Journal of Sound and Vibration (1998) **215**(4), 723–741
Article No. sv981601

ON THE HEALTH RISK OF THE LUMBAR SPINE DUE TO WHOLE-BODY VIBRATION—THEORETICAL APPROACH, EXPERIMENTAL DATA AND EVALUATION OF WHOLE-BODY VIBRATION

H. SEIDEL, R. BLÜTHNER, B. HINZ AND M. SCHUST

Federal Institute for Occupational Safety and Health, *Research Group AM* 4.3
Biological Effects of Vibration and Noise, *Nöldnerstr*. 40–42, *D*-10317 *Berlin*, *Germany*

(*Accepted* 18 *March* 1998)

The guidance on the effects of vibration on health in standards for whole-body vibration (WBV) does not provide quantitative relationships between WBV and health risk. The paper aims at the elucidation of exposure–response relationships. An analysis of published data on the static and dynamic strength of vertebrae and bone, loaded with various frequencies under different conditions, provided the basis for a theoretical approach to evaluate repetitive loads on the lumbar spine (“internal loads”). The approach enabled the calculation of “equivalent”—with respect to cumulative fatigue failure—combinations of amplitudes and numbers of internal cyclic stress. In order to discover the relation between external peak accelerations at the seat and internal peak loads, biodynamic data of experiments (36 subjects, three somatotypes, two different postures—relaxed and bent forward; random WBV, a_w, r.m.s. 1·4 ms^{-2}, containing high transients) were used as input to a biomechanical model. Internal pressure changes were calculated using individual areas of vertebral endplates. The assessment of WBV was based on the quantitative relations between peak accelerations at the seat and pressures predicted for the disk L5/S1. For identical exposures clearly higher rates of pressure rise in the bent forward compared to the relaxed posture were predicted. The risk assessment for internal forces considered the combined internal static and dynamic loads, in relation to the predicted individual strength, and Miner’s hypothesis. For exposure durations between 1 min and 8 h, energy equivalent vibration magnitudes (formula B.1, ISO 2631-1, 1997) and equivalent vibration magnitudes according to formula B.2 (time dependence over-energetic) were compared with equivalent combinations of upward peak accelerations and exposure durations according to predicted cumulative fatigue failures of lumbar vertebrae. Formula B.1 seems to underestimate the health risk caused by high magnitudes, formula B.2 is recommended for the evaluation of such conditions.

1. BACKGROUND

The guidance on the effects of vibration on health in the present standard for whole-body vibration (WBV) [1] does not provide quantitative relationships between vibration exposure and the risk of health effects, because sufficient data are missing. Current gaps of knowledge concern also the effect of WBV containing high acceleration events on the lumbar spine, the influence of different sitting postures, and the variability of WBV effects caused by the individual variability of the skeleton Sandover [2] suggested fatigue failure of vertebral endplates as the pathogenetic mechanism that causes subsequent degenerative changes of the lumbar spine. Based on this hypothesis, the paper is aimed at the

0022–460X/98/340723 + 19 $30.00/0

elucidation of exposure–response relationships in order to derive quantitative relations for the assessment of the health risk due to WBV. For that purpose, data published in literature (cf. [3] for a review) on the ultimate strength of lumbar vertebrae or vertebral segments and on the endurance limit of bone were compared with results of human experiments [3].

2. METHOD

2.1. ANALYSIS OF STRENGTH DATA

A comprehensive analysis of published data on the static and dynamic strength of vertebrae and bone, loaded, with various frequencies under different conditions, provided the basis for a theoretical approach to evaluate repetitive loads on the lumbar spine ("internal loads"). During occupational WBV, a combined static and dynamic load acts on the spine. The internal static load is determined by the body mass and posture, the dynamic load is caused by WBV. Data on the ultimate strength of two groups of authors [4, 5] were selected which provided detailed data on all cases tested. The data were carefully checked and cases with possible errors or pathological changes were eliminated [3]. The reported ultimate compressive forces were transformed into pressure using the surface areas of vertebral endplates reported by the authors. Any internal static compressive stress x_{int} (in MPa) can be expressed as normalized internal static stress x_n according to

$$x_n = x_{int}/x, \tag{1}$$

with x = ultimate compressive strength in MPa. (A list of symbols and abbreviations is given in the appendix.)

Data on the endurance limit of vertebrae in the high cycle range were not available, therefore, results of testing other bones had to be used [6, 7]. Without an additional static load, the fatigue damage F was assumed to be a linear function of the number of cycles according to

$$\sum n_k/N_k = F. \tag{2}$$

(Palmgren—Miner approach), with n_k = number of internal cyclic peak-to-peak loads with a peak-to-peak amplitude k and N_K = number of cycles to failure at k.

The Goodman relation [8] was applied to re-calculate the published data in order to consider the reduction of the resultant strength when a static stress x_{int} is combined with an internal variable stress y_{int} in MPa. Any y_{int} can be expressed as normalized internal cyclic stress y_n according to

$$y_n = y_{int}/y, \tag{3}$$

with y = endurance limit in MPa.

By derivation from the Goodman relation [8, Figure 42.26], the combined normalized steady x_n and variable stresses y_n were expressed as an "equivalent cyclic stress" y_{eqk}:

$$y_{eqk} = x_n + y_n. \tag{4}$$

According to the conservative Goodman relation, the fatigue behaviour caused by y_{eqk} is predicted to be equal to a behaviour that would be caused by a sole variable stress.

The calculation of a normalized variable stress y_n, i.e., the ratio variable stress divided by the endurance limit, required a reasonable assumption with regard to the relation between the endurance limit and the ultimate strength. In order to estimate this relation, non-linear curve fitting and regression [3] were applied to the relations between the number of cycles to failure N_k and various ratios between the endurance limit y and the ultimate strength limit x, until an equivalent cyclic stress of 1 was realted to a number of about 10^6 cycles that was considered as endurance limit.

2.2. HUMAN EXPERIMENTS

2.2.1. *Vibration exposure*

The exposure was applied by means of a displacement controlled, electro-hydraulic vibrator (Hydropuls) modified for human experiments, with a maximum stroke of 400 mm and maximum force 10 kN. A total mass of approx. 200 kg (seat construction with four steel plates and piston rods) served as a mechanical safety measure against excessive upwards acceleration. An emergency stop button was provided on the control panel which when actuated by the subject interrupted the test and caused the seat to move slowly and smoothly to the static set point. In addition, a mechanical short-circuit valve (Multiventil) limited the differential pressure and thus the peak acceleration against gravity to 12 ms^{-2}. The test rig was controlled by a computer and a 12 bit ADC (DASH 16, Metrabyte) using the HVLab software (ISVR Southampton) at a sampling frequency of 500 Hz. Digital channels were used for the link between the computer and the electronic monitoring system. The connection of a digital channel to a synchronisation unit made it possible to start the exposure by external triggering.

The simulation of a WBV signal obtained from field measurements (365 kilowatt, tracked hydraulic excavator while loosening rocks from a pile of detached rock) required the creation of a displacement control signal which took the transfer function of the Hydropuls test rig into account. The latter was initially determined with a random signal for the seat and a mechanical model of a human [9]. The transfer function for the seat construction and a subject was subsequently determined. A subroutine, which took into account the previously determined specific transfer function for the test rig, generated the displacement control signal for the acceleration to be simulated. Comparison of the spectral density distributions of the in-field acceleration signal and the simulated exposure gave very good agreement. The main frequency content was located between 1 and 8 Hz with the maximum near 3 Hz. Within the recordings of actual acceleration of 65 s, four 10 second long data blocks (see Table 1), separated from each other in time, were selected for data processing and frequency weighting in accordance with reference [10].

