Applied Ergonomics 36 (2005) 481–488

APPLIED ERGONOMICS

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Vibration induced low back disorders—comparison of the vibration evaluation according to ISO 2631 with a force-related evaluation

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Received 28 October 2003; accepted 10 January 2005

Abstract

Long-term vibration stress can contribute to degenerative changes in the joints of the human body, especially in the lumbar spine. An important factor in the development of these diseases is given by the forces transmitted in the joints. Because the forces can hardly be measured a biomechanical model was developed which simulates the human body in the standing and the sitting posture. The vibration properties of the model were adapted to the transfer function provided in the standards and the literature. With the model the compressive forces at the driving point of the body, in the leg joints, and in two motion segments of the spine were simulated under a vertical pseudo random vibration. Transfer functions between the accelerations of the ground or of the seat and the forces were computed. Furthermore, based on the transfer function between seat acceleration and compressive force in the spinal motion segment L3–L4 weighting factors were derived. By means of these factors characteristic vibration values were computed for 57 realistic vibration spectra measured on 17 machines and vehicles. The consideration of the forces resulted in a stronger weighting of low-frequency vibrations compared to the weighted acceleration as suggested by ISO 2631-1. In order to enable an assessment of the health risk a force-related guidance value was derived which amounts to $0.81\,\mathrm{ms}^{-2}$ (rms).

Keywords: Biomechanical model; Weighting factors; Force-related guidance value

1. Introduction

In the annex of the international standard ISO 2631-1 (1997) it is emphasized that biodynamic research as well as epidemiological studies have given evidence for an elevated health impairment due to long-term exposure to high-intensity whole-body vibration. Mainly the lumbar spine and the connected nervous system may be affected. Symptoms of the low-back disorders are not specifically induced by the vibration stress. However, due to vibration the symptoms occur earlier and much more frequently than in the case of non-exposed persons (Dupuis and Zerlett, 1986). Therefore, it is necessary to assess the vibration stress in order to perform preventive measures or during the process of the acknowledgement of the disorders as an occupational disease, as it is possible in Germany for example. According to ISO 2631-1 the vibration should not exceed the weighted acceleration $a_{\mathrm{we}} = 0.8\,\mathrm{ms}^{-2}$ (rms) in the case of long-term exposure and a daily exposure time of 8 h.

For the computation of the weighted acceleration, the time-course of the acceleration has to be measured between the seat cushion and the pelvis (buttocks) and analysed in the frequency domain (ISO 2631-1). Each frequency component is weighted by a factor which is derived from the human vibration sensation. In case of time-variable or interrupted vibration exposure, the different exposure periods of a working day have to be summed up. Then the daily vibration stress is assessed by the root of the sum of all products of the squared frequency-weighted accelerations and their exposure

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0003-6870/$ - see front matter
doi:10.1016/j.apergo.2005.01.008

durations over the daily exposure (Eq. A.3 in ISO 2631-1).

The vibration exposure of the human can induce disorders of the same organ which is also affected by lifting or carrying heavy objects or work with the trunk highly inclined. Concerning these activities it is assumed that the mechanical loading of the lumbar spine and especially the compressive forces transmitted in the spinal segments are important triggers for the pathogenesis of the lumbar spine disorders. Thus it seems to be reasonable to consider the spine forces instead of the vibration sensation for the evaluation of the vibration stress. Furthermore, the force-related evaluation is desirable because lifting and carrying objects are combined with vibration stress in several occupational activities, e.g. in agriculture and farming, in road and underground construction, or in transport industry, etc. (Schäfer and Hartung, 1999).

The forces in the joints of the human body can be measured only with great efforts under clinical conditions. However, they can be assessed by means of biomechanical models which contain rigid and elastic elements to imitate the spine. E.g. Buck (1997) developed a model of the seated human including a detailed presentation of the lumbar spine. For the routine application this model was simplified by Buck et al. (1997) in essential details. In both models the arms and legs are imitated by rigid segments which are fixed to the trunk by hinges. In contrast, Fritz (2000) included in his model force elements for the trunk and also for the leg muscles so that the model is able to imitate the human body in the standing or in the sitting posture by changing only the geometrical input data. In the present study the forces in the leg joints and the spine are simulated by the model and the relationships between these internal forces and the vibration acceleration are described by transfer functions.

Furthermore a method is developed for the force-related evaluation of whole-body vibration and for the computation of characteristic vibration values. The evaluation is carried out for realistic machine vibrations and compared to the evaluation given in the standards.

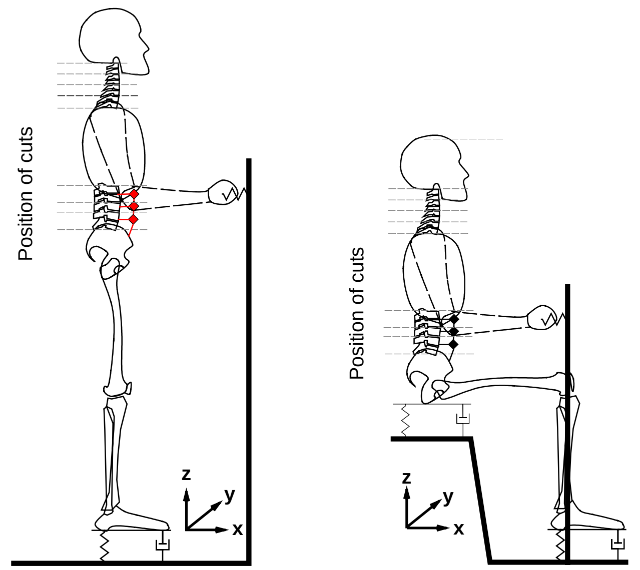

Fig. 1. Schematic of the skeleton of the human trunk, neck, head, and legs and of the arms and the viscera (◆) in the standing and sitting posture. The dashed lines indicate the position of the nine cuts through the trunk and neck (coordinate system according to ISO 2631-1).

2. Methods

The applied biomechanical model was described in detail by Fritz (2000). For a better understanding of the following computations it seems to be useful to describe the important elements of the model.

2.1. Structure of the biomechanical model

In the model head, trunk, abdomen, arms, and legs are imitated by 27 rigid bodies (Fig. 1). An additional body represents the vibrating ground or seat including the vibrating handles which are grasped by the hands. The bodies are connected by ‘ideal’ joints enabling not only rotational but also translational motions of the bodies. Furthermore, in the ideal joints forces and torques are transmitted between the bodies.

As indicated by the dashed lines of the cuts in Fig. 1 three bodies imitate the lumbar spine and the surrounding muscles and four bodies imitate the neck. According to the limited torsion range in the motion segments of the lumbar spine, the mobility in the three inferior joints is restricted to three-dimensional translations and rotations around the two horizontal axes. In the other joints translations and rotations in each of the three directions are possible. The relationship between the forces and torques, respectively, and the motions in the joints is given by stiffness matrices. Panjabi et al. (1976) measured the stiffness matrices for the displacements of the thoracic spine in positive and negative directions. However, the stiffness properties of the thoracic spine differ from those of the other spine segments. Therefore, the matrices were multiplied by correction factors which corresponded to the cross sectional area of different spine segments.

The abdominal viscera are represented by three rigid bodies which can shift against the contiguous bodies of the lumbar spine (Fig. 1). The movements of the bodies are restricted by force elements tensioned to the pelvis, the lumbar spine, the thorax and between the three bodies.

Thigh, shank, foot, and patella are represented by four bodies at which the patella serves as a support of the force elements imitating the quadriceps femoris. The rotational mobility in the joints corresponds to the mobility in the leg joints. Further, thigh, shank, and foot can shift in their longitudinal direction. Hereby the deformations of the articular surfaces are simulated which result from the forces transmitted in the joints.

The function of the trunk and neck muscles is imitated by 58 force elements, namely 28 elements in the abdominal and lumbar region and 30 elements in the neck-shoulder region. The muscles of each leg are represented by 24 force elements. In order to maintain an erect posture leg, trunk, and neck muscles have to contract actively. The vibration induced movements of the human body can be controlled by the muscles only at frequencies lower than 8 Hz (Freund 1983; Seidel et al., 1986). At higher frequencies the muscles are passively stretched. Accordingly, the force elements in the model have to exert forces composed of two parts. The constant force parts enable the posture stabilization of the model and the time dependent parts simulate the vibration-induced variation of the muscle forces. The forces result from the stretching of the parallel and the serial elastic muscle elements and of the contractile elements according to Hill's model of a muscle fibre. The relationship between the muscle stretching and the muscle forces are approximated by an adequate equation (see Fritz, 2000).

