Annex F – SUPPLEMENTAL INFORMATION ON THORACO-LUMBAR SPINE INJURY ASSESSMENT

F.1 INTRODUCTION

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. 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].

The probability of injuries due to loading mechanisms in dorso-ventral and lateral (X, Y) directions are quite low because of the predominance of axial (Z) load scenarios with a mine threat.

There is a lack of available detailed epidemiological data and PMHS data for the specific loading conditions as occur due to a mine strike under a vehicle. Therefore, test results and incidents with similar or at least comparable loading regimes were analyzed. The reader is referred to the main report (Chapter 3) for anatomy of the spinal thoraco-lumbar spine as well as for more general information on the injury assessment method.

The present annex is divided in 6 sections and focuses on the loads and tolerance levels in axial direction:

- Section F.1 – Introduction
- Section F.2 – Quasi-Static Strength
- Section F.3 – Dynamic Strength
- Section F.4 – Miscellaneous
- Section F.5 – Summary
- Section F.6 – References

F.2 QUASI-STATIC STRENGTH

Already in the mid 1920s Junk gave a few values on the compressive strength of vertebrae in his Tabulae Biologicae [Junk, 1925 – 1963]. Since detailed data on the boundary conditions (e.g. age) were not mentioned Geertz investigated PMHS between 19 and 46 years during WW II. Geertz tested the strength of individual vertebrae and vertebral complexes by shaping two end vertebrae for a central load transmission of a tension testing machine. The specimens were observed with X-ray pictures to identify the first peak of the stress-strain curve which was determined as an indication for fracture. The strength of the individual vertebrae (Table F.1) increases in caudal direction. As already pointed out in Section 3.3, Ruff summarized the investigations of Geertz [Ruff, 1950]. The lowest tolerance level for acceleration is determined for L1 which is also validated by operational data of US Army helicopter crashes [Shanahan and Shanahan, 1979] which have similar loading directions as Geertz investigations and mine strikes under vehicles.

Table F.1: Breaking Mass Load (m_F) as well as Maximum and Minimum Tolerance Level for Acceleration (a_F) [Ruff, 1950]

	m_F [kg]										a_F [g]	
	age [years]											
	19	21	21	23	33	36	38	43	44	46	min	max
T8	–	640		540	–	600	–	–	–	–	20.8	24.9
T9	–	–	–	610	–	720	–	700	–	–	21.0	24.9
T10	–	800	–	–	660	770	–	–	730	–	21.0	25.7
T11	750	–	720	–	–	–	–	860	–	755	20.8	25.1
T12	–	900	690	–	800	–	800	–	–	–	18.6	24.5
L1	720	–	840	–	–	–	–	900	800	800	18.2	23.0
L2	–	990	–	–	800	–	830	–	–	–	19.1	23.9
L3	900	–	–	–	–	–	–	940	–	1100	20.4	25.2
L4	–	–	–	–	1100	–	900	–	950	–	19.7	24.3
L5	–	1020	–	–	–	–	–	1000	–	1200	21.2	25.7

Additionally Ruff analysed the portion of the total weight supported by the individual vertebrae and calculated the tolerance level for acceleration a_F for each vertebral body with:

$$a_F = \frac{m_F}{\varepsilon \cdot m_G} - 1$$

where m_F is the breaking mass load, ε is the portion of body weight carried by the individual vertebrae and m_G is the total body weight (75 kg). The minimum and maximum values are listed in Table F.1. The related average breaking force of L1 is 7966 N.

Sonoda [Sonoda, 1962] investigated the vertebral strength by using 22 PMHS with an age between 22 and 79 years old. These investigations are also summarized in Yamada's Strength of Biological Material [Yamada, 1970]. Sonoda used wet isolated vertebral bodies and lined the deck plates with hard plaster to guarantee a uniform load transfer. The vertebral bodies were compressed with a tensile testing machine so that stress and strain values could be analysed. Table F.2 illustrates the breaking mass loads for specific vertebral bodies dependent on the age. Only the data which are related to an age between 20 and 39 years old are comparable to the data which were published by Ruff due to similar PMHS properties. Both data series show an increase in strength from T8 to L1, although the vertebral strength of Sonoda's data is round about 15% lower. Furthermore the decrease of vertebral strength from L1 to L5 seems to be questionable due to the geometrical properties of the specific vertebrae and is contradictory to Ruff's results. The general decrease in vertebral strength with increasing age is strongly correlated with the decrease of mineral density.