TABLE 1

Characteristic values of unweighted (a) *and frequency weighted* (*after ISO* 2631, [10]) *seat acceleration* (a_w) *in* [ms^{-2}] *for the four* 10 *s data blocks* S2, S3, S5, S8, *analyzed*; *r.m.s.*, *root mean square*; *d*, *extreme acceleration downward*; *u*, *extreme acceleration upward*

	a_{rms}	$a_{w,rms}$	a_d	$a_{w,d}$	a_u	$a_{w,u}$
S2	1·75	1·50	12·13	7·98	−10·37	−12·93
S3	1·90	1·64	11·06	6·04	−10·53	−11·72
S5	2·42	2·07	7·88	8·07	−7·19	−7·80
S8	1·89	1·54	8·54	5·07	−6·56	−7·56

2.2.2. *Subjects*

36 young healthy male subjects of different somatotype were selected as a step towards answering the question concerning representative inter-individual variability of the effects. All subjects were informed about the aims of the experiment and gave their informed consent. A "robust–frail" variation ranking was used to characterise the robustness of the skeleton of the population adequately. The Humeral Index (HI) from Greil [11] was preferred for this ranking [12]:

$$HI = (\text{elbow width/upper arm length}) \times 100 \tag{5}$$

12 subjects were selected for each robustness group such that there were significant differences in the means for the HI between all three groups (see Table 2). Discal areas at the L3/4, L4/5 and L5/S1 levels [12, 13] were different between the frail and robust groups. There were, however, no significant differences in the body mass of the subjects selected. Further details on anthropometric characteristics are given in references [3, 12].

2.2.3. *Experimental procedure and data acquisition*

The subjects sat on a hard, anatomically formed seat without a backrest. Two postures were examined: posture D—steering wheel held with both hands, upper torso relaxed, loose, upright and subjectively comfortable posture; posture BF—upper torso bent forward with a stretched lumbar spine, tense back muscles and with the pelvis tipped forward. Appropriate spherical control elements, side mounted below the plane of the seat, were grasped with the hands, back of the hand ventrally directed, elbow bent, and resting on the operating elements was to be avoided. For the motion analysis system (MacReflex, Qualysis), markers on triangular shaped polystyrene tags were fastened to the subject's back. With the aid of a video, the tags enabled the subject to maintain a posture constant throughout the test period. Seat acceleration was measured by one transducer BWH 101 (Metra) that could be calibrated statically by using the $\pm 1\,\boldsymbol{g}$ method. In addition, the signs of the acceleration were dynamically assigned with respect to the sign of the displacement for a sinusoidal test signal.

The resultant force FZ at the interface between the subject and the seat, was measured by using three type KWH 100 (Messelektronik Dresden) load cells arranged in the form of an isosceles triangle. For an operating voltage of 4 V the sensitivity of the load cells is 1 mV/9·81 N. The determination of the summed force was carried out using special hardware. Before beginning the exposures, the pertinent mean values of FZ were determined for each test person, in the two defined postures, under static conditions, over a four s measurement period. This value corresponds to the gravitationally induced force of the body mass supported by the seat. The sign of this force was negative due to the measuring arrangements. Acceleration and force signals were conditioned, digitized and stored by using a 22 channel, battery operated, measurement recording system SCADASII (DIFA, Netherlands) in conjunction with an Amstrad 286 Laptop (sampling rate 1 ms). With the help of MacReflex (Qualisys) motion analysis system and a Macintosh IIsi computer, the co-ordinates of the marks positioned over the spinous processes of T5, T11, S1, S3 and two marks on the test seat were measured with a frequency of 50 Hz during the exposure. Using reference measurements, the co-ordinates of the spinous processes L3, L4 and L5 were calculated [3]. The position of the centre of gravity and the midpoints of the intervertebral disks were calculated with the aid of the anthropometric measurements previously made on each subject, and data from the literature [3]. The precise temporal synchronisation of the different systems used in the tests for the control of the vibrator and data acquisition were synchronized to a tolerance of $\pm 0{\cdot}4$ ms [3].

TABLE 2

Selected measured anthropometric characteristics of the frail, robust and average (interm.) subjects (S): *BM = Body mass in* kg, *BH = height to temples in* cm, *Age* (*in years*), *HI = humeral index*, *BF*(L5) = *surface area of the intervertebral disc* L5/S1 *in* cm^2, *MV = mean value*, *SD = standard deviation*, *interm = intermediate*

S no.	Type	*BM*	*BH*	Age	*HI*	*BF*(L5)	S no.	Type	*BM*	*BH*	Age	*HI*	*BF*(L5)	S no.	Type	*BM*	*BH*	Age	*HI*	*BF*(L5)
100	frail	76·8	183·2	23	17·34	18·79	202	robust	63·0	173·5	19	24·01	19·00	113	interm	74·4	175·2	21	20·50	19·28
121	frail	69·4	175·8	23	18·46	18·57	213	robust	65·6	168·3	21	23·83	19·54	171	interm	79·6	169·4	23	20·75	19·31
131	frail	73·2	172·1	23	18·24	18·37	243	robust	64·8	166·6	21	23·23	17·98	176	interm	82·0	180·2	20	20·82	20·59
139	frail	71·2	168·7	20	18·34	16·26	249	robust	67·6	167·1	22	23·57	18·84	206	interm	61·6	176·9	20	21·19	19·76
151	frail	59·2	166·3	22	18·32	15·04	261	robust	73·4	171·4	20	23·32	19·60	224	interm	72·8	175·6	19	21·21	19·47
156	frail	74·6	176·7	20	16·67	17·81	282	robust	72·4	173·7	21	23·01	21·41	235	interm	67·0	170·6	19	20·95	18·63
162	frail	64·4	173·8	25	18·48	17·18	318	robust	64·6	167·9	21	23·20	19·25	277	interm	57·0	182·8	19	20·59	20·03
173	frail	59·2	170·2	22	18·15	16·47	329	robust	58·4	165·7	20	22·95	18·64	301	interm	64·8	173·6	22	20·96	19·10
185	frail	59·8	165·6	23	18·37	16·76	333	robust	71·0	172·2	21	23·44	22·00	308	interm	63·6	170·9	21	20·77	19·15
207	frail	71·6	176·7	21	17·91	19·14	349	robust	80·6	176·7	19	23·99	23·97	311	interm	70·8	174·0	21	20·90	19·46
401	frail	58·2	178·1	20	18·54	18·41	350	robust	59·6	164·3	21	23·10	18·77	321	interm	78·0	186·3	19	20·61	22·09
403	frail	49·0	163·3	42	17·57	16·54	361	robust	74·4	186·2	22	23·35	23·88	322	interm	57·2	172·8	20	21·21	18·72
MV	frail	67·1	173·4	23·7	18·08	17·53	MV	robust	67·9	171·1	20·7	23·42	20·24	MV	interm	69·1	175·7	20·3	20·87	19·63
SD	frail	7·0	5·4	6·0	0·58	1·28	SD	robust	6·6	6·0	1·0	0·36	2·06	SD	interm	8·5	5·1	1·3	0·25	0·94

differences in the means for the *HI* between all three groups (see Table 2). Discal areas at the L3/4, L4/5 and L5/S1 levels [12, 13] were different between the frail and robust groups. There were, however, no significant differences in the body mass of the subjects selected. Further details on anthropometric characteristics are given in references [3, 12].

2.2.3. *Experimental procedure and data acquisition*

The subjects sat on a hard, anatomically formed seat without a backrest. Two postures were examined: posture D—steering wheel held with both hands, upper torso relaxed, loose, upright and subjectively comfortable posture; posture BF—upper torso bent forward with a stretched lumbar spine, tense back muscles and with the pelvis tipped forward. Appropriate spherical control elements, side mounted below the plane of the seat, were grasped with the hands, back of the hand ventrally directed, elbow bent, and resting on the operating elements was to be avoided. For the motion analysis system (MacReflex, Qualysis), markers on triangular shaped polystyrene tags were fastened to the subject's back. With the aid of a video, the tags enabled the subject to maintain a posture constant throughout the test period. Seat acceleration was measured by one transducer BWH 101 (Metra) that could be calibrated statically by using the $\pm 1\,\boldsymbol{g}$ method. In addition, the signs of the acceleration were dynamically assigned with respect to the sign of the displacement for a sinusoidal test signal.