In the standing posture the motions of the vibrating ground and the handles are transmitted by springs and dampers to the feet and the hands of the model. In the case of the sitting operator the motions of the seat are transmitted by additional springs and dampers to the pelvis of the model (Fig. 1).

2.2. Transfer functions

In general, the relationship between the input signal and the output signal of a system is described by means of the transfer functions. In this study the relationships between the vibration excitation representing the input signal and several responses of the human body are considered. Although the input and output signals are measured or simulated in the time domain the transfer functions are computed in the frequency domain in order to facilitate the interpretation of the functions. In the standards concerning the vibration stress of the human body the following transfer functions are specified:

- between the accelerations of the seat–body interface and of the head (called seat-to-head transmissibility);
- between the acceleration of the seat-body interface and the forces transmitted from the seat to the buttocks (called apparent mass).

Whereas the first transfer function compares the motions of two points of the system, the second function describes the relationship between the seat motion and the oscillating part of the force at the driving point of the human body. According to Newton's second law, forces are the cause of the acceleration of a body. In contrast to measurements a biomechanical model enables the assessment of the oscillating forces at points within the human body too. Thus corresponding transfer functions can be computed for the internal forces.

For the computation of the different transfer functions the input signal imitated a vertical pseudo random vibration. This excitation enables the presentation of the transfer functions in a broad frequency range by performing only one simulation trial.

2.3. Evaluation of the vibration

As mentioned earlier for the evaluation of the vibration of machinery or vehicles the acceleration of the seat–body interface has to be measured and weighted with frequency dependent factors. At first 57 spectra of vertical seat vibration were weighted according to ISO 2631-1 (1997) and the weighted accelerations a_{we} were computed. The spectra were measured by the "Berufsgenossenschaftliches Institut für Arbeitsschutz—BIA (Sankt Augustin, Germany)" on 17 different vehicles, mobile machinery, helicopters, and motorcycles.

In the next step the model's transfer function between the seat acceleration and the compressive force in the lumbar motion segment L3–L4 was used to derive force-related weighting factors. For this the moduli of the transfer function were divided by the mass which is supported by the seat. According to the mass distribution of the human body this mass was set to 53.3 kg for a whole-body mass of 75.0 kg. With these dimensionless factors the 57 given acceleration spectra were weighted and then the force-related characteristic values of the vibrations were calculated by computing rms values as described in ISO 2631-1. The computation of the characteristic values in the form of rms values has the advantage that the values do not depend upon the phase shifts of the different frequency parts of the vibration which simplifies the calculation of the characteristic values and thus reduces the expense.

Finally, in order to compare the two evaluation methods the ratio of the characteristic vibration values for the 57 recorded acceleration spectra, resulting from the weighting according to ISO 2631-1 (1997) and the force-related weighting, were calculated. Using non-linear regression, an exponential function was fitted to relate these ratios to the median frequency of the spectra. The median frequency divides the integral of the acceleration over the frequency range into two halves and is slightly influenced by the noise of the signal.

3. Results

The vibration properties of the model depend upon the input data realizing the body posture, the anthropometric data, the elasticity of the spine, the elasticity of the connection between the model and the vibrating ground and seat, and upon the muscle forces. The model was adjusted to the transfer functions given in the standards and in the literature by varying the elasticity of the connection between the model and the ground and seat and by the muscles forces (Fritz, 2000).

The time courses of the following compressive forces were computed by means of the model imitating the standing operator: between the ground and the foot, in the ankle, the knee, and the hip as well as in the motion segments L3–L4 and C6–C7. The transfer functions resulting from the vibration induced cyclic parts of these six compressive forces are shown on Fig. 2. The moduli of these functions reach their maximal values between 5 and 6 Hz. After a more or less marked plateau at 8 Hz the moduli decrease to small values at high frequencies. The moduli of the transfer functions of the forces between the ground and the foot, in the ankle, and in the knee reach nearly the same level. They are approximately one third lower than the moduli calculated for the motion segment L3–L4 although each leg is loaded by one half of the body weight only. By means of the phase curves the time delay between the peak values of the vibration acceleration and of the compressive forces can be assessed. The delay increases from the ankle to the neck with increasing frequency. The discontinuity in the curve of the phase shift of the cervical spine from -180° to $+180^\circ$ can also be interpreted as a further backward shift. Observing two cyclic motions it is not possible to distinguish between a backward or a forward phase shift at angles greater than 180°.