Table F.2: Breaking Mass Load (m_F) [Sonoda, 1962]

	m_F [kg]			
	age [years]			Mean value
	20 – 39	40 – 59	60 – 79	
T8	498	411	253	387
T9	567	431	263	420
T10	625	457	268	450
T11	678	486	273	479
T12	705	469	274	482
L1	755	477	291	508
L2	740	481	289	503
L3	744	484	310	513
L4	719	481	339	513
L5	691	461	311	488

Yoganandan et al investigated the compressive strength of vertebral bodies by using 38 PMHS [Yoganandan et al, 1988]. They positioned isolated vertebrae between two cylinders of a MTS (Material Testing System) device piston and compressed them uniformly to 50% of the original height. The applied load and deformation was recorded as a function of time. The first peak of the stress-strain curve was determined as an indication for fracture. Afterwards the force at failure was analysed by grouping vertebrae of the upper (T1 – T6) and lower thoracic (T7 – T12) as well as the lumbar spine (L1 – L5). The results are presented in Table F.3. The quite low values for failure forces seem to be allegeable due to variations in age of the PMHS and in the testing method. Based on the description of the specimen mounting it is possible that the primarily contact surface was reduced significantly by using a direct contact between the isolated vertebral bodies and the cylinders of the MTS device piston.

Table F.3: Breaking Force F and Further Properties at Different Regions of the Spine [Yoganandan, 1988]

	Force [N]	Pressure [MPa]	Deformation [mm]	Strain [%]
T1 – T6	2642 (*555*)	4.2 (2.7)	6.0 (1.8)	30.8 (9.0)
T7 – T12	3264 (*1211*)	3.2 (1.4)	4.9 (1.7)	23.4 (9.5)
L1 – L5	4590 (*2061*)	6.9 (3.2)	6.9 (3.2)	24.4 (9.5)

The focus of the investigations of Vesterby et al [Vesterby et al, 1991] was age-related effects on the compressive strength of the lumbar vertebrae. They compressed individual L2 vertebrae which were sawed 2 mm below the end-plates of 15 male and female PMHS with a material testing machine at a constant rate of 4.5 mm/min. The average breaking force of the vertebrae was 6289 N. The correlation of the compressive strength of the L2 vertebra with age and sex of the PMHS was significant.

Prasad et al [Prasad et al, 1974] focused on the significant weight bearing function of the articular facets during cranial acceleration based on approximately 70 runs with 4 PMHS. The axial force as well as geometrical force eccentricity was measured with a so-called intervertebral load cell between L3 and L4 and compared with the total spine load. Hence, the facet load was deduced. One conclusion of these investigations was the capability of the articular facets bearing compressive and tensile loads. Thus, the cranial load is transmitted via two paths, the vertebral bodies and the articular facets. The vertebral column is initially exposed to a compressive load. The facets tend to unload and cause a tensile load resulting in a forward flexion of head and torso. This flexion affect increasing loading on the anterior components of the vertebra by means anterior wedge fractures. These results are important for designing effective counter measures against cranial compressive loads to the vertebral column.

It is not clear if the aforementioned injury models did consider the effect of facet load during their investigations.

Furthermore Ripple and Mundie [Ripple and Mundie, 1989] proposed bending moment criteria including 1235 Nm for forward-flexion, 370 Nm for rearward-extension, 675 Nm for lateral bending. The only mentioned reference related to these values is Laananen [Laananen, 1980]. However, checking this reference does not give information on these bending moment criteria, but showed the DRIz criterion (DRIz is explained in main report Chapter 3, as well as in section Annex F.3). Based on engineering judgement it seemed that the momentum criteria were derived directly from the neck criteria as defined by Mertz and Patrick [Mertz and Patrick, 1967] with a factor of approximately 6.5. The background of this factor is unknown.

Chandler [Chandler, 1988] used the dynamic tolerance levels (see Annex F.3) to define a quasi-static tolerance level. He derived a quasi-static lumbar spine compression force limit of 6700 N (1,500 lb) based on a vertical (axial) DRIz value of 19 (operational data: 9% injury risk AIS 2+; laboratory data: 20% injury risk AIS 2+) using a Hybrid II ATD with a straight spine [Chandler, 1988]. This transfer from DRIz to force was done to overcome the general application problems of the DRIz for development and tests of civil aircraft seat and restraint systems. Chandler stated that the DRIz does not consider effects of restraint loads which are directly related to the stress in the thoraco-lumbar spine. Although the static compression force criterion is derived by the DRIz, deviations between DRIz and static compression force assessment have to be expected due to the fact that the DRIz analysis considers the dependency of injury risks and acceleration durations. The input for a DRIz analysis is the acceleration gradient which is directly related to durations. The input for the compression force criterion is only the maximum applied compression force value. This significant limitation has different impacts on air crashworthiness and AV mine protection applications. The impact load to seats in air crashworthiness incidents is characterized by only slight variations. Thus, the air crashworthiness standard provides only a specific standard impact impulse for tests. In opposite to this AV mine strikes provide a huge variation of impact loads due to different materials of the vehicle structure, as well as variations in the threat conditions and deploy conditions (e.g. depth of burial, soil conditions). The variation of this relatively huge number of variables results in significantly different loading regimes. Hence, the duration of the loads becomes more and more important.

The relationship between axial force and DRIz for full scale mine trails performed in DEU and NLD was studied and compared to data from aerospace [Dosquet, 2003]. In general a quite sufficient linear relationship between DRIz and measured force was seen. Future work should focus on updating the relationship, using more recent data from several countries.

Rapaport et al [Rapaport et al, 1997] tried to validate the force criterion evaluated by Chandler [Chandler, 1988] for military seat applications. The team used a similar procedure as Chandler and derived a force criterion based on the Eiband tolerance curve [Eiband, 1950] (see Annex F.3) and the DRIz model by using a FAA-Hybrid III ATD on a sled platform normalized for the occupant weight. The peak acceleration of the crash pulse was varied between 10, 20 and 30 g with a 6.1, 7.6 and 9.1 m/s run each. The proposed values are 8632 N for the 5th percentile Hybrid III, 9656 N for the 50th percentile Hybrid III and 11124 N for the 95th percentile Hybrid III. However, it is not directly stated which DRIz value is comparable with these values.

F.3 DYNAMIC STRENGTH

The dynamic properties of the thoraco-lumbar spine are of special interest for mine protection of vehicles properties since the peak accelerations and forces as well as durations vary significantly for various threats and different vehicle structures. This section describes different injury models which take the dynamic strength into account.

The first model which considered the dependency of load tolerance limits on duration was presented by Ruff [Ruff, 1950]. He postulated a tolerance limit for durations between 5 ms and 1 s of 18 to 23 g which reflect the lower and upper limits of the investigations by Geertz for L1 (Figure F.1). The tolerance limit for durations less than 5 ms was approximated on the basis of data acquired in catapult test accidents. The seat acceleration came up to 26 g for 5 ms in these accidents. The tolerance is determined by the dynamic properties of the bones and the intervertebral disks. Hence, the decrease in acceleration is related to an increase of duration due to the fact that some distance must be obtained in a breaking process.

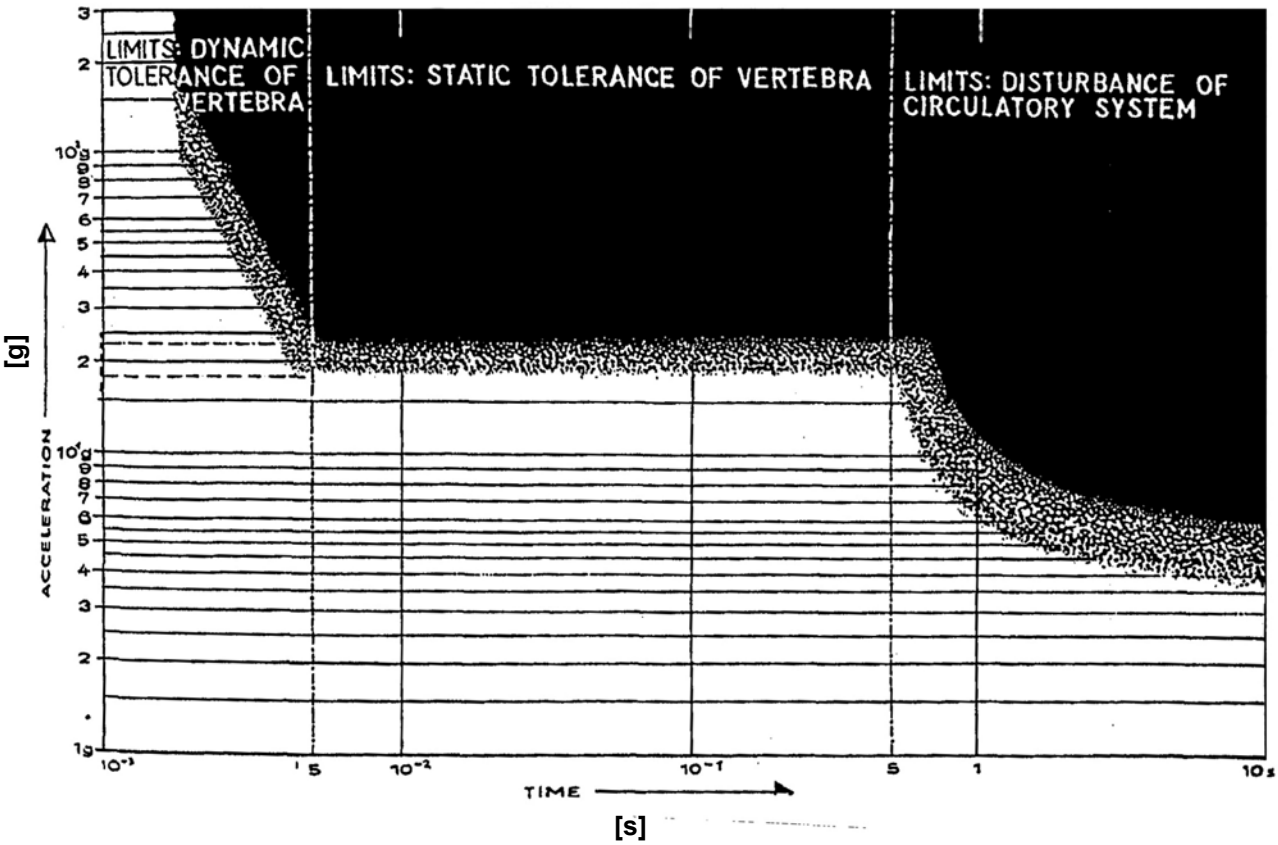

Figure F.1: Tolerance Curve for Vertebrae [Ruff, 1950].

Eiband summarized available acceleration limits related to different durations for a trapezoid pulse of ejection seat experiments in his memorandum to model the spinal strength [Eiband, 1959] (Figure F.2). Human volunteers tolerated uniform seat accelerations of 16 g for durations up to 40 ms in catapult seat experiments. The tolerance levels of human volunteers for longer durations were also confirmed in these experiments. The 110 g limit for 2 ms as well as the 42 g limit for 7 ms was established in hog (domestic pig) experiments. The animals did not suffer any permanent injury. Chimpanzees sustained 42 g for 48 ms and 28 g for 140 ms without permanent injury.

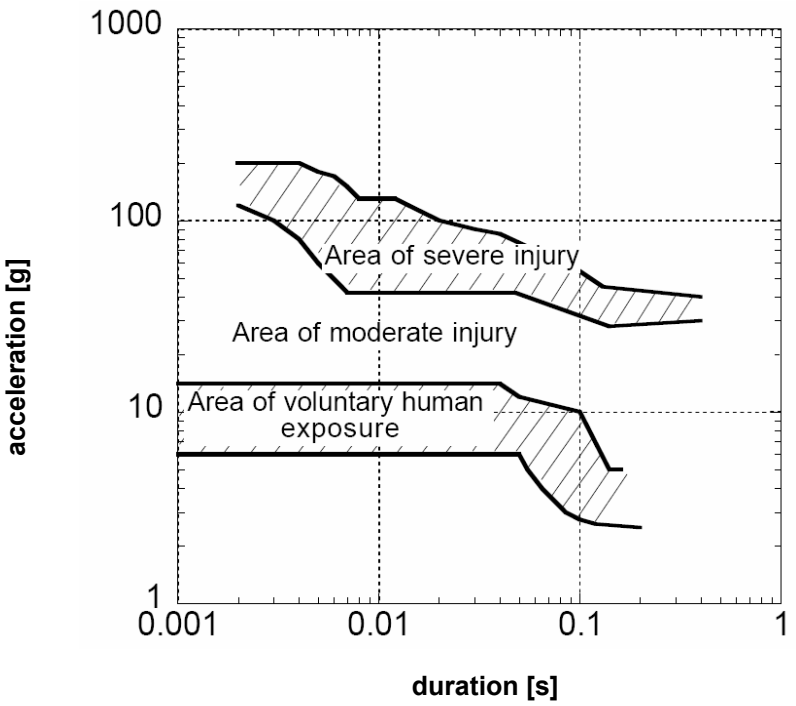

Figure F.2: Eiband Tolerance Curve [Eiband, 1959].

In 1964 Hirsch [Hirsch, 1964] presented a tolerance curve of seated unrestrained men in upright position to shock motion of short durations as a design guideline for the protection of ship personnel (Figure F.3). He calculated the tolerance curve on the baseline of Kornhauser's sensitivity analysis approach [Kornhauser, 1954]. Kornhauser assumed that the sensitivity analysis which is valid for inorganic structures is also applicable for animate structures. Hence, the maximum tolerable load of a single degree of freedom mechanical system is related to peak acceleration if the pulse duration is long in comparison to the natural period and to a peak velocity change if the duration is short related to the natural period. The results of Ruff were used to estimate to tolerance curve for long durations. By postulating a uniform strength of the lower vertebral bodies the appropriate load on L5 is 60% of the whole body weight. Thus, using the 1400 lb limit for vertebral strength the acceleration threshold for long durations is 15 g. The threshold will be 30 g by using the 2900 lb upper limit of Ruffs investigations (dashed line). The time when the minimum distance between rib cage and the platform of a ship shock motion simulator occurs was estimated as 40 ms derived by shock tests with volunteers. This time is taken as a quarter period of the natural frequency. Thus, the natural frequency f is about 6 Hz for a seated man. The peak velocity v_0 was determined by using Kornhauser's equation with the static acceleration sensitivity g_0:

$$v_0 = 5.13 \cdot \frac{g_0}{f}$$

Hence, the peak velocity change threshold is 13 fps (~ 4 m/s) or 26 fps (~ 8 m/s, dashed line). Overall, the significant assumption in the Hirsch model causes a substantial simplification of the real biomechanical response of a seated man.

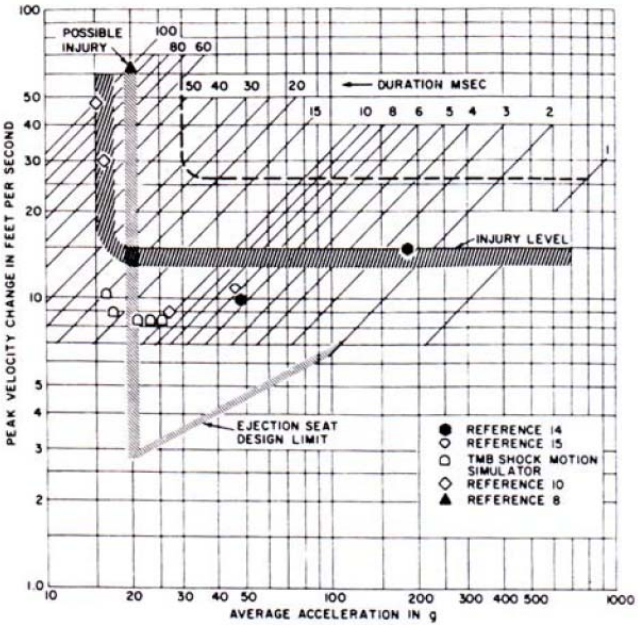

Figure F.3: Tolerance of Seated Men to Shock Motion of Short Duration [Hirsch, 1964].

TTCP [Tremblay et al, 1998] proposed a quasi-static value of 3800 N and a linear interpolation between 3800 N (t = 30 ms) and 6673 N (t = 0 ms) for vehicular mine protection applications by referencing the previous mentioned report of Ripple and Mundie [Ripple and Mundie, 1989] (Figure F.4), but the tolerance values are not described in that reference. However, these criteria seem to be derived from the Mertz criteria for neck loads [Mertz and Patrick, 1967] by using a factor of 3.4 – 3.8. This factor reflects the ratio of the geometrical contact surface area of lumbar and cervical vertebrae (~ 1550/420 mm^2).

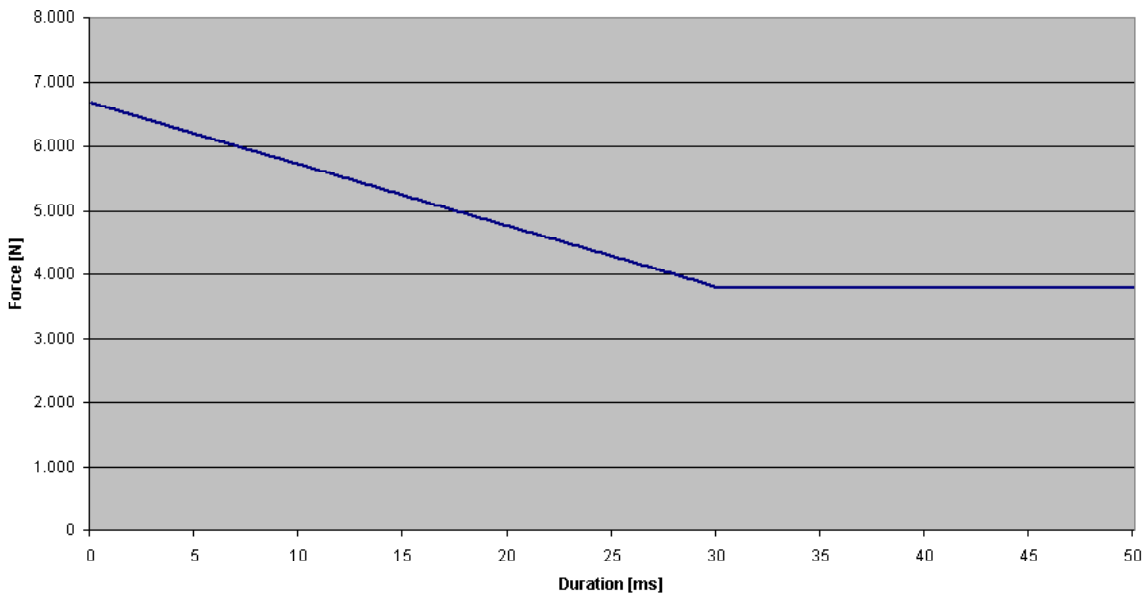

Figure F.4: Lumbar Spine Compression Force Criterion [TTCP, 1998].

In 1957, Latham introduced a general model to describe the impact of ejection seat to the human body [Latham, 1957]. Stech and Payne [Stech, 1969] evaluated this model as a general model to simulate the biomechanical response due to human body dynamics by using a single mass-spring-damper system (Figure F.5). The model is called: Dynamic Response Index (DRI). The DRIz is the DRI model for the vertical direction.

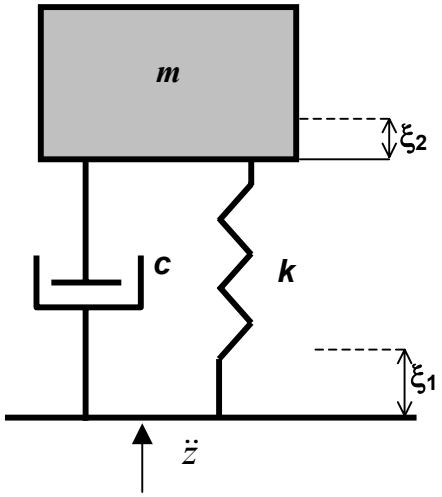

Figure F.5: DRIz Model [Stech, 1969].

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 $\zeta = \dfrac{c}{2 \cdot m \cdot \omega_n}$; and
- 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 ξ and ω_n, 0.224 and 52.9 rad/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 F.6 as the laboratory data curve.

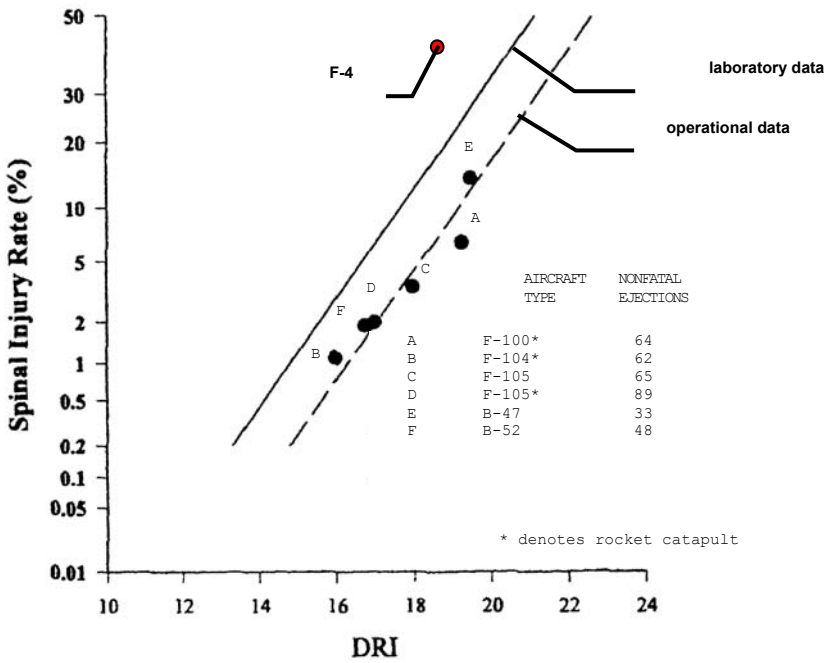

Figure F.6: Spinal Injury Risk Calculated from Laboratory and Operational Data Supplemented with F-4 Operational Data [Brinkley, 1970] Valid for AIS 2+ Injuries.

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 F.6 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.

The physical parameter which affects fracture is always force. Using a model which is based on another physical parameter causes less accuracy. However force based injury models taking the visco-elastic properties of the vertebral column into account are not available yet.

F.4 MISCELLANEOUS

In relation to the cranio-caudal (axial) direction only a little information is available for the lateral and dorso-ventral loading direction of the thoraco-lumbar spine. However, as stated earlier [Dosquet, 2003,

2004], it is expected that the injury risks in these lateral and dorso-ventral directions are low when a vertical load is generated by a mine strike under the vehicle.

Brinkley et al [Brinkley et al, 1989] introduced a general injury-risk criterion based on the DRI values related to lateral (Y), dorso-ventral (X) and cranio-caudal (Z) loading direction. These values have to be calculated with the measurement data and the limit values DRI_L for specific injury risks (low: 0.5%, moderate: 5%, high: 50%) and the three loading directions:

$$\beta = \sqrt{\left(\frac{DR_x}{DR_{x_L}}\right)^2 + \left(\frac{DR_y}{DR_{y_L}}\right)^2 + \left(\frac{DR_z}{DR_{z_L}}\right)^2}$$

The defined injury risk will be exceeded if β is greater than 1 for the related DRI limits, but this kind of analysis does not consider the different characteristics of the injury risk curves for the three loading directions. The risk curves represent different exponential functions which will have a significant impact on the quotients.

Furthermore the validation of DRI related to lateral and dorso-ventral loading directions is only rudimental. The evaluation of the DRI in dorso-ventral directions is based on tests that were not specifically designed for this purpose. Hence, there is a wide variation in experimental methods and measurements. For example, most of the time to peak acceleration values for the +X direction was between 20 to 50 ms with some data in the range between 8 and 10 ms. The data for the –X direction ranged from 25 to 160 ms. By using a half-sinewave approximation technique the natural frequency and the damping coefficient in –X direction was calculated with 62.8 rad/s and 0.2. These results were confirmed by a transfer function analysis with data from 11 impact tests without dynamic preload (54.5 rad/s and 0.26). By using data of tests with human subjects for evaluating the risk of injury levels the authors choose a conservative approach.

The most critical part of the general injury risk criterion is the calculation and validation in lateral direction. Damping coefficient and natural frequency were calculated by using only one set of data with 13 human subject tests at 8 g deceleration level, an impact velocity of 8.84 m/s and a time to peak acceleration of 22 ms. The injury risk levels were based only on existing expert opinions and available data with a poor applicability.

F.5 SUMMARY

Considering relative low injury risks due to lateral and dorso-ventral loads HFM-090/TG-25 recommended an injury assessment only for the vertical (axial, z) loading direction of the thoraco-lumbar. Loads with durations less than 10 ms presuppose injury risk models which consider the dynamic behaviour of the thoraco-lumber 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-lumber spine disregard the duration dependency or are not validated. The acceleration measurements in the pelvis of the anthropomorphic test device have to be used as input for the DRIz. This is also the most consistent and practical way of measuring the input accelerations for the DRIz model. Based on discussions on the risk curves available in literature, TG-25 decided to use the more conservative risk curve derived from laboratory data. When using this curve (Figure F.6) the tolerance level of 17.7 for DRIz refers to a 10% risk of AIS 2+ injuries.

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