The resultant force *FZ* at the interface between the subject and the seat, was measured by using three type KWH 100 (Messelektronik Dresden) load cells arranged in the form of an isosceles triangle. For an operating voltage of 4 V the sensitivity of the load cells is 1 mV/9·81 N. The determination of the summed force was carried out using special hardware. Before beginning the exposures, the pertinent mean values of *FZ* were determined for each test person, in the two defined postures, under static conditions, over a four s measurement period. This value corresponds to the gravitationally induced force of the body mass supported by the seat. The sign of this force was negative due to the measuring arrangements. Acceleration and force signals were conditioned, digitized and stored by using a 22 channel, battery operated, measurement recording system SCADASII (DIFA, Netherlands) in conjunction with an Amstrad 286 Laptop (sampling rate 1 ms). With the help of MacReflex (Qualisys) motion analysis system and a Macintosh IIsi computer, the co-ordinates of the marks positioned over the spinous processes of T5, T11, S1, S3 and two marks on the test seat were measured with a frequency of 50 Hz during the exposure. Using reference measurements, the co-ordinates of the spinous processes L3, L4 and L5 were calculated [3]. The position of the centre of gravity and the midpoints of the intervertebral disks were calculated with the aid of the anthropometric measurements previously made on each subject, and data from the literature [3]. The precise temporal synchronisation of the different systems used in the tests for the control of the vibrator and data acquisition were synchronized to a tolerance of $\pm 0{\cdot}4$ ms [3].

2.2.4. *Calculation of internal loads*

The model used for the calculation of internal compressive loads contains the following assumptions: *FZ* is closely related to the vertical force exerted by the body segment above the intervertebral disc, whereby the ratio between this force and *FZ* was assumed to be the same as that between the mass above the intervertebral disc and the body mass supported by the seat. The mass above the intervertebral discs L3/L4, L4/L5, and L5/S1 was estimated for each subject based on his anthropometric characteristics in connection with regression equations and proportions published in [3, 14, 15]. Further assumptions were as follows. There are no spring and damper elements between the seat and the intervertebral disc. There are no rotational movements and the muscle forces at the respective level of the spine guarantee an equilibrium of force moments in the sagittal plane. The time history of the compressive force acting, e.g., on the disc L5/S1 was calculated by

$$C_{\mathrm{L5/S1}} = FZ_{korr}q_{\mathrm{L5/S1}}(\cos\theta_{\mathrm{L5/S1}} + |w_{\mathrm{L5/S1v}}/m_{\mathrm{L5/S1}}|), \tag{6}$$

with $C_{\mathrm{L5/S1}}$ = compressive force acting on the intervertebral disc L5/S1, $FZ_{korr} = FZ$ reduced by the force which is exerted by the accelerated interface, with this force calculated as the product of the mass of the interface and its acceleration, $q_{\mathrm{L5/S1}}$ = factor for the reduction of the total mass on the seat to a partial mass above L5/S1, $\cos\theta_{\mathrm{L5/S1}}$ = cosine of the angle between the vertical and the normal to the intervertebral disk L5/S1, $w_{\mathrm{L5/S1v}}$ length of ventral lever arm for the partial mass of the body above disk L5/S1 and $m_{\mathrm{L5/S1}}$ = length of the back muscle lever arm estimated individually [3].

Co-ordinates of the centre of gravity of the torso were determined according to reference [16]. Values published in references [17–19] were used for the calculation of lever arms of the muscles (m. erector spinae) and ligaments at the level of the spinous processes of L3, L4 and L5 [12].

An averaging of the compressive force and pressure time histories was carried out separately for the three subject types and summed up over all 36 subjects. The standard deviations were also calculated. The curves of mean compressive force and pressure with their respective standard deviations served as the basis of a determination of the relationship between extremes of the compressive force or pressure and the weighted (a_w) [10] and unweighted seat acceleration. 135 visually determined time windows were used to enable an unambiguous, automatic, identification of valid temporal correlations for these extremes. Figure 1 illustrates this procedure.

Statistical calculations were performed by the SPSS-PC for Windows package.

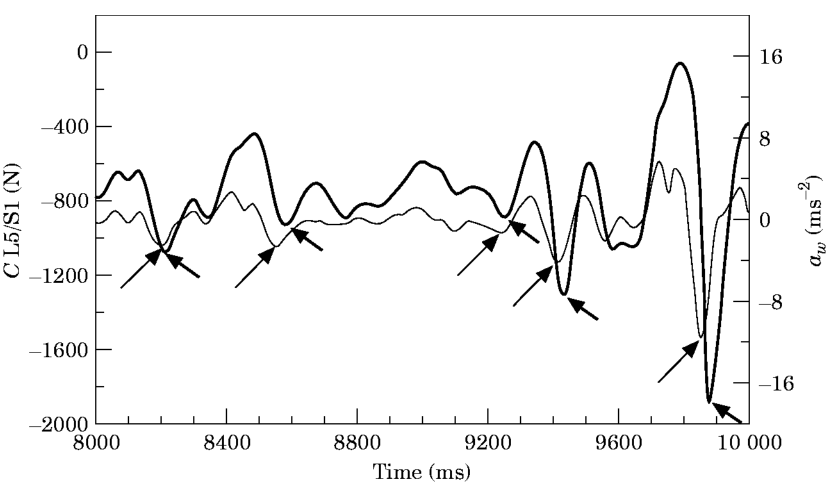

Figure 1. Example (bent forward posture I3, segments, 8000–10000 samples) of the method used to determine the extreme values for the mean ($N = 36$) calculated compressive force (C) time history for the L5/S1 intervertebral disc (bold line). Pertinent frequency weighted (ISO 2631/1-1985 [10]) seat acceleration a_w (thin line). The extreme values used in the analysis are marked by ↗ (a_w) and ↖ (C). The time windows used to find the extreme values were 8100–8300, 8500–8700, 9200–9300, 9350–9500 and 9800–9950.

3. RESULTS

3.1. STRENGTH DATA

3.1.1. *Ultimate static strength*

Regrettably, strength limits have, previously, rarely been expressed as pressures [20] and therefore the data of references [4, 5], which appeared to be the most reliable, were chosen for reference purposes. This decision was also taken because of the possible danger that tests on a single vertebrae could cause the ultimate static strength to be under or over estimated as reviews of ultimate static strength show [21].

If preparations derived from persons whose medical histories show "maximum reduced physical activity in the last 2 years" are excluded, the data of reference [5] yield an average compressive strength of 3·65 MPa (standard deviation $\pm 1{\cdot}22$ MPa) for the remaining 68 vertebral segments. The 10th and 90th percentile limits were 2·31 and 5·29 MPa respectively. The age of the deceased varied from 19 to 79 years (46·65 $\pm$ 14·45 years). The intermediate ages and standard deviations were 46 $\pm$ 13 years for the 39 male segments and 47 $\pm$ 16 years for the 29 female segments. Gender and level in the spine factors had no significant influence on the compressive strength (ANOVA). While the disc area increased significantly from cranial to caudal, the mineral density remained roughly constant [5]. Age had a significant influence both for men and women collectively as well as for men alone. The regression equations for the estimation of the ultimate compressive strength x were

$$x[\mathrm{MPa}] = -0{\cdot}055141 \times \text{years} + 6{\cdot}224754, \tag{7}$$

$$\text{coefficient of determination} = 0{\cdot}44, \text{ men and women,}$$

$$x[\mathrm{MPa}] = -0{\cdot}061148 \times \text{years} + 6{\cdot}643324, \tag{8}$$

$$\text{coefficient of determination} = 0{\cdot}55, \text{ men.}$$

Thus it is estimated that from the age of 20 to 60 the compressive strength for men decreases by about 2·45 MPa.