In addition to the standing human model a seated operator was imitated by the model. The operator's back was unsupported and the feet were resting on the vibrating platform. For this posture the apparent mass and the transfer functions between the vertical seat acceleration and the compressive forces acting in the motion segments L3–L4 and C6–C7 are shown in Fig. 3. The moduli of these three functions reach their maximum values at frequencies between 4 and 5 Hz and then decrease with increasing frequency. The ratios of the moduli indicate that the highest cyclic forces are acting between the seat and the buttocks whereas the forces acting in the cervical spine are small. From the phase curves it can be inferred that the peak values of the forces are delayed with respect to the peaks of the acceleration. Due to the vibration transmission in the human body the time delay is greater for the lumbar and cervical spine than for the connection between the seat and the pelvis.

Fig. 4 displays the curve of the force-related weighting factor w_{force} as a function of the frequency. As described earlier the factor was derived from the transfer function

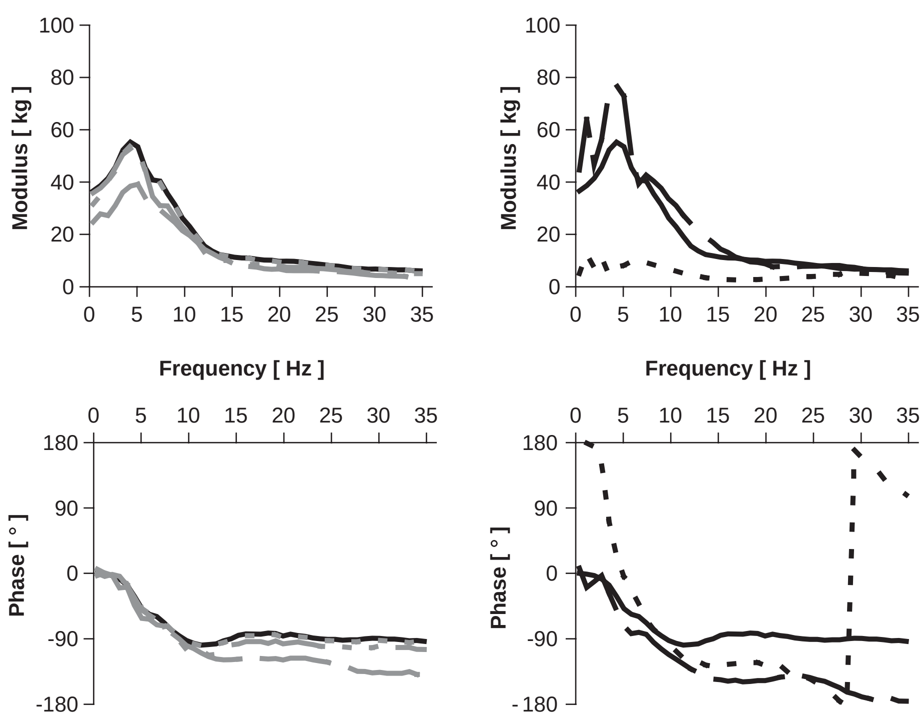

Fig. 2. Transfer function between the vertical ground acceleration and the compressive force —— between ground and foot (apparent mass), left side: — — — in the hip joint, — · — in the knee joint, – – – – in the ankle joint, right side: ——— in the spinal motion segment L3–L4, – – – – in the spinal motion segment C6–C7.

of the motion segment L3–L4 for a seated body under vertical vibration. The maximum value of the factor amounts to 1.19 at 4 Hz which is the resonance frequency of the human body in the sitting posture. At frequencies above 30 Hz the factor amounts to about 0.07. The curve can be approximated by the rational polynomial function with the degree of 5

$$w_{\mathrm{force}}(f) = \frac{\sum_{i=1}^{6} a_i f^{i-1}}{\sum_{i=1}^{6} b_i f^{i-1}}, \tag{1}$$

where the frequency f is within the range

$0.5\,\mathrm{Hz} \leqslant f \leqslant 35\,\mathrm{Hz}$.

The parameters a_i and b_i of this function are listed in Table 1.

For each of the 57 given acceleration spectra the ratio of the force-related characteristic value a_{fe} over the weighted acceleration a_{we} (ISO 2631-1) were computed. In Fig. 5 the ratios are plotted as a function of the median frequencies of the spectra. At frequencies lower than 5 Hz the ratio is greater than 1. With increasing median frequency the ratio decreases and above 20 Hz it

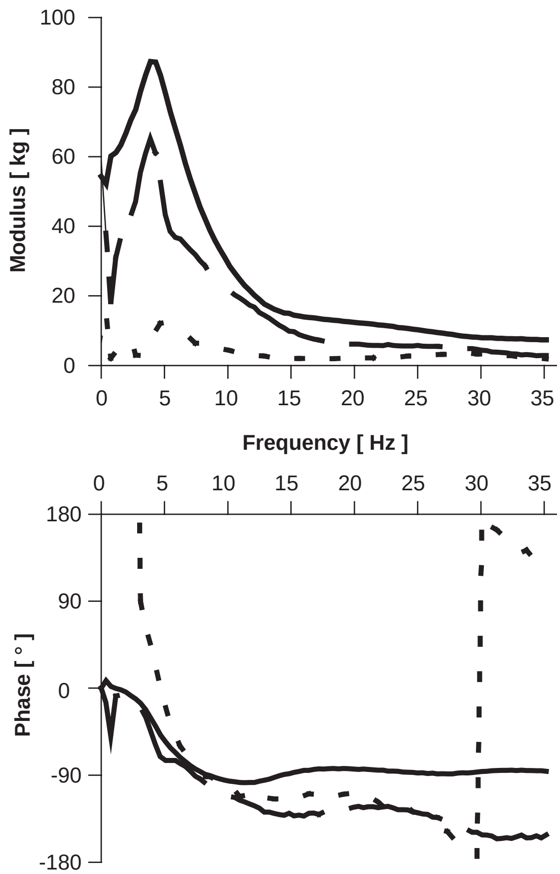

Fig. 3. Transfer function between the vertical seat acceleration and the compressive force —— between seat and pelvis (apparent mass), ——— in the spinal motion segment L3–L4, –––– in the spinal motion segment C6–C7.

Table 1
Parameters a_i, b_i, of Eq. (1) and c_i of Eq. (2)

Index i	Eq. (1) force-related weighting factor		Eq. (2) Ratio $a_{\mathrm{fe}}/a_{\mathrm{we}}$
	a_i	b_i	c_i
1	0.40752038	1.0	0.58805976
2	−0.13893113	−0.60946555	0.59122121
3	0.00817374	0.13998675	7.2348584
4	0.00099088	−0.01398941	—
5	−0.00002371	0.00036447	—
6	0.00000223	0.00003466	—

The functions given by these equations approximate the curves plotted in Figs. 4 and 5.

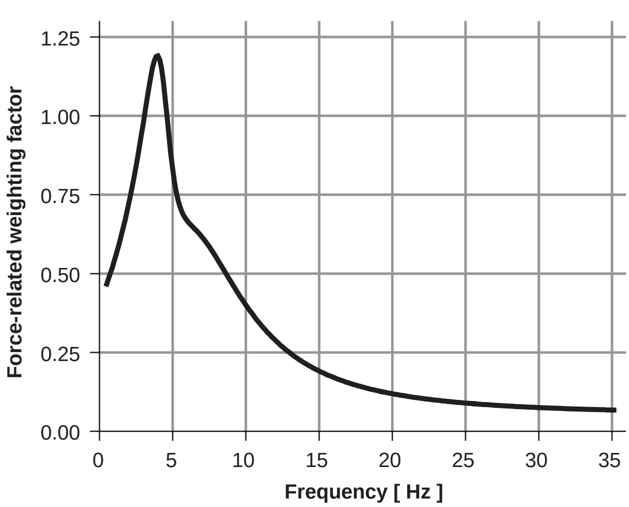

Fig. 4. Force-related weighting factors, computed by means of the cyclic compressive forces in the motion segment L3–L4 under vertical vibration in an erect sitting posture.

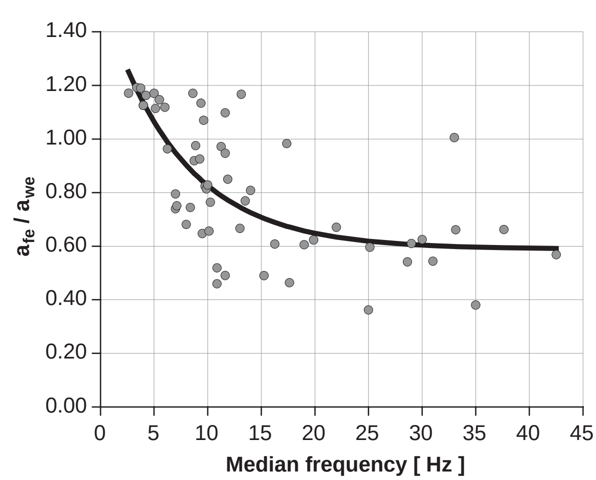

Fig. 5. Ratio of the force-related weighting a_{fe} over the weighted acceleration a_{we} related to the median frequency for 57 acceleration spectra, solid line: approximation of the data by an exponential function ($r = 0.714$).

is nearly constant at an average value of 0.6. The relationship between the 57 ratios and the frequencies can be approximated by the exponential function

$$\frac{a_{\text{fe}}}{a_{\text{we}}}(f_{\text{med}}) = c_1 + c_2 + \exp(-f_{\text{med}}/c_3) \tag{2}$$

with f_{med} denoting the median frequency of the spectrum. The three parameters c_1, c_2, and c_3 of the function are listed in Table 1, fourth column.

4. Discussion

The transfer function between the acceleration of the vibrating ground and the compressive force transmitted from the ground to the feet represents the apparent mass of the standing operator. Matsumoto and Griffin (1998) investigated the influence of the posture of the legs on this apparent mass. The biomechanical model used for the simulations imitated a standing posture with straight legs. A comparison between the simulated apparent mass (Fig. 2) and the function measured with straight legs under equal vibration level performed by Fritz (2000) showed that the simulated moduli were somewhat lower than the measured moduli. The greatest differences up to 50% were found in the frequency range between 10 and 20 Hz. In the human body the compressive forces decrease from joint to joint since a part of the forces is used for the acceleration of the body segments lying distal from the joint. Thus it can be assumed that also the transfer functions of the compressive forces in the leg joints and the spine reflect a small underestimate of the cyclic part of the forces.

In ISO 5982 the vibration response of the seated operator is described by the seat-to-head transmissibility and by the apparent mass. The idealized curves are valid for the posture which is imitated by the model, namely sitting erect without backrest and the feet resting on the vibrating base platform. A satisfactory conformity is obtained between the apparent mass given in the standard and the simulated function in Fig. 3 whereas differences exist between the corresponding seat-to-head transmissibilities. The simulated modulus curve of the transmissibility, which is not shown in the present study, reaches a secondary maximum value at about 30 Hz which is not included in the standard curve. However, the secondary maximum value lies between the lower and upper range of the transmissibility as listed by Paddan and Griffin (1998). Compared to the early model of Fritz (1998) this improvement resulted from the consideration of the rotation of the pelvis around its transverse axis, the pitch motion, as it was already mentioned and required by Broman et al. (1996).

To obtain the force-related weighting factor the transfer function of the compressive force acting in the motion segment L3–L4 was divided by the mass which was supported by the seat (Fig. 4), so that the weighting factors are independent of the operator's mass. Furthermore, it is assumed that this weighting factor has the same property as the normalized modulus of the apparent mass. Measurements by Fairley and Griffin (1989) showed that the inter-individual variability in the normalized moduli was remarkably small.

For the comparison of the two evaluation methods the weighted acceleration a_{we} (ISO 2631-1) and the force-related characteristic value a_{fe} were computed for 57 realistic vibrations spectra. Fig. 5 shows that vibrations with low median frequencies are evaluated stronger by the force-related values than by the weighted acceleration and that this relationship is inverted at median frequencies above 6 Hz. