weight and withstand compressions.
- The posterior arch-shaped bone provides protection for the spinal cord.
- Muscles are attached to the posterior processes. The facet joints allow a limited backwards motion of the spine.

The spinal column is one of the vulnerable parts of crew members in vehicular mine incidents due to different loading mechanisms in cranial (axial) direction. Most critical is a direct impact of the elastic structural deformation via the seat system to the spinal column. Furthermore the transfer of the shock wave via the structure and the seat system can cause serious injuries of the spinal region.

Various types of injuries have to be expected: damage to the bony structure (i.e. vertebral fractures) as well as ligamentous and muscle injuries. Vertebral fractures cause the bone to loose height and form a wedge shape. In very severe compression fractures, the back of the vertebral body may protrude into the spinal canal putting pressure on the spinal cord. The spine can withstand an enormous amount of pressure; however, if the force is too great for the vertebra to withstand, one or more of them may break (fractures of the spinous and transverse process (AIS 2), the facet (AIS 3), the lamina (AIS 3), the pedicle (AIS 3) and the vertebral body (AIS 2-3)).

The thoraco-lumbar spine is the most vulnerable part of the spinal column when subjected to axial loading, as presented in a study of U.S. Army non-fatal helicopter crash injuries from 1979 – 1985 [Shanahan, 1989]. The lower thoracic (T12) and the upper lumbar vertebral bodies (L1 – L2), are specifically at risk because of the relation of the strength of and loads on the vertebral bodies due to the effective body weight [Dosquet, 2003].

The probability of injuries due to loading mechanisms in lateral and dorso-ventral direction are quite low because of the predominance of axial load scenarios with a mine threat. More than 50 mine protection trials with a huge variation of the boundary conditions (vehicle mass, seat adaptation, mine types, etc.) illustrated this [Dosquet, 2003].

3.3.2 Injury Risk Model

Loads with durations less than 10 ms presuppose injury risk models which consider the dynamic behaviour of the thoraco-lumbar spine. Hence a dynamic function and not a constant value would reflect the requirements for an assessment criterion for mine related incidents. The Dynamic Response Index (DRIz) model meets these requirements, whereas other available injury models for the thoraco-lumbar spine disregard the duration dependency or are not validated. In this section, only the recommended Dynamic Response Index (DRIz) for the cranial direction (Z-axis) will be described in detail, since TG-25 decided that this DRIz model fits the best for the injury assessment in AV mine strike tests. Other models are presented in Annex F.

3.3.2.1 Description of DRI Model

Stech and Payne [Stech, 1969] evaluated the Dynamic Response Index (DRI) as a general model to simulate the biomechanical response due to human body dynamics by using a single mass-spring-damper system (see Figure 3.11). The general model was already introduced by Latham in 1957 [Latham, 1957] to describe the impact of ejection seats to the human body. The DRIz is the Dynamix Response Index in axial direction (z-direction).

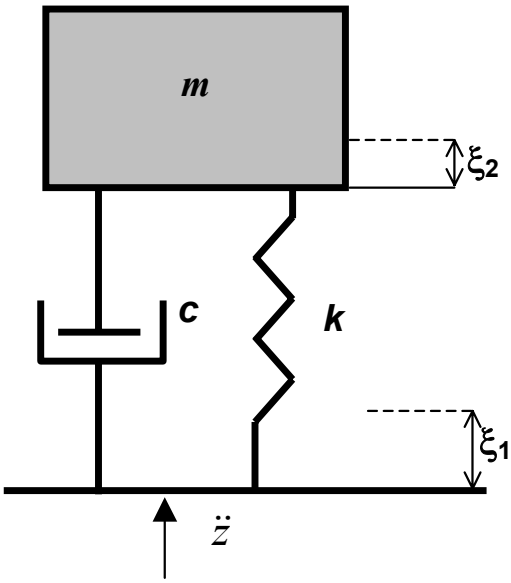


Figure 3.11: DRIz Model [Stech, 1969].

The equation of motion for this model is:

$$\ddot{z}(t) = \ddot{\delta} + 2 \cdot \zeta \cdot \omega_n \cdot \dot{\delta} + \omega_n^2 \cdot \delta$$

where

- $\ddot{z}(t)$ is the acceleration in the vertical direction measured at the position of initiation
- δ is the relative displacement of the system with $\delta = \xi_1 - \xi_2$; and $\delta > 0$ => compression
- ζ is the damping coefficient with $\zeta = \dfrac{c}{2 \cdot m \cdot \omega_n}$
- ω_n is the natural frequency with $\omega_n = \sqrt{\dfrac{k}{m}}$

The DRIz is calculated by the maximum relative displacement δ_{max}, ω_n and the gravity acceleration g:

$$DRIz = \frac{\omega_n^2 \cdot \delta_{max}}{g}$$

The first application was a dynamic model of the human body in spinal direction by Stech and Payne (vertical compression of the spinal column). They selected the values of ζ and ω_n, 0.224 and 52.9 radians/s respectively, as values for a representative population of Air Force pilots with a mean age of 27.9 years. These values were estimated based on investigations on compressive individual vertebral strength by Geertz [Ruff, 1950] as well as on load-deflection curves [Yorra, 1956]. Ruff [Ruff, 1950] summarized the investigations by Geertz during WW II. The tests were made either with individual vertebrae or vertebral complexes of PMHS between 19 and 46 years of age. By using Geertz data as an indication for vertebral compression fractures, Stech and Payne related the DRIz value to an injury risk of 50% depending on the age of the population. For an average age of 27.9 years, they calculated a DRIz of

21.3. This value was used as a baseline to introduce the function of spinal injury risk due to compressive loads versus DRIz values by assuming a normal distribution [Brinkley, 1970]. This function is presented in Figure 3.12 as the laboratory data curve.

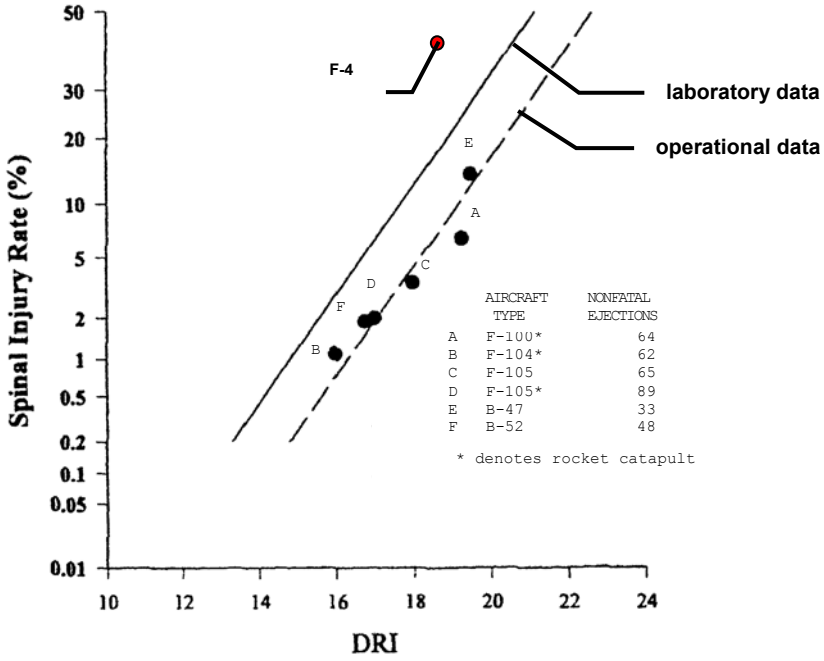


Figure 3.12: Spinal Injury Risk Calculated from Laboratory and Operational Data and F-4 Operational Data [Brinkley, 1970] Valid for AIS 2+ Injuries.