Hansson *et al*. [4] investigated lumbar vertebrae from 15 men (average age 59·7 years) and 21 women (average age 57·7 years). In both cases the age span was large. The mean age and standard deviation for the 45 male vertebrae examined were 58 $\pm$ 15 years respectively. Corresponding values for the 64 female vertebrae were 58 $\pm$ 10 years. The compressive loads were only applied to vertebrae if it had been possible to maintain a 3 mm piece of disc on both end plates during the preparation and thus achieve the most realistic application of the force possible. The average compressive strength was 2·29 MPa (standard deviation $\pm 0{\cdot}87$ MPa). The 10th and 90th percentiles limits were 1·20 and 3·44 MPa respectively. Age also had a significant influence on compressive strength in these investigations. The regression equations for calculation of the ultimate compressive strength x were

$$x[\mathrm{MPa}] = -0{\cdot}032836 \times \text{years} + 4{\cdot}191814, \tag{9}$$

$$\text{coefficient of determination} = 0{\cdot}22, \text{ men and women,}$$

$$x[\mathrm{MPa}] = -0{\cdot}032537 \times \text{years} + 4{\cdot}412940, \tag{10}$$

$$\text{coefficient of determination} = 0{\cdot}29, \text{ men.}$$

Thus the compressive strengths of reference [4] are, even when the influence of age is taken into consideration, clearly lower than the values found in reference [5], particularly in the

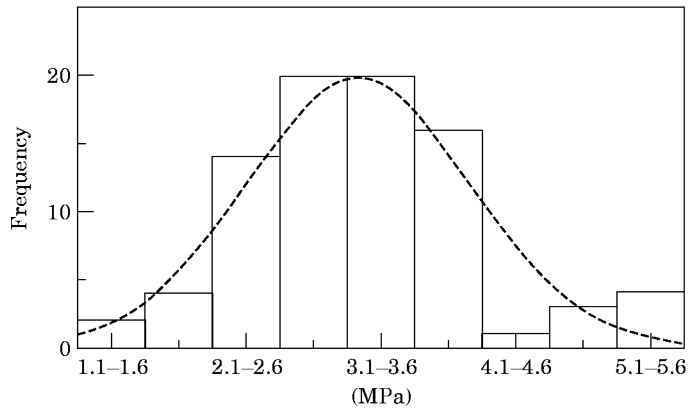

Figure 2. Frequency distribution of weighted and age-normalized (50 years) ultimate static strengths of human vertebrae and vertebral segments, derived from the data on ultimate static strengths of vertebrae (Hansson *et al*. [4]) and vertebral segments (Brinckmann *et al*. [5]). For details see text.

younger age group, but above the age of 60 the differences are very slight. The causes of these differences cannot be positively identified.

For an orientating assessment of compressive strength as a function of age, a mean regression for men was used. If one determines this from the data published by both groups of authors, then an unwanted impairment would inevitably occur due to the differences in the mean values for the two groups. The combination of a higher mean value and lower age in the data of reference [5] in contrast to the combination of a lower mean value and higher age in the data of reference [4], would lead to the influence of age being overestimated. For this reason the age function was determined after weighting the single values of references [4] and [5], multiplying them by the quotients of the overall mean across references [4] and [5] divided by the group means of reference [4] or [5], respectively. Using this approach yields the regession equation, with 95% confidence intervals given in brackets (italics), for males as

$$x[\mathrm{MPa}] = -0{\cdot}037747 \text{ (from } -0{\cdot}049784 \text{ to } -0{\cdot}025710) \times \text{age (years)} \\ + 5{\cdot}106713 \text{ (from } 4{\cdot}449031 \text{ to } 5{\cdot}764395), \quad (11)$$

with a coefficient of determination of 0·32.

The mean ultimate static strengths estimated in this way are, 4·35 MPa, 3·97 MPa, 3·60 MPa, 3·22 MPa and 2·84 MPa for the ages of 20, 30, 40, 50 and 60 years respectively.

The very small number of preparations for persons with ages under 25 years and the increasing calculation uncertainty with decreasing age resulting from this, should be noted. To obtain a reference point for the normal scatter of ultimate static strength values, the weighted data for male preparations (mean age 47·1 years [4, 5]) were adjusted to an age of 50 years by using separate regression equations (estimate of the ultimate static strength as a function of age) for each author group. After combining the weighted and age (50 years) adjusted ultimate static strengths of both investigations (see Figure 2), the data can be characterised as follows: mean value = 3·21 MPa, standard deviation = 0·84 MPa, 5th percentile = 1·89 MPa, 10th percentile = 2·29 MPa, 50th percentile = 3·15 MPa, 90th percentile = 4·24 MPa, 95th percentile = 5·06 MPa. These distributions can be considered as guide values for the determination of the range of ultimate static strengths which are relevant in practice. Ultimate strength data were also presented in references [22, 23]. The compressive loads were applied to differently flexed vertebral segments with a high load rate of 3 kN/s.

Upon considering the reduced compression due to the flexion, the data for men aged between 18 and 65 were combined with the raw data of references [4, 5] of the same age (mean age of all references [4, 5, 22, 23] preparations 42·8 years). They lead to the regression equation, with 95% confidence intervals given in brackets (italics):

$$x[\mathrm{MPa}] = -0{\cdot}067184 \text{ (from } -0{\cdot}082407 \text{ to } -0{\cdot}051961) \times \text{age (years)} + 6{\cdot}765024 \text{ (from } 6{\cdot}077286 \text{ to } 7{\cdot}452763), \quad (12)$$

with a coefficient of determination of 0·46.

The dependence on age is more pronounced and the mean ultimate static strengths estimated for 20 to 40 year old men are somewhat higher than those predicted by equation (11). 5·42 MPa, 4·75 MPa, 4·08 MPa, 3·41 MPa and 2·73 MPa are predicted for the ages of 20, 30, 40, 50 and 60 years, respectively. These combined references [4, 5, 22, 23] and age (40 years) adjusted ultimate static strengths were characterized by a mean value of 4·08 MPa, standard deviation 1·06 MPa, 5th percentile = 2·45 MPa, 10th percentile = 2·74 MPa, 50th percentile = 4·00 MPa, 90th percentile = 5·39 MPa, 95th percentile = 6·24 MPa. This distribution could be alternatively used as guidance for the determination of the range of ultimate static strengths.

If 95 percent of the male population are to be considered, a reduction of the predicted strength by 1·3 (equation (11)) or 1·5 MPa (equation (12)) seems to be appropriate.

3.1.2. *Endurance limit*

Previous findings on the fatigue strength of vertebral segments in cyclic compression [21, 24, 25] were placed in the low cycle fatigue area with regard to the external load and the number of load cycles. In this area, the fatigue failure for bones should depend much more on differences in the material and structure than in the high cycle fatigue area [21]. A different fatigue behaviour for both areas was described by Michel *et al*. [7] and Guo *et al*. [26]. Unlike the WBV situation, vertebrae were examined *in vitro* at only very low frequencies (0·25 to 0·67 Hz). According to Lafferty [27] however, the influence of frequency should be small at frequencies of less than 30 Hz. Morrow [28] stated that the load cycle frequency is of secondary importance for the determination of the fatigue behaviour.

Brinckmann *et al*. [21] pointing to the increased strength under short-term transient dynamic loads relative to static loads [29–31] called for separate fatigue strength investigations for WBV.

The analysis of published data is made difficult by the different load conditions in the relevant papers which were, to some extent, not considered by the authors when interpreting their results. Liu *et al*. [25] examined 11 segments from male corpses. They indicated a minimal load of 0·022 kN for fluctuating sinusoidal loads (0·5 Hz) with relative peak stresses of between 37 and 80% of an inappropriately estimated strength [3]. Therefore, the descriptions of failures as a function of the number of load cycles and assumed ultimate static strength by Liu *et al*. [25] are quantitatively not reliable. They merely suggest the possibility of fatigue failures under repeated dynamic loads below the ultimate static strength. Liu *et al*. [25] suggested a damage mechanism related to microfractures of the inorganic components of the subchondral bone. The data of Hansson *et al*. [24] and Brinckmann *et al*. [21] were taken from the published tables, checked for possible errors and analyzed. They contain, with certain exceptions, comprehensive statements with regard to both the characteristics of the preparations and the stress parameter so that a wide ranging degree of comparability is obtained. The stresses in the two studies differ primarily with regard to the relative (with reference to an ultimate static strength) minimum cyclic stresses from which a cyclic stress range would start. Consequently the average normalized static stresses, defined as the sum of the relative minimum stress plus $0{\cdot}5 \times$ the relative cyclic stress range, also differed.