Furthermore, they presented the injury risk as a function of DRIz based on operationally experienced non-fatal spinal injury probabilities, which were calculated in ejection seat tests (Figure 3.12 operational curve). The relation of injury risk and DRIz is only valid for misalignments of the ejection seat compared to the catapult acceleration vector of less than 5°. Brinkley and Shaffer stated that the seat of an F-4 aircraft did not permit the crewman vertebral column to be aligned with the catapult acceleration vector as in other Air Force ejection seats; therefore, they excluded the F-4 data. When not aware of this misalignments and just using the operational curve, a risk of 9% would be predicted for these F-4 seat. However, when looking at the real data a DRIz of 19 related to a spinal injury probability of 34% for the F-4 seat, which is much higher. Therefore, it is assumed that for higher misalignments than 5° the tolerance levels decrease, resulting in higher injury risk at the same DRIz value.

3.3.2.2 Applicability of DRI Model

It should be noted that durations and maximal values of the transmitted load to the crew members varies significantly due to different threat types and vehicle concepts such as different vehicle masses, seat adaptations and seat concepts. However, the DRIz includes a dynamic behaviour which is necessary for injury assessment of the short duration load occurring during a mine strike.

The injury model based on the laboratory data (compressive strength of vertebrae) has the pelvis as point of initiation (location where load is transferred to the model), whereas the model based on the operational data has the seat as point of initiation [Brinkley, 1970]. However, in several mine protection trials, the seat acceleration data have shown a high variation and a lack of reproducibility. Although the characteristics of the DRIz model is comparable to a filter, the variation of the seat acceleration input has an impact to the

reproducibility of the DRIz data, whereas the variation of DRIz data based on acceleration measurements in the pelvis of the anthropomorphic test device (ATD) was quite low. Therefore, using the pelvis accelerations of the ATD is recommended.

Originally, the DRIz calculations are based on seat accelerations. However, in some countries seat cushion accelerations or pelvis accelerations are used as input for the DRIz model. Investigations on the DRIz calculation based on acceleration data measured in a seat itself, the seat cushion and measured in the ATD pelvis in 63 tests of different test facilities demonstrated a sufficient correlation of the results [Dosquet, 2004]. The DRIz values based on pelvic accelerations are slightly lower than the seat cushion DRIz values, but these differences are not significant.

Based on these studies TG-25 decided that the DRIz calculations for the mine strike trails should be based directly on measurements of the ATD pelvis.

As aforementioned, the operational data curve is not strict enough for specific boundary conditions. Some military vehicles provide misalignments of 15° and more. Furthermore, seats of protected vehicles are often adapted to the roof or the wall of the vehicle. This conceptual boundary condition causes similar effects on the load on the crew member as higher misalignments. When the misalignments of the seats are higher than 5°, the injury risk will be higher than calculated with the operational data curve, which only accounts for loads in spinal direction. Although the operational data is based on real incidents, the underestimation of the injury risk for different seat concepts resulted in the TG-25 decision to use the laboratory data curve, which is more conservative than the operational data curve. When using this curve (see Figure 3.12) the tolerance level of 17.7 for DRIz refers to a 10% risk of AIS 2+ injuries.

3.3.2.3 Limitations of DRI Model

The spine injury mechanism is force driven. Using a model which is based on another physical parameter than force, e.g. the pelvis acceleration, may introduce uncertainties. However force based injury models taking the visco-elastic properties of the vertebral column into account are not available yet.

3.3.3 Conclusions

In conclusion, the Dynamic Response Index (DRIz) model is at this point the best available model for thoraco-lumbar spine injury assessment. Considering relative low injury risks of spine injury due to lateral (y) and dorso-ventral (x) loads TG-25 decided to only assess spine injuries in the vertical (z) direction and use the DRIz model. The DRIz is to be calculated with the pelvis ATD vertical acceleration. Based on discussions on the two risk curves available in literature (see Figure 3.12), HFM-090/TG-25 decided to use the more conservative risk curve derived from laboratory data. When using this curve (see Figure 3.12) the tolerance level of 17.7 for DRIz refers to a 10% risk of AIS 2+ injuries.

3.4 NECK

This section provides information on neck anatomy and injuries, and on the neck injury criteria to be used when performing AV mine protection testing. Additional information related to neck injury assessment is provided in Annex G.

3.4.1 Anatomy, Loading Mechanisms and Injuries to the Neck

3.4.1.1 Anatomy

The cervical spine is the most important part of the neck and it comprises of seven vertebrae forming eight motion segments between the base of the skull and first thoracic vertebra (see Figure 3.13, the most upper

vertebra is denoted C1 and the lowest is C7). The cervical spine (neck) is divided morphologically and mechanically into two regions, the upper cervical and lower cervical spine. The lower cervical spine contains vertebrae C3 through C7. The upper cervical spine consists of three bony elements: the occiput (the base of the skull), the atlas (C1) and the axis (C2). Enlarged facets on the atlas allow for articulation with the base of the skull. The axis has a bony projection called the dens, or ordontoid process, which projects upward and serves as the vertebral body of the atlas. The dens also serves as the axis about which the head and atlas rotate.

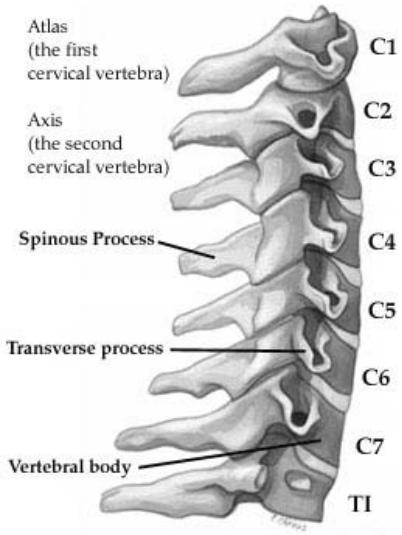


Figure 3.13: Cervical Vertebrae [Gray, 1985].

A number of ligaments hold the vertebral bodies together. These include the anterior and posterior longitudinal ligaments, which span the anterior and posterior portions of the vertebral bodies. There are several additional ligaments, which are named for the parts of the vertebral structure, which they connect. The vertebral bodies are connected to each other by means of an intervertebral disk. The disk is a fibrocartilanginous structure composed of a central nucleus pulposus and surrounded by the annulus fibrosus, a laminar set of fibrous sheets.

3.4.1.2 Loading Mechanisms

Head and neck motion can be described anatomically as shown in Figure 3.14, which depicts these and associates terms with the movements of the head/neck complex.

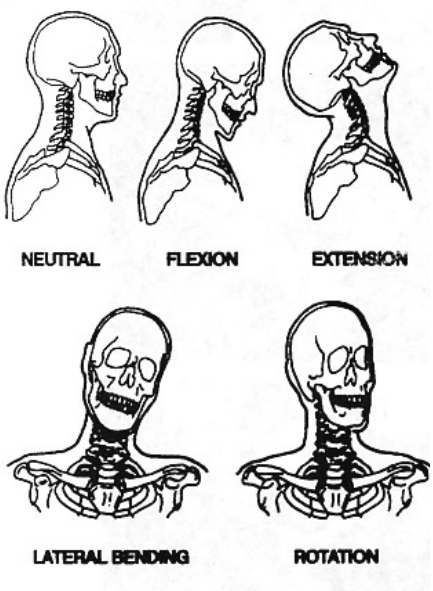


Figure 3.14: Anatomical Description of Head Movement [McElhaney, 2002].

A useful way to classify neck injury mechanisms is by an engineering description of loading as depicted in Figure 3.15.

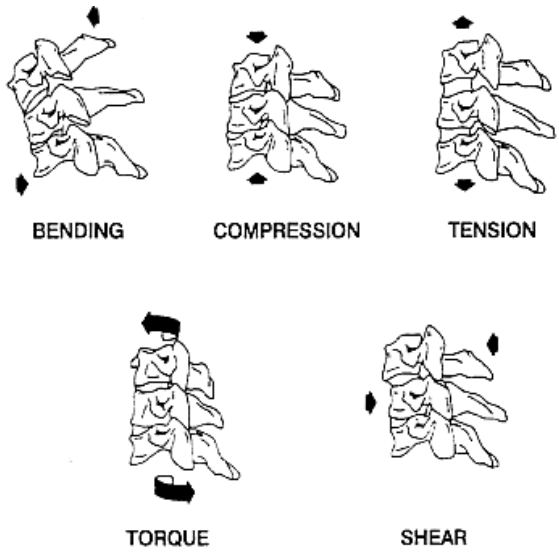


Figure 3.15: Engineering Description of Neck Loading [McElhaney, 1993].

The loading mechanisms described by Figure 3.15 may occur also at thoracic and lumbar region of the spine. However, neck injuries are usually more life-threatening since they may affect the phrenic nerve, responsible of the respiration functions.

In anti-vehicular (AV) mine blast incidents it is impossible to predict precisely which injury mechanism will occur. Primary location of the blast, type of vehicle, positioning of the occupant, geometry of the interior of the vehicle relative to the position of the crew member, and movement of the individual after the initial loading phase all play a part in determining what mechanism, if any, contributes to an injury.

However, the most important neck injury mechanisms expected to occur during a mine strike are the followings:

- Axial (vertical) compression;
- Forward bending (due to flexion);
- Rearward bending (due to extension); and
- Lateral bending.

In order to facilitate the reading of the next paragraphs, the following terms will be used to discuss each of these injury mechanisms mentioned above: axial compression, flexion, extension and lateral bending.

While purely compressive loading of the neck is considered to occur infrequently, many AV mine incidents may be considered to be predominantly **axial compressive loading**. With the detonation of an anti-vehicular landmine, the major component of vehicle and occupant acceleration is vertical. Many tactical vehicles have very little space between the crown (top of helmet) and the roof structure. Although the intent is to restrain the occupant to prevent excessive movement, necessary limited stretching of the upper restraints and movement of the seat complex may result in a nearly compressive strike of the occupants head into a roof structure. Based on full-scale test data, when there is no impact of the head on the roof, the axial load transmitted through the spine is not considered being important enough to cause neck injuries since the load will mainly be absorbed by the thoraco-lumbar region.

Neck injuries will usually be the result of more than one injury mechanisms. For example, automotive crash investigations indicated that compression-extension injuries resulted from head impact with the windshield, resulting in the simultaneous extension and compression on the neck, giving rise to posterior structure injuries (King, 2000). During AV mine strike, head impact on the roof will probably result in **axial compression combined with extension, flexion or lateral bending**, depending on the head position prior to impact. Finally, injury to the neck may also result of inertial loading, i.e. violent head movement without direct impact on vehicle structure.

3.4.1.3 Injuries

Trauma to the neck has been divided into three main categories (Pikes, 2002):

1) Osseous (bone) injury;
2) Soft tissue injury; and
3) Neurological injury.

Osseous injury refers to vertebral fractures. The severity level of these fractures will depend on: whether or not the fracture resulted in impingement of the neurological structures, and whether or not the fracture is structurally stable. For example, fracture of the atlas (C1) by compression is called the 'Jefferson' fracture, and may occur if the head impact the vehicle roof. Other examples of vertebra fractures are given in Annex G.

Neurological injury is the most severe class of neck injuries, which may result in paralysis, long-term pain and even death. They generally are the result of primary osseous or soft tissue injuries.

Soft tissue injury is the most common type of neck injury in vehicular crash environment. For example, neck sprain, in which the ligaments are stretched beyond their normal limits, is a result of flexion-extension (also called 'whiplash'). Ligamentous injuries are usually minor, but when severe, they can impart severe neurological impairment.

3.4.2 Injury Criteria

Numerous experimental studies, both static and dynamic, have been conducted over the past 40 years to determine injury producing loads and develop injury criteria. The well known Neck Injury Criterion (Nij), which considers combined injury mechanisms and is used as a standard for frontal car crashes, was not chosen for the reasons mentioned in Annex G. The proposed neck injury assessment method considers all injury mechanisms separately. The method includes axial compression, flexion and extension injury criteria, which are described in the next paragraphs. Due to lack of information on Hybrid III neck biofidelity and injury criteria for lateral bending, no injury criterion for this specific loading mechanism was included.

3.4.2.1 Description of Axial Compression Injury Criterion

Mertz et al. [Mertz, 1978] undertook a study to develop axial compression neck injury tolerance curves. Impact tests of a spring-loaded tackling block on football helmets [Mertz, 1978] were conducted with a 50th percentile male Hybrid III and then compared them to real accidents of high school aged football players. One player (17 years old, 1.88 m, 101 kg) suffered immediate paralysis of the arms and legs and X-rays showed fractures of C3, C4 and C5. A second 17 year old player (1.82 m, 95 kg) suffered fatal injuries including hemorrhages in the brain stem and pons (white matter nerve fibers) and subarachnoid blood (area of brain containing cerebrospinal fluid, cushioning brain from mechanical shock). In a third incident, a high school player was allegedly struck and rendered quadriplegic by the same tackling block used in Mertz's experiments.

In the tests, the Hybrid III was oriented so that the load was applied to the top of the head, loading the neck structure in compression with minimal head rotation (appropriate figure of the test set-up was not available). The configuration was chosen to produce the maximum value of neck compression force for the impact velocity chosen. The neck compression load measured by the Hybrid III was considered representative of the upper bound of maximum axial compressive load that an equivalent weight human would have experienced for the same impact velocity.

Based on this study, Mertz [Mertz, 1978] derived two injury tolerance curves based on the upper neck axial compression force measured on a 50th percentile male Hybrid III (see Figure 3.16). The coordinates of the 'Upper' curve are 0 ms and 6670 N, 35 ms and 1110 N, and greater than 35 ms, 1110 N. The coordinates of the 'Lower' curve are 0 ms and 4000 N, 30 ms and 1110 N, and greater than 30 ms, 1110 N. To evaluate neck load signal, pairs of points (force, duration) are plotted on the graph (shown in Figure 3.16) with the two injury assessment curves. The points are connected together by a series of straight lines. If any of the line segments lie above the upper curve (red), the neck axial compression force is considered to have the potential to produce serious neck injury. If any of the line segments lie above the lower curve (blue), the potential for neck injury from the axial compressive force is considered less likely. If none of the line exceeds the lower curve, the probability of neck injury from axial compressive force is considered remote.

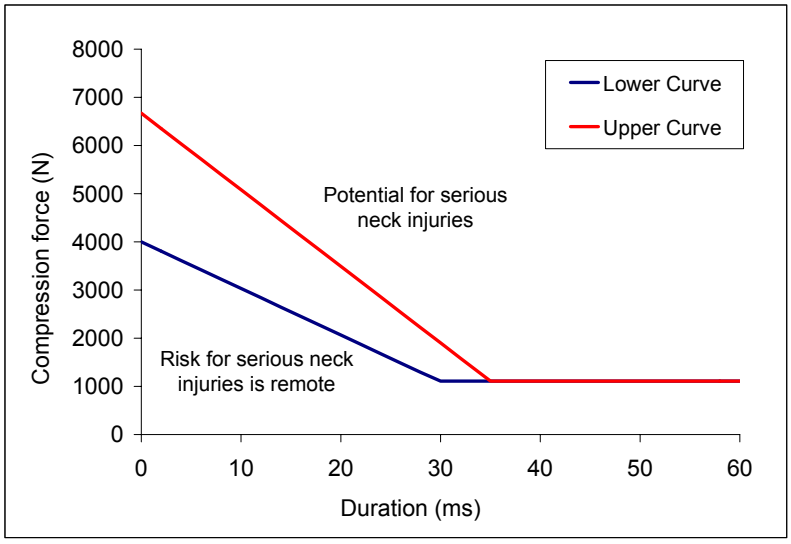


Figure 3.16: Injury Tolerance Curves for Axial Neck Compression Force when Using a Hybrid III 50th Percentile Male ATD [Mertz, 1978].

These levels were proposed for an adult population that was considerably older (exact age range not known) and much less conditioned than a high school football athlete. The time durations were determined from the loading times observed during the experiment, which were on the order of 30 – 40 ms.

The HFM-090/TG-25 proposed to use **the lower curve (blue)** in Figure 3.16 as the tolerance level for neck compression injuries in AV mine strike tests. Exceeding this level implies that serious neck injuries are likely.

3.4.2.2 Description of Flexion/Extension Injury Criteria

Mertz and Patrick [Mertz, 1971] as well as Patrick and Chou [Patrick, 1976] conducted tests on volunteers and PMHSs to determine neck reaction on the head under dynamic conditions. Human volunteers were subjected to static and dynamic tests to produce non-injurious response for neck flexion and extension. PMHSs were used to extend the data into the injury region. Their analysis of the data indicated that equivalent moment at the occipital condyles (protusions on the back of the skull which articulate with the cervical vertebra) was the critical injury parameter in flexion (forward) and extension (backward).

The dynamic tests were conducted with one human volunteer and four PMHS. The subjects were restrained in a rigid chair mounted on an impact sled (appropriate figure of the test set-up was not available). The sled was accelerated pneumatically over a distance of 2 meters to a prescribed velocity. A headrest was used to maintain the head in the upright position. After reaching the prescribed velocity the sled was stopped with a hydraulic cylinder, which produced a repeatable stroke. The occupant was restrained with a lap belt and two shoulder harnesses, which crisscrossed at the sternum. The human volunteer was subjected to 46 sled runs of various degrees of severity for four configurations of head weight. The subject attempted to achieve two different degrees of muscle tenseness, relaxed and tensed. With muscles tensed the volunteer was subjected to sled decelerations of 1.9 – 6.8 g. With weight placed above the center of mass of the head the volunteer was subjected to a 9.6 g deceleration. This particular run resulted in neck and back pain that lasted several days. The PMHS were similarly fixed so as to obtain comparisons of responses to the same sled deceleration pulses for the various configurations and head weights, and then subjected to more severe conditions. A total of 132 PMHS runs were completed using four subjects.

The volunteer withstood a static flexion moment about his occipital condyles of 35 N·m. This was without contribution from chin contact to the chest. A dynamic tolerance level for the flexion moment about the occipital condyles for the initiation of pain occurred with the maximum equivalent resisting moment of 59 Nm. The maximum dynamic flexion moment generated by the volunteer was measured to be 88 Nm. This level produced a sharp pain in the neck and upper back region with soreness persisting for several days. It was considered non-injurious, but close to the injury threshold. The PMHS were exposed too much greater decelerations. Based on the PMHS data, it was observed that the 50th percentile human could withstand equivalent flexion moments of 190 Nm without suffering ligamentous injury (AIS 1 or 2) or bone injury (AIS 2 to 6). However, it is expected that muscle injuries (AIS 1) would occur at a lower value of the flexion moment. Based on these results Mertz [Mertz, 1978] proposed an injury tolerance level of 190 Nm for neck flexion moments measured with a 50th percentile male Hybrid III. Similarly from the aforementioned study, corridors for extension were proposed with an injury tolerance level of 57 Nm [Mertz, 1978]. This level was associated with ligamentous damage (AIS 1 or 2), to a PMHS.

The HFM-090/TG-25 proposed to use the tolerance levels defined by Mertz, as described above for neck flexion (190 Nm) and extension (57 Nm). Exceeding this level implies that serious neck injuries are likely.

3.4.2.3 Limitations of the Proposed Injury Criteria

A first limitation of the proposed neck injury assessment method is that no injury risk curves were available such as for the lower leg and the spine. Also, the definition of the injury severity associated with the proposed injury tolerance values was not quite clear. However, based on Mertz [Mertz, 1984], exceeding the proposed tolerance levels for compression as well as flexion/extension implies that **serious** neck injuries are likely. Based on the Abbreviated Injury Scale (AIS, 1990), serious injuries have a score of 3 and according to Mertz [Mertz, 1984], a serious injury refers to AIS 2 or more.

Another limitation is that the method considers injury mechanisms separately while the loading is typically a combination of more than one injury mechanism. However, the well known Nij, which considered combined injury mechanisms, was not appropriate (see Annex G) for the current purpose. Also, only three injury mechanisms (compression, flexion and extension) were considered while many other may occurs especially when the AV mine is not detonation directly under the occupant location. Three-dimensional motions of the human body and its parts may also result in lateral bending, shear, tension or torsion. Injury criteria for these other injury mechanisms are summarized in Annex G.

Finally, the proposed injury criteria were developed for the loading duration seen in automotive crash situations, while during an AV mine strike the initial loading duration could be much shorter. However, the loading durations for the global effects and the drop down phase are expected to have a similar duration.

3.4.3 Conclusions

Neck injury tolerance has been extensively researched by the commercial automotive industry for the past 50 years. Concern regarding head strike during front crash, whiplash in rear strike and contact injuries in rollovers has yielded criteria utilized by the European and United States regulating bodies. HFM-090/ TG-25 proposed to use the criteria developed by Mertz and Patrick (1971), Patrick and Chou (1976) and Mertz (1978) for axial compression, flexion and extension. The injury tolerance values for neck compression was set to 4 kN at 0 ms and 1.11 kN at 30 ms (see Figure 3.16). If all force-duration data points fall below this tolerance limit, then the risk of serious neck injuries due to compression is unlikely. If flexion and extension bending moment peak values are below 190 Nm and 57 Nm, respectively then the risk serious neck injuries are unlikely. Although no specific information on AIS level was given for these tolerance values, they are believed to represent a low risk of AIS 2+ injuries based on information given by Mertz [Mertz, 1984].

Finally, since pure compressive or flexion/extension loadings are relatively rare (for an AV mine strike as well as for car crashes), an injury assessment method considering combined injury mechanisms, would have been more appropriate. However, the current available model (Nij) appeared not to be suitable for AV mine loading conditions. Also, it is still recommended to extend neck injury assessment to other injury mechanisms such as shear, tension and lateral bending if appropriate methods exist.

3.5 HEAD

This section provides a general overview on injuries, injury mechanisms and injury assessment models for the head.

3.5.1 Anatomy, Loading Mechanisms and Injuries

The head can be divided into two parts: The face and the centre, and rear part of the head. The face is defined as the front part of the head, comprising the facial cranium, its skin, muscles, the blood vessels to the facial structures, as well as the facial nerves. The centre and rear part of the head include the neurocranium, its skin covering (the scalp) and its contents: the brain and meninges (Figure 3.17). The brain is connected to the spinal cord and serves as the upper, greatly expanded part of the central nervous system.

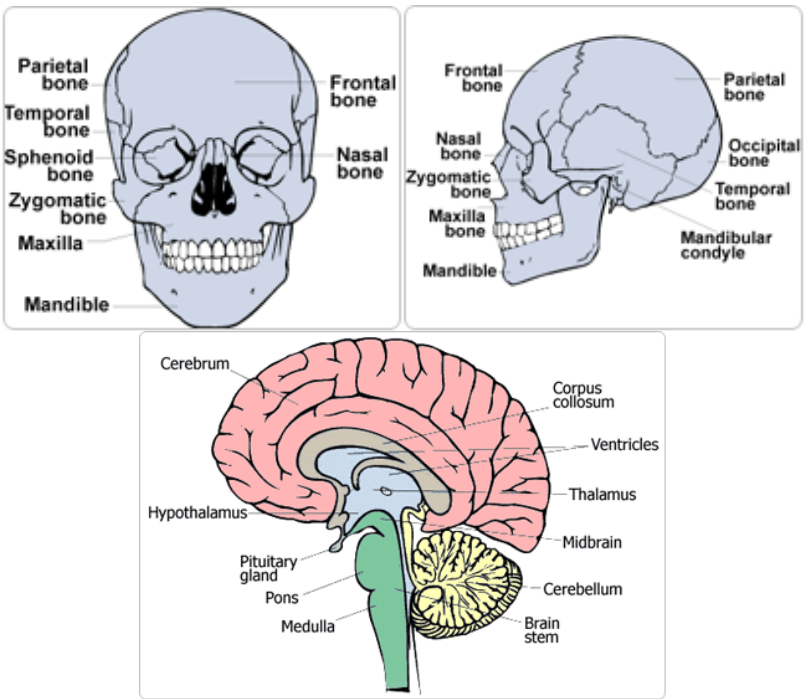


Figure 3.17: Anatomical Structure of the Head and Brain [AMA, 1998].

Head injuries were classified by Ommaya [Ommaya, 1985] as scalp, skull, extracerebral bleeding (focal or diffuse) and brain tissue damage (neural and/or vascular). Brain tissue damage can arise from concussions, contusion, intracerebral hemorrhage or brain laceration. The types of brain injury resulting from this tissue damage include focal injuries (contusions), diffuse axonal injuries (DAI) and mass lesions (hematomas).

DAI were first discovered by Sabina and Strich [Sabina, 1961]. DAIs are typically referred to as “closed head injuries”, which are damages to axons in the white matter of the brain. They are caused by both rotational and/or translational accelerations. Focal injuries are primarily due to direct impact of the head.

Mass lesions are due largely to angular accelerations suffered by the brain. Injuries to the skull include suture separation, indentation, linear fracture, depressed fracture and crushed skull.

Many current researchers in head injury biomechanics consider angular acceleration together with linear accelerations to be the principal cause of brain injury, whereas traditional biomechanical research focuses on the linear accelerations only [King, 2003]. Pressure, shear, stress, strain rate and the product of strain and strain rate are mentioned as injury mechanisms. Until now research on head injury biomechanics is still ongoing.

An AV mine strike under a vehicle will result in a global movement of the body of the vehicle occupants, which may result in violent head 'direct' impact (on hard surface) and may cause skull and brain injuries. Brain injuries may also result of inertial loading (no 'direct' impact), where as skull fracture occurs as a result of 'direct' impact only [Little, 1993]. The injury mechanism of impact involves a short-duration impulsive loading and a high-acceleration peak while the inertial loading is associated with linear and/or angular acceleration pulse over a significant longer time period [Little, 1993]. In case the occupants wear a protective helmet, skull injury risk is minimized. The influence of the mass of the helmet on the neck injuries is still ongoing research, however, it is not expected that for the vertical impacts of short duration the helmet will influence the risk on neck injuries significantly.

3.5.2 Injury Assessment Models

Most of the head injury risk models use skull fracture as a predictor of brain injuries and show that fracture is linearly related to head linear acceleration. An example of one model, the Head Injury Criterion (HIC) [Mertz and Prasad 1997], was developed to assess skull fracture and brain damage caused by frontal head impact, and is largely used in automotive and aviation testing standards. The use of the HIC in cases where there is no head impact on hard surface or when the impact location is not frontal, is presently a controversial topic in the automotive research community. Other head injury models such as the Severity Index [Gadd, 1966] and the Power Index [Newman, 2000] were developed to assess head and brain injuries in case of 'direct' head impact, whereas the brain injury assessment models by Ommaya [Ommaya, 1984], Thibault and Gennarelli [Thibault, 1990] and Glaister [Glaister, 1997] were developed for the inertial loading conditions and focus on the head angular accelerations. Still much research is going on with respect to head injury criteria, resulting in contradictory statements, therefore a modification of the automotive standard is not available and still the HIC is used.

Because of lack of information on the applicability of the different available head injury assessment models, the HFM-090/TG-25 decided not to include mandatory assessment of this body region. This decision was also supported by the two following assumptions:

1) For an AV mine strike it is assumed that in case of hard head contact the neck loads will exceed the tolerance limits as well.
2) For the more severe cases of head inertial loadings caused by the vertical impact on the vehicle (for e.g. impact similar to a whiplash) it is assumed that other body regions (neck and thoraco-lumbar spine) will exceed their injury tolerance levels first.

3.5.3 Conclusions

Based on the assumption that head injuries would not occur without exceeding injury risk tolerance values of other body regions and because there is actually no satisfying head injury assessment model available, HFM-090/TG-25 decided not to include head injury assessment. In the case of a 'direct' head impact (on hard surface), it is assumed that the neck loads will be a predictor of neck (and head) injuries and in the case of inertial loading only (no 'direct' impact), it is assumed that no brain injury will occur without neck or spine failure.

It is still recommended to record head linear and angular (if appropriate techniques exist) accelerations in order to understand head impact mechanisms under the strike and also to collect data for possible future research on development of head injury assessment methods. Injury assessment models presented in this section may be used, however, their limitations must be taken into consideration before interpreting the results.

3.6 INTERNAL ORGANS

Organs (or systems) containing air or fluids are the most sensitive to overpressure caused by blast waves. These organs (or systems) are the auditory system (ear), the respiratory system, the gastrointestinal (G. I.) tract and some solid intra-abdominal organs [Axelsson, 1996]. This section provides on internal organs and blast overpressure injuries, and on the injury risk model to be used when performing AV mine protection testing. Additional information related to overpressure injury assessment is provided in Annex H.

3.6.1 Anatomy, Loading Mechanisms and Injuries

The following paragraphs give a short anatomical description of the auditory system and the non-auditory organs (respiratory system and gastrointestinal tract) vulnerable to blast overpressure. For further details on these body regions, the reader is referred to [Tortora, 2003].

3.6.1.1 The Auditory System

The auditory system (ear), shown in Figure 3.18, is the most susceptible body region to blast overpressure. Overpressure may cause ear injuries at the eardrum, the middle ear or the inner ear of different degrees of severity. Tympanic membrane rupture, dislocation of ossicles (malleus, incus, stapes) and damage of cochlea hair cells are examples of ear injuries (or damage) caused by overpressure. Middle and inner ear damage usually represent a risk of hearing loss. Ear injuries have an AIS score of only 1, even when they result in permanent hearing loss. The risk of ear injuries can be reduced by wearing ear protection such as earplugs or earmuffs.

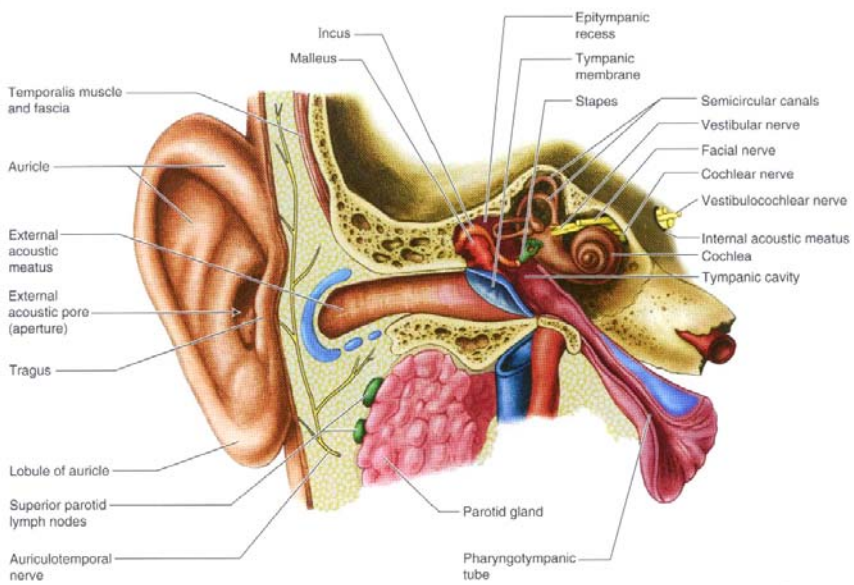


Figure 3.18: The Auditory System [Moore, 1999].

3.6.1.2 The Respiratory System

The respiratory system, shown in Figure 3.19, is divided in two parts: The upper respiratory system (including the nose and the pharynx) and the lower respiratory systems (including the larynx, trachea, bronchi and lungs) [Tortora, 2003]. The most fatal blast overpressure injuries are the ones to the lungs, because they may involve air embolisms. The air embolism is produced when the air enters the circulation from the disrupted lung tissues and may enter the brain or the heart. The *blast lung* is the most common overpressure blast injury. The term *blast lung* is used to describe the direct damage to the lung produced by the interaction of the blast-wave with the body generated by an explosion and is associated with lung hemorrhage (AIS 3 or more). The pharynx, larynx and trachea are also very vulnerable to blast overpressure, but their injuries will usually represent a lower threat to life. The severity of respiratory system injuries depends on the size of the lesion and the numbers of affected organ layers (for pharynx, larynx and trachea).

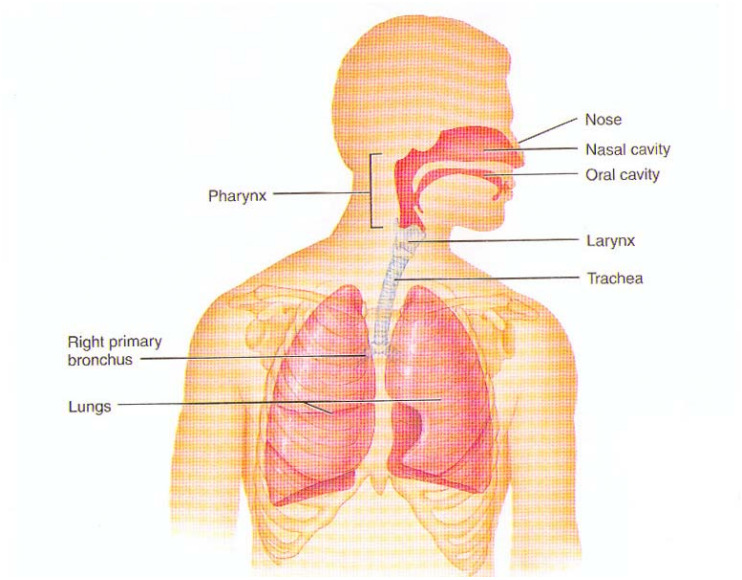


Figure 3.19: The Respiratory System [Tortora, 2003].

3.6.1.3 The Gastrointestinal (GI) Tract

The organs of the gastrointestinal tract (GI tract), shown in Figure 3.19, are the mouth, pharynx, esophagus, stomach, small intestine and large intestine [Tortora, 2003]. The wall of the GI tract is composed of four layers. The severity of GI tract injuries depends on the number of layers affected and the size of the lesion. Also shown in Figure 3.20, intra-abdominal solid organs such as the liver and the bladder may be injured when the body is subjected to blast overpressure.

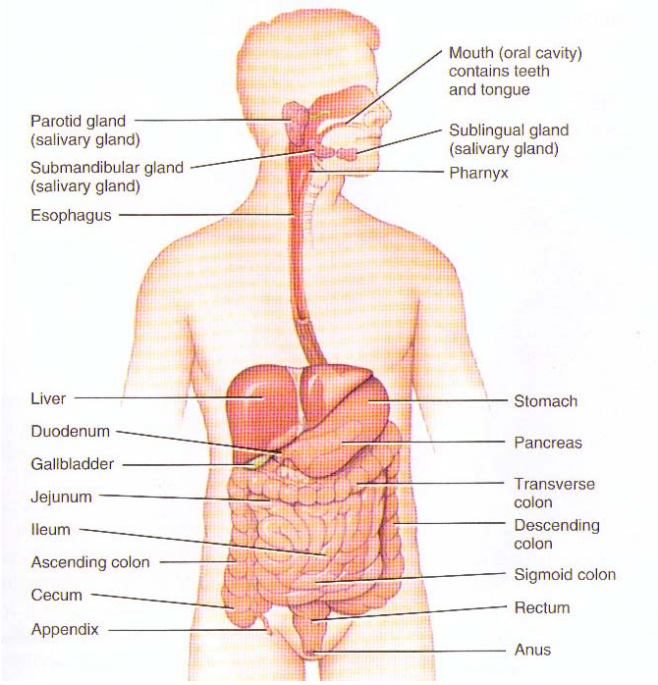


Figure 3.20: The Gastrointestinal Tract [Tortora, 2003].

3.6.2 Injury Risk Criteria

3.6.2.1 Auditory Organs

Several criteria are available for injury assessment to the auditory organs [e.g. Chan, 2001; MIL-STD-1474D, 1997; Richmond, 1992; NATO, 2003; Dancer, 1995]. However, in case of a mine detonation under a vehicle the risk of permanent auditory injuries can be minimized when proper ear protection is worn. For this reason and because ear injuries do not represent any threat to life (AIS 1), auditory injury assessment is not a mandatory criterion for the STANAG 4569.

3.6.2.2 Non Auditory Internal Organs

Non-auditory overpressure injuries are not expected to occur when the vehicle structure integrity is preserved during a mine strike. However, these body regions need to be assessed in order to ensure a minimum risk of life-threatening injuries caused by blast overpressure.

Previous non-auditory blast damage risk criteria have been based on exposure of mammals to approximately ideal blast waves, typically recorded in free stream in the open. These ideal blast waves (or Friedlander waves) are characterized by a nearly instantaneous rise in pressure followed by an exponential decay leading into a negative pressure phase. The tolerance of humans to single exposure to these ideal blast waves depends on the peak overpressure and the overpressure phase duration [Bowen, 1968]. However, the actual exposure environment in many situations is of more complex nature. For example, a detonation in an enclosure or a shaped charge warhead penetrating an armoured vehicle produces a reverberant wave or complex blast wave inside. A complex blast wave typically consists of a series of multiple shocks that may be superimposed on a slow rising pressure and may cause a variety of biological effects. In an AV mine strike complex blast waves are more common.

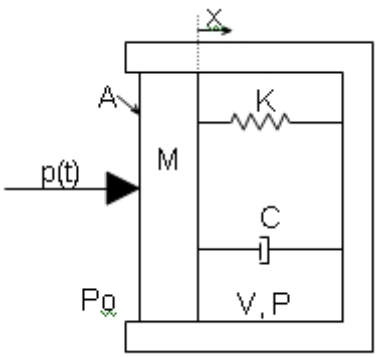
Stuhmiller et al. [Yu, 1990; Stuhmiller, 1998] as well as Axelsson & Yelverton [Axelsson, 1996] developed injury assessment models for the complex blast waves. However, Stuhmillers models focus on

the lungs only, while the model by Axelsson & Yelverton includes all non-auditory internal organs of the upper body. At this point, it is not known if the lungs are more susceptible to complex blast waves than other non-auditory internal organs. For this reason, the model of Axelsson was chosen in order to evaluate the risk of all potential injuries to the non-auditory internal organs. The model is described below, while the other models are discussed in Annex H.

3.6.2.3 Description of Axelsson Model

The objective of Axelsson & Yelverton study [Axelsson, 1996] was to understand the effects of complex blast waves on the human body in order to find a simple tool for vulnerability assessment. The model developed by Axelsson is based on experiments performed by Yelverton [Yelverton, 1993] in which 255 sheep and an instrumented cylinder were exposed to complex blast waves in enclosures. The instrumented cylinder (Blast Test Device) was placed where the sheep were positioned. The cylinder (described in details in Chapter 4) was an aluminium tubular test module approximating the size of a sheep. The cylinder was instrumented with four pressure gauges recording the blast loading coming from four different directions. Axelsson used data of 177 of the 255 sheep submitted to complex blast waves in Yelverton study in order to develop a transfer function between injury severity and blast loading recorded on the cylinder.

A mathematical model of the thorax was developed to predict injury severity as a function of the loading recorded by the cylinder submitted to complex blast waves in different enclosures. Initially a two-chamber spring-mass system (two lungs) was developed by Bowen [Bowen, 1965] and Fletcher [Fletcher, 1970]. The model was then simplified to a single chamber one-lung model [shown in Figure 3.21] assuming that the blast load, p(t), is acting simultaneously on both lungs.



A is the effective area;

M is the effective mass;

V is the initial gaseous volume of the lungs;

x is the displacement;

C is the damper coefficient;

K is the spring constant;

P_o is the ambient pressure;

p(t) is the overpressure over the time; and

γ is the polytropic exponent for gas in lungs.

Figure 3.21: Single-Chamber One-Lung Model [Axelsson, 1996].

The model is a single degree of freedom system in which chest wall response (displacement, velocity and acceleration) and intra-thoracic (lung) pressure can be calculated for different complex blast waves and ideal blast waves as well. The equation for the model is the following

$$M \cdot \frac{d^2x}{dt^2} + C \cdot \frac{dx}{dt} + K \cdot x = A \cdot \left[p(t) + P_o - \left(\frac{V}{V - A \cdot x} \right)^{\gamma} \cdot P_o \right]$$

with the model parameters given in Table 3.2.

Table 3.2: Model Parameters for a 70-kg Mammal*

Parameter	Units	70-kg body*
M	kg	2.03
C	Ns/m	696
K	N/m	989
A	m^2	0.082
V	m^3	0.00182
γ	–	1.2

* For mammals of different body weight, scaling factors can be used as described in Axelsson, [Axelsson, 1996].

The input to the mathematical model is the measured external reflected overpressure p(t), the blast loading on the cylinder. The maximal chest wall velocity can be calculated for the four gauges located on the cylinder (v_1, v_2, v_3, v_4). The average velocity V is then calculated (using v_1, v_2, v_3, v_4) and is used to determine the injury severity expressed by the ASII (Adjusted Injury of Severity Index).

The ASII was developed by Yelverton [Yelverton, 1996] and can be expressed as follow:

$$\mathrm{ASII} = (0.124 + 0.117 \cdot \mathrm{V})^{2.63}$$

Based on the injury description given by Axelsson and Yelverton for each of the 'injury levels' (designated as negative, trace, slight, moderate and extensive), it was possible to correlate these injury levels with an AIS score. Based on the AIS range given for each injury level, it was possible to establish a 'global' AIS range as presented in Table 3.3.

Table 3.3: Injury Levels with Corresponding AIS Range for Each Body Region*

Injury Level	Lungs	Pharynx/Larynx and Trachea	GI Tract (Bowel)	Solid Abdominal Organs	Global
Negative (no injury)	0	0	0	0	0
Trace to slight	3 to 4	2 to 3	2 to 4	1 to 2	1 to 4
Slight to moderate	3 to 4	2 to 4	3 to 4	2 to 4	2 to 4
Moderate to extensive	3 to 5	3 to 5	3 to 5	3 to 5	3 to 5

* Annex C gives more detailed injury descriptions.

Table 3.4 present the 'injury levels' with associated ASII and CWVP ranges (given by Axelsson and Yelverton) with the associated global AIS ranges (from Table 3.1).

Table 3.4: Injury Levels with Corresponding ASII and CWVP, and Estimated AIS Levels

Injury Level	ASII (-)	V (m/s)	AIS Range
Negative (no injury)	0.0 – 0.2	0.0 – 3.6	0
Trace to slight	0.2 – 1.0	3.6 – 7.5	1 to 4
Slight to moderate	0.3 – 1.9	4.3 – 9.8	2 to 4
Moderate to extensive	1.0 – 7.1	7.5 – 16.9	3 to 5
> 50% lethality	> 3.6	> 12.8	Up to 6

Considering that AIS 4 injuries might occur when the CWVP values is between 3.6 and 7.5 m/s (injury level: trace to slight), the acceptable CWVP tolerance value was set to 3.6 m/s. Based on the summary given in Table 3.3, this value is believed to represent a very low risk of AIS 2 injuries, which is in accordance with the guideline of 10% risk of AIS 2+.

3.6.2.4 Applicability and Limitations of Axelsson Model

Axelssson injury model can be used for the free field as well as the complex blast situation. It should be noted that the Axelsson injury model should be used together with the experimental techniques as described in his paper. However, practical problems are raised when introducing these techniques for the vehicle occupants, and modifications are necessary. This is further discussed in Chapter 4. Finally, the injury model by Axelsson predicts the injury severity, however it does not predict the injury risk.

3.6.3 Conclusions

In conclusion, the Axelsson & Yelverton model is the best available model for non-auditory blast injury assessment occurring during an AV blast mine strike. Considering that there are no risk curves available, it was decided to use a conservative approach and take the no injury level (3.6 m/s) as the limit for the chest wall velocity predictor.

Although it is known that the auditory system is the most vulnerable body region to blast overpressure, no pass/fail criterion has been proposed, because it is assumed that the crew will wear proper hearing protection. This will minimize the risk on temporary and permanent hearing injuries. Besides it should be noted that when the vehicle integrity is guaranteed during a mine strike, the overpressure in the vehicle should be low, resulting into low injury risks for the non-auditory organs.

3.7 SUMMARY ON INJURY CRITERIA AND TOLERANCE LEVELS

In this chapter the injury criteria and associated tolerance levels for the different body regions are discussed. The injury criteria and tolerance levels as advised by HFM-090/TG-25 and those to be used for injury assessment for vehicle mine protection are summarized in Table 3.5. More background can be found in the previous sections as well in the appendices.

Table 3.5: Summary of Injury Criteria and Tolerance Levels Proposed by HFM-090/TG-25 to be Used for Vehicle Mine Protection

Body Region	Injury Criteria	Tolerance Level	Signification	Specification
Lower leg	Peak lower tibia compression force (-Fz)	5.4 kN	10% risk of AIS 2+	Lower leg position should follow the recommendations of Chapter 4.
Thoraco-lumbar spine	Dynamic Response Index (DRIz)	17.7	10% risk of AIS 2+	Calculated with H3 pelvis vertical upward acceleration.
Neck	Compression force (-Fz)	4 kN @ 0 ms 1.1 kN @ 30 ms (see lower curve in Figure 3.16)	Serious (AIS 3) injuries are unlikely below the tolerance level	Measured at the H3 upper neck.
	Peak flexion bending moment (My)	190 N·m	Significant (AIS 2+) injuries are unlikely below the tolerance level	Measured at the H3 upper neck.
	Peak extension bending moment (-My)	57 N·m	Significant (AIS 2+) injuries are unlikely below the tolerance level	Measured at the H3 upper neck.
Non-auditory internal* organs	Chest wall velocity Predictor (CWVP)	3.6 m/s	No injury	Based on reflected pressure measurement follow the recommendations of Chapter 4.

* Respiratory system, gastrointestinal tract and solid intra-abdominal organs.

It should be noted again that there is no proposal for assessment of injuries caused by fragments and loose objects. So far, it has been assumed that these injuries will not occur due to preservation of vehicle integrity and prevention of flying objects.

Furthermore, it has been assumed that risk on auditory injuries will be negligible when wearing proper hearing protection. Therefore, injury assessment of auditory injuries has not been proposed.

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