The cyclic loading used by Hansson *et al*. [24] consisted of a maximum 1000 sinusoidal stress cycles at a frequency of 0·5 Hz. Their data had to be corrected for several cases in order to eliminate obvious discrepancies [3]. Brinckmann *et al*. [21] chose a similar approach. They did not, however, estimate the ultimate static strength of the vertebral segments they examined from the mineral content, like Hansson *et al*. [24]. They [21] predicted it from the experimentally ultimate determined static strength of adjacent segments. The minimum cyclic load of the initially triangular (saw tooth) cyclic loads were 700, 750, 800 or 1000 N. Loads were applied for a maximum of 5000 cycles (1 cycle in 4 s). In addition to the lower frequency a further possible important difference between this experiment and that of reference [24] might have been the temperature at which the preparations were tested ([24]—no statement, presumably room temperature, [21]—at 36·5 degrees Celsius). The normalized static stress was, for a similar cyclic stress, fractionally higher in the experiments of reference [21].

The low number of cycles in experiments with vertebrae or vertebral segments made it necessary to have regard to test results on the fatigue strength of bones with a high number of load cycles and to check their usability. Reviews are found in, amongst others, references [2, 6, 7, 21]. Lafferty *et al*. [6] examined the dorsal elements of vertebrae (three vertebra from one individual and 14 vertebra from Rhesus apes) for fatigue strength in bending with up to about 200 000 cycles. Michel *et al*. [7] tested the fatigue strength of cancellous bone under uniaxial compression load at 2 Hz with up to 350 000 cycles. When the results of both groups of authors were analyzed as described in section 2.1, they exhibited good agreement (see Figure 3). Near 10^6 cycles, the equivalent cyclic stress of about 1 was roughly equal to 20 percent of the ultimate strength limit, i.e., the equivalent cyclic stress of about 5 near 1 cycle. Without consideration of the Goodman relation [8], apparent differences in the high cycle area might be assumed [3].

The reduction of the different data to linear regression equations of the form $\log(N) = b0 \times \log(y_{eqk}) + b1$, with N = number of cycles to failure, $b0$ = slope of straight

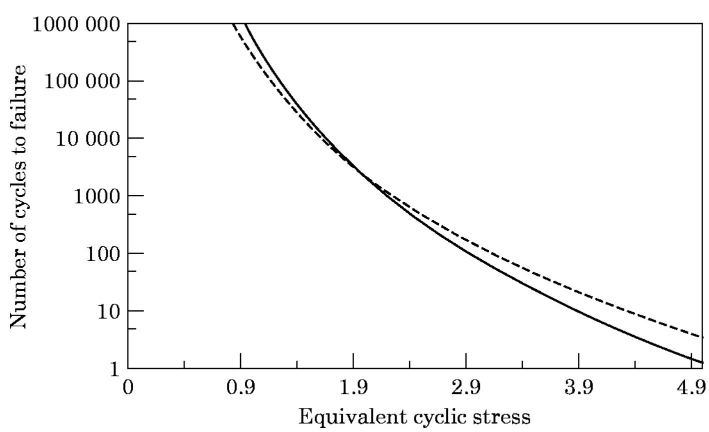

Figure 3. Comparison of estimates of the number of cycles to failure expressed as a function of the equivalent stress, calculated using different non-linear regressions (see text). Data from Michel *et al*. [7] (full line) and Lafferty *et al*. [6] (broken line), recalculated according to the Goodman relation and adjusted to an equivalent cyclic stress ≈ 1 near the endurance limit. For details see text.

line, y_{eqk} = equivalent cyclic stress assuming an endurance limit at 10^6 cycles, and $b1$ = constant (corresponds to that log (N) for a $y_{eqk} = 1$), led to the following results (*Rsq*: coefficient of determination):

$$\log(N) = -7{\cdot}102525 \log(y_{eqk}) + 5{\cdot}518835, \qquad Rsq\text{: } 0{\cdot}86 \text{ for data of reference [6]}, \tag{13}$$

$$\log(N) = -8{\cdot}140544 \log(y_{eqk}) + 5{\cdot}804797, \qquad Rsq\text{: } 0{\cdot}99 \text{ for data of reference [7]}, \tag{14}$$

$$\log(N) = -3{\cdot}396130 \log(y_{eqk}) + 3{\cdot}852875, \quad Rsq\text{: } 0{\cdot}17 \text{ for data of references [21, 24]}, \tag{15}$$

The very low coefficient of determination for the data of vertebrae from references [21, 24] underlines the justification for concerns expressed by Brinckmann *et al*. [21] against such data reduction. An application of equation (15) cannot be recommended. Thus for a first, comparative, assessment of the possible risk due to WBV the use of relationships which derive from the data of Michel *et al*. ([7] equation (14)) appears to be indicated.

Limited information is available on the variability of fatigue strength. On the basis of data from references [21, 24] and from the summary presentation in reference [27], it could be broadly assumed that identical effects will result for relative (with reference to an ultimate static strength) cyclic loads which differ by a relation of up to 1 to 2. From this derives the factor of 0·67 by which y, the mean endurance limit, could be multiplied to reflect the lower limit of this variability as a safety margin. (The mean of 1 and 2 equals 1·5, to be multiplied by 0·67 in order to get 1, the lower limit.)

3.2. EXPERIMENTAL DATA

Evaluation of the relationships between the 135 negative upward peak seat acceleration values (a_u and $a_{w,u}$) and the pertinent mean (*MV*) or mean minus one standard deviation (*MV–SD*) peak compression values for the L3/4 (C3min), L4/5 (C4min) and L5/S1 (C5min) intervertebral discs, obtained from the time series averaged over all 36 subjects, yielded the linear regression equations and related coefficients of determinations given in Table 3. At identical peak accelerations, considerably higher forces were predicted for the bent forward posture.

The ranges between directly successive least and largest pressures were identified in the averaged time histories, by first determining the maximum and then determining the corresponding minimum in an appropriate time window (compare Figure 1). Figure 4 shows the correlation between these calculated mean internal compressive stress ranges and the frequency weighted peak values of seat acceleration upward. In addition to a distinct effect of the posture, the influence of the somatotype of the subjects is obvious.

Table 4 contains all the information necessary to estimate the cyclic stress in the L5/S1 intervertebral disc in MPa from $a_{w,u}$ by using a posture dependent and where required, somatotype dependent, regression equation of the type

$$\text{cyclic stress (peak to peak)} = m a_{w,u} + c. \tag{16}$$

In order to consider the variability of the human biodynamic response, the equations are given for the average time histories of calculated pressure and the average time histories plus one standard deviation of 12 or 36 time series for each somatotype or all subjects, respectively.

TABLE 3

Prediction of the peak compression forces C in N *for the* L3/4 (C3), L4/5 (C4) *and* L5/S1 (C5) *intervertebral discs*, *based on mean values of time series of C averaged over all* 36 *subjects* (*MV*) *or based on mean values of averaged time series minus one standard deviation* (*MV-SD*); D = *driving posture*, BF = *bent forward posture*; m = *slope*; c = *constant of the linear regression equations for the calculation as function of the peak upward accelerations of the seat in* ms^{-2}; a_u *unweighted*; $a_{w,u}$ *weighted according to ISO* 2631 [10]; *coeff*. = *coefficient of determination*

Posture	*C*	*m*	*c*	*a*	Coeff.	Posture	*C*	*m*	*c*	*a*	Coeff.
D	*MV*C3	56·76	−436·52	a_u	0·84	BF	*MV*C3	102·52	−552·43	a_u	0·90
D	*MV*C3	46·13	−471·76	$a_{w,u}$	0·74	BF	*MV*C3	87·26	−604·82	$a_{w,u}$	0·87
D	*MV*C4	57·66	−442·02	a_u	0·84	BF	*MV*C4	111·29	−607·55	a_u	0·90
D	*MV*C4	46·82	−477·95	$a_{w,u}$	0·74	BF	*MV*C4	94·75	−664·36	$a_{w,u}$	0·88
D	*MV*C5	63·55	−491·82	a_u	0·83	BF	*MV*C5	124·91	−689·18	a_u	0·90
D	*MV*C5	51·50	−531·65	$a_{w,u}$	0·74	BF	*MV*C5	106·42	−752·71	$a_{w,u}$	0·88
D	*MV-SD*C3	70·46	−542·14	a_u	0·83	BF	*MV-SD*C3	128·89	−699·02	a_u	0·89
D	*MV-SD*C3	57·80	−584·61	$a_{w,u}$	0·75	BF	*MV-SD*C3	110·39	−763·13	$a_{w,u}$	0·88
D	*MV-SD*C4	71·26	−537·60	a_u	0·84	BF	*MV-SD*C4	138·70	−756·97	a_u	0·90
D	*MV-SD*C4	58·44	−580·54	$a_{w,u}$	0·76	BF	*MV-SD*C4	118·93	−825·68	$a_{w,u}$	0·89
D	*MV-SD*C5	76·84	−599·28	a_u	0·83	BF	*MV-SD*C5	155·58	−855·32	a_u	0·83
D	*MV-SD*C5	62·70	−646·39	$a_{w,u}$	0·74	BF	*MV-SD*C5	133·52	−932·11	$a_{w,u}$	0·89

4. DISCUSSION

The peak compressive forces created within the spine by WBV, have been considered up to now as an important cause of degenerative changes [2, 32]. The occurrence of fatigue failures after repeated, long duration, loading below the ultimate strength has been known in medicine for a long time. Sandover [2] was the first to develop the hypothesis that fatigue failures in the region of the vertebral endplates are the dominant form of primary damage which initiates WBV related diseases of the spine. *In vitro* studies confirmed that fatigue failures occurred in the area of the vertebral endplates under repeated compression significantly below the ultimate strength for a single loading [21, 24, 33]. In contrast, primary changes in the intervertebral disc are not to be expected under these conditions [34–36]. However, other patho-physiological mechanisms than fatigue failure could contribute to the development of spinal disorders after long-term occupational exposure to WBV. These mechanisms seem to be less significant and cannot be quantified at present. Comprehensive discussions on that topic can be found in references [3, 32, 37–39]. Several other factors which cannot be quantified at present are assumed to modify the health risk caused by WBV and high acceleration events: the order in which the external loads are applied, low temperature, periods of rest, state of adaptation to compressive loads, relation between the z-basicentric [1] axis and gravity, non-lateral–symmetric postures.

As the analysis of the literature has shown, experimental *in vitro* data are available as a basis for the prediction of fatigue failure by using the Palmgren–Miner approach and considering the static preload, e.g., by the Goodman relation [8], for the evaluation of internal cyclic repetitive loads. Internal dynamic loads and the internal static preload can be predicted using experimental data, anthropometry, and modeling. The relation between the predicted pressure changes and the predicted tolerance of the exposed person can be used to calculate the individual health risk. As strength data and experimental data demonstrate, the individual health risk depends on exposure and person related factors. Essential exposure related factors are the number and peak intensity of upward accelerations as well as the posture. Essential person related factors are anthropometric data (mass of the body above the endplate, size of endplates—"somatotype") and age. The data presented in this paper could be used to derive a procedure for the prediction of fatigue failure caused by WBV and repetitive shocks that should include the following eight steps.

1. Prediction of the internal static compressive stress x_{int} (in MPa) acting on the endplate. The calculation can be performed using biomechanical models and anthropometric data [3, 12, 40].

2. Calculation of the normalized internal static stress x_n (see equation (1)). The normalization considers the decreasing static strength of the endplate with an increasing age of the person.

3. Estimation of the internal cyclic peak-to-peak stress y_{int} (in MPa) from $a_{w,u}$ by using one of the regression equations presented in section 3.2, equation (16) and Table 4.

4. Calculation of the normalized cyclic stress y_n (see equation (3)), $y_n = y_{int}/(x = 0{\cdot}2)$, upon assuming an endurance limit equal to about 20% of the ultimate strength y. The normalization considers the decreasing static strength of the endplate with an increasing

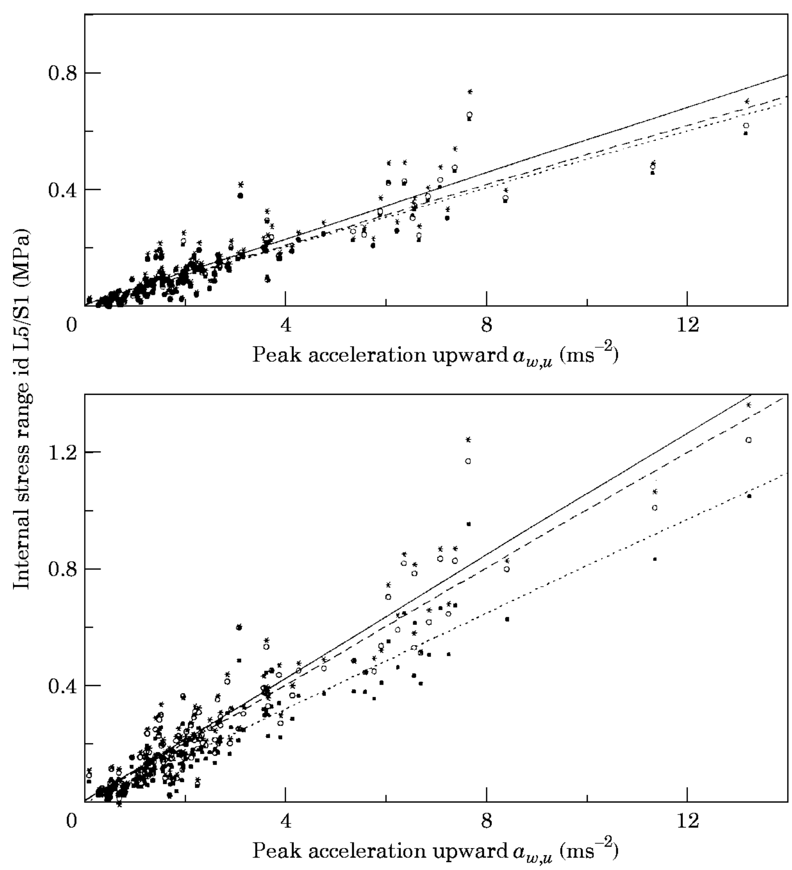

Figure 4. Scatter plots of the ranges between maxima and subsequent minima of the time histories of mean ($N = 36$) pressure calculated for the intervertebral disc (ID) L5/S1 (internal stress range in MPa) of the three groups of frail (*), intermediate (○) and robust (■) subjects, and the frequency weighted (ISO 2631, 1985) upward peak acceleration $a_{w,u}$ (absolute values in ms^{-2}) measured at the seat, with least squares linear regression lines for frail (——), intermediate (- - - -) and robust (· · · · ·) subjects. The values relate to an exposure in the driving (top) and bent forward (bottom) postures. For further details see text.

TABLE 4

Prediction of the absolute cyclic pressure [MPa] *in the* L5/S1 *intervertebral disc by means of linear regressions, based on the extreme values of the mean pressure curves* (*MV*) *or of the mean curves minus one standard deviation* (*MV-SD*) *for* 12 *frail*, 12 *robust*, 12 *intermediate* (*interm*), *and all* 36 *subjects*; D = *driving posture*, BF = *bent forward posture*; *m* = *slopes*, *c* = *constants of the linear regression equations for the calculation as a function of the negative peak value of the frequency weighted* (*ISO* 2631 [10]) *upward seat acceleration* $a_{w,u}$ [ms^{-2}], *Coeff*. = *coefficient of determination*

Subject group	Posture	Pressure maximum	m	c	Coeff.
frail	D	*MV*	−0·054818	0·014931	0·90
robust	D	*MV*	−0·048092	0·014776	0·90
interm	D	*MV*	−0·049620	0·013864	0·90
all	D	*MV*	−0·050843	0·014524	0·90
frail	BF	*MV*	−0·101431	0·026853	0·93
robust	BF	*MV*	−0·078044	0·014118	0·94
interm	BF	*MV*	−0·096245	0·024560	0·93
all	BF	*MV*	−0·091907	0·021844	0·94
frail	D	*MV-SD*	−0·054994	0·151009	0·87
robust	D	*MV-SD*	−0·049553	0·113041	0·90
interm	D	*MV-SD*	−0·052457	0·133375	0·90
all	D	*MV-SD*	−0·052335	0·132475	0·89
frail	BF	*MV-SD*	−0·108237	0·258229	0·94
robust	BF	*MV-SD*	−0·088139	0·177870	0·94
interm	BF	*MV-SD*	−0·100637	0·199876	0·94
all	BF	*MV-SD*	−0·099004	0·211992	0·94

age of the person in order to make the different effects of an identical y at different ages comparable.

5. Calculation of the equivalent cyclic stress $y_{eqk} = x_n + y_n$ (see equation (4)).
6. Calculation of the number of cycles N_k at which failure is likely to occur for an equivalent stress y_{eqk} by using equation (14).
7. Calculation of the quotient $1/N_k$ for $y_{eqk} \geqslant 1$; $Y_{eqk} < 1$ may be ignored, because the internal loads are below the endurance limit (cf. section 3.1.2).
8. Summation of the quotients $1/N_k$ to $\Sigma 1/N_k$. A fatigue failure would be unlikely for $\Sigma 1/N_k < 1$.

In this procedure (steps 1 to 8), $n_k = 1$ is assumed. If $a_{w,u}$ are available as a histogram, the frequencies n_k of the classes k could be used in steps 5 to 8, and the quotient $\Sigma n_k/N_k$ could be calculated [28] instead of $\Sigma 1/N_k$.

To transpose the results into a criterion suitable for the risk prevention purposes, it seems appropriate to consider the normal lower ranges of variation of relevant variables, since the exclusive reference to mean values would, theoretically, result in an assessment which was only correct in approximately 50% of cases. There are four kinds of variability. (1) The variability of the ultimate strength x could be regarded by reducing the mean value predicted by about 1·3 or 1·5 MPa (cf., section 3.1.1.). That would cover approximately 95% of the male population of the same age. (2) The variability of the mean endurance limit $y = 0{\cdot}2x$ could be taken into account by multiplying it by 0·67 (cf., section 3.1.2.). (3) The static preload varies with the body mass, posture and surface area of the endplate. Table 5 illustrates this variability for the subjects who participated in our experiments. A significantly larger variability could be expected in the normal male population with an age up to 60 years. (4) The regression equations (see equation (16)) that are based on the

TABLE 5

Mean values (*MV*), *standard deviations* (*SD*) *and* 75*th* (75*th P*) *and* 95*th percentile* (95*th P*) *of calculated static pressures in* MPa (12 *measurements of each subject for* 10 s) *acting on the lumbar disc* L5/S1 of 3 *groups of subjects* ($N = 12$ *for each group*) *with a frail*, *intermediate and robust somatotype during relaxed driving* (D) *and bent forward* (BF) *postures*

Posture	Somatotype	*MV*	*SD*	75th *P*	95th *P*
D	frail	0·287	0·060	0·301	0·448
D	intermediate	0·263	0·061	0·304	0·366
D	robust	0·264	0·048	0·290	0·361
BF	frail	0·429	0·104	0·496	0·610
BF	intermediate	0·409	0·089	0·461	0·573
BF	robust	0·341	0·077	0·382	0·522

average time series with the standard deviation added to the maxima and subtracted from the minima (see Table 4, rows with "*MV–SD*" in the column "Pressure") give a clue to the variability of the internal cyclic load.

Figure 5 illustrates an example for the exposure times until fatigue failure is predicted as a function of the magnitude of an upward, frequency weighted, peak acceleration $a_{w,u}$ of WBV with 4 Hz acting on a 40 year old male of the "intermediate" somatotype. For these predictions, all four kinds of biological variability were considered additively: i.e., these predictions are extremely conservative and they should not be misinterpreted as exposure limits. The shape of predictions, however, could be compared with the shape of the curves resulting from the two relationships indicated in reference [1] for the time dependence of the health guidance caution zone. This comparison suggests that the time dependence resulting from energy equivalent vibration magnitudes according to Formula

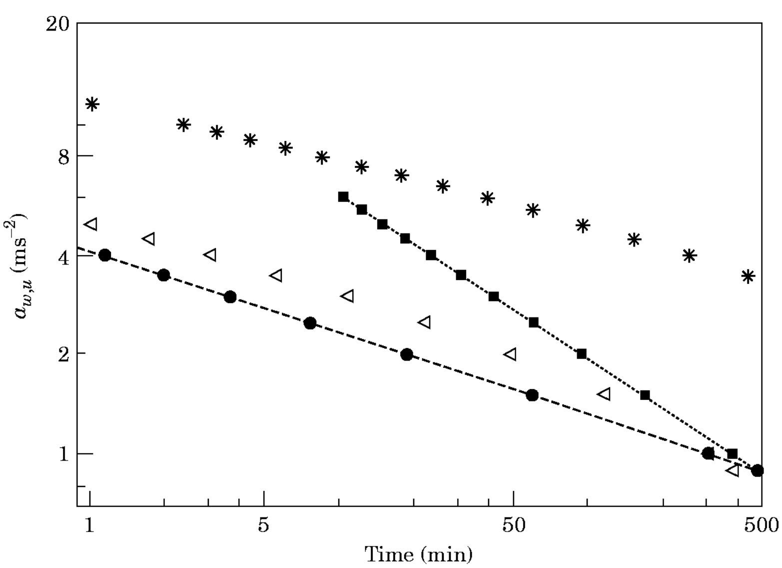

Figure 5. Equivalent daily exposures according to Formula B.1 (■···) and Formula B.2 (●---) of ISO 2631-1 [1] and exposure times to fatigue failure predicted very conservatively as function of the magnitude of the frequency weighted upward peak acceleration $a_{w,u}$ of WBV with 4 Hz acting on a 40 year old male of the intermediate somatotype in the driving (*) and bent forward (◁) postures. For details see text.

B.1 of reference [1] may underestimate the health risk caused by high magnitudes. Formula B.2 exhibits a similar shape as that based on predicted fatigue failure. Therefore Formula B.2 [1] seems to represent the human response better and can be recommended for the evaluation of such conditions.

ACKNOWLEDGMENT

The authors gratefully acknowledge the collaboration and assistance of L. Gericke, J. Keitel, H.-J. Lindner, G. Menzel, R. Vizcaino and W. Wagner.

REFERENCES

1. International Organization For Standardization 1997 *ISO* 2631-1, Mechanical vibration and shock—evaluation of human exposure to whole-body vibration. Part 1: general requirements. Second edition.
2. J. Sandover 1981 *University of Technology*, *Department of Human Sciences*, *Loughborough*. *Research Report No DHS* 402. Vibration, posture and low-back disorders of professional drivers.
3. H. Seidel, B. Blüthner, B. Hinz, M. Schust 1997 *Stresses in the lumbar spine due to whole-body vibration containing shocks* (*Final Report*). Bremerhaven: Wirtschaftsverlag NW.
4. T. Hansson, B. Roos, A. Nachemson 1980 *Spine* **5**, 46–55. The bone mineral content and ultimate compressive strength of lumbar vertebrae.
5. P. Brinckmann, M. Biggemann, D. Hilweg 1989 *Clinical Biomechanics* **4**, Suppl. 2, S1–S27. Prediction of the compressive strength of human lumbar vertebrae.
6. J. F. Lafferty, W. G. Winter, S. A. Gambaro 1977 *The Journal of Bone and Joint Surgery*, **59**-A, 154–158. Fatigue characteristics of posterior elements of vertebrae.
7. M. C. Michel, X.-D. E. Guo, L. J. Gibson, T. A. McMahon, W. C. Hayes 1993 *Journal of Biomechanics* **26**, 453–463. Compressive fatigue behaviour of bovine trabecular bone.
8. E. G. Fischer 1976 In *Shock and Vibration Handbook* (Editors H. M. Harris and C. E. Crede) New York: McGraw-Hill Book company; Theory of equipment design 42.1–42.36.
9. J. Kinne and R. Melzig-Thiel 1993 *ErgoMed* **17**, 154–157. The optimisation and testing of suspension seats using mechanical models of humans—average frequency characteristics and input impedance of seated persons (Part 1, German). (Optimierung und Prüfung von Schwingungsschutzsitzen mit mechanischen Mensch-Modellen—Mittlere Frequenzgänge und Eingangimpedanz von sitzenden Menschen (Teil 1)).
10. International Organization For Standardisation 1985 *ISO* 2631/1-1985(*E*). International standard evaluation of human exposure to whole-body vibration—Part 1: general requirements. First edition.
11. H. Greil 1988 *Habilitation*, *Humboldt-University Berlin*. The physique of adults—DDR representative cross sectional anthropometrical study 1982/84 (German). (Der Körperbau im Erwachsenenalter—DDR—repräsentetive anthropologische Querschnittsstudie 1982/84).
12. B. Hinz and H. Seidel 1995 *Arbeitsmedizin*, *Sozialmedizin*, *Umweltmedizin* **30**, 487–499. Anthropometrical methods and quantification of spinal strength under internal mechanical loads. Theoretical approach and examples of its application. (German) (Anthropometrische Methoden und Quantifizierung der Wirbelsäulenbelastbarkeit für interne mechanische Belastungen. Theoretischer Ansatz und Anwendungsbeispiel.)
13. D. Colombini, E. Occhipinti, A. Grieco, M. Faccini 1989 *Spine* **14**, 51–55. Estimation of lumbar disc areas by means of anthropometric parameters.
14. W. M. Saziorski, A. S. Aruin, W. N. Selujanow 1984 *Biomechanics of the movement apparatus in humans*. (*German*), (*Biomechanik des menschlichen Bewegungsapparates*). Berlin: Sportverlag.
15. Y. K. Liu, J. M. Laborde, W. C. Van Buskirk 1971 *Aerospace Medicine* **42**, 650–657. Inertial properties of a segmented cadaver trunk: their implications in acceleration injuries.
16. S. Plagenhoef, E. G. Evans, T. Abdelnour 1983 *Research Quarterly for Exercise and Sport* **54**, 169–178. Anatomical data for analysing human motion.
17. E. Nemeth and H. Ohlsen 1986 *Spine* **11**, 158–160. Moment arm length of trunk muscles to the lumbosacral joint obtained *in vivo* with computed tomography.

18. P. J. Moga, M. Erig, D. B. Chaffin and M. A. Nussbaum 1993 *Spine* **18,** 2305–2309. Torso muscle moment arms at intervetebral levels T10 through L5 from CT scans on eleven male and eight female subjects.
19. P. Tveit, A. G. Cresswell, K. Daggfeld, A. Thorstensson 1991 XIII*th*. *Int*. *Congress on Biomechanics*, 9–13 *December*, *Perth Abstracts*, 249. Spinal curvature changes lever arm length for erector spinae.
20. M. Jäger, A. Luttmann, W. Laurig 1991 *International Journal Industrial Ergonomics B*, 261–277. Lumbar load during one-handed brick-laying.
21. P. Brinckmann, M. Biggemann, D. Hilweg 1988 *Clinical Biomechanics* **3**, Suppl. 1, S1–S23. Fatigue fracture of human lumbar vertebrae.
22. W. D. Hutton and M. A. Adams 1982 *Spine* **7**, 586–590. Can the lumbar spine be crushed in heavy lifting?
23. R. W. Porter, M. A. Adams and W. C. Hutton 1989 *Spine* **14**, 201–203. Physical activity and the strength of the lumbar spine
24. T. H. Hansson, T. S. Keller, D. M. Spengler 1987 *Journal of Orthopaedic Research* **5**, 479–487. Mechanical behaviour of the human lumbar spine. II. Fatigue strength during dynamic compressive loading.
25. Y. K. Liu, G. Njus, J. Buckwaller, K. Wakano 1983 *Spine* **8**, 857–865. Fatigue response of lumbar intervertebral joints under axial cyclic loading.
26. X.-D. E. Guo, Th. McMahon, T. M. Keaveny, W. C. Hayes, L. F. Gibson 1994 *Journal of Biomechanics* **27**, 145–155. Finite element modelling of damage accumulation in trabecular bone under cyclic loading.
27. J. F. Lafferty 1978 *Aviation Space Environmental Medicine* **49**, 170–174. Analytical model of the fatigue characteristics of bone.
28. C. T. Morrow 1976 In *Shock and Vibration Handbook* (Editors H. M. Harris und C. E. Crede.) New York: McGraw-Hill Book Company; Environmental specifications and testing 24.1–24.13.
29. W. C. Hutton, B. M. Cyron, J. R. R. Stott 1979 *Journal Anatomy* **129**, 753–758. The compressive strength of lumbar vertebrae.
30. L. Kazarian and G. A. Graves 1977 *Spine* **2**, 1–14. Compressive strength characteristics of the human vertebral centrum.
31. O. Perey 1957 *Acta Orthopedica Scandinavica*, Suppl. No. XXV, 1–101. Fracture of the vertebral end-plate in the lumbar spine. An experimental biomechanical investigation.
32. H. Seidel 1993 *American Journal of Industrial Medicine* **23**, 589–604. Selected health risks caused by long term whole-body vibration.
33. M. E. Amman, F. W. Winkelbauer, K. Hittmair, S. Thurnher, 1993 *The British Journal of Radiology* **66**, 381–383. Short communication: Dynamic forces influencing vertebral fracture risk.
34. M. A. Adams and P. Dolan 1995 *Clinical Biomechanics* **10**, 3–19. Recent advances in lumbar spinal mechanics and their clinical significance.
35. M. Kasra, A. Shirazi-Adl, G. Drouin 1992 *Spine* **17**, 93–102. Dynamics of human lumbar intervertebral joints. Experimental and finite-element investigations.
36. T. S. Keller, D. M. Spengler, T. H. Hansson 1987 *Journal Orthopaedic Research* **5**, 467–478. Mechanical behaviour of the human lumbar spine. I. Creep analysis during static compressive loading.
37. H. Seidel and R. Heide 1986 *International Archives Occupational Environmental Health* **58**, 1–26. Long-term effects of whole-body vibration—a critical survey of the literature.
38. P. M. Bongers and H. C. Boshuizen 1990 *Doctoral Thesis*, *University of Amsterdam*. Back disorders and whole-body vibration at work.
39. H. Seidel 1995 *Back-muscle fatigue and whole-body vibration*. In: European Commission. Work and Health. Scientific basis of progress in the working environment. International Conference 22 to 25 February 1993, Copenhagen, Denmark. Invited lectures. Eds.: O. Svane and Ch. Johansen. Report EUR 15980 EN. Luxembourg: Office for Official Publications of the European Communities. 89–94.
40. D. B. Chaffin and G. B. J. Andersson 1991 *Occupational Biomechanics*. New York, Chichester, Brisbane, Toronto, Singapore: John Wiley; second edition.

APPENDIX: SYMBOLS AND ABBREVIATIONS

a	unweighted acceleration
a_d	extreme unweighted acceleration downward

a_u extreme unweighted acceleration upward
a_{rms} root mean square of the unweighted acceleration
a_w frequency weighted [10] acceleration
$a_{w,d}$ extreme frequency weighted [10] acceleration downward
$a_{w,u}$ extreme frequency weighted [10] acceleration upward
$a_{w,rms}$ root mean square of the frequency weighted [10] acceleration
F fatigue damage
BF bent forward sitting posture
D relaxed driving posture
FZ resultant force at the interface between the subject and the seat
HI humeral index
k amplitude of the equivalent cyclic stress
L1 first lumbar vertebrae
L3 third lumbar vertebrae
L4 fourth lumbar vertebrae
L5 fifth lumbar vertebrae
MV mean value
n_k number of internal cyclic peak-to-peak loads with k
N_k number of cycles to failure at k
Rsq coefficient of determination
S1 first spinous process in the Crista sacralis intermedia of the Os sacrum
S3 third spinous process in the Crista sacralis intermedia of the Os sacrum
SD standard deviation
T5 fifth thoracic vertebra
T11 eleventh thoracic vertebra
T12 twelfth thoracic vertebra
x ultimate static strength in MPa
x_{int} internal stress in MPa
x_n normalized internal static stress
y_{int} internal cyclic peak-to-peak stress in MPa
y_n normalized internal cyclic stress
y endurance limit in MPa
y_{eqk} equivalent cyclic stress with the amplitude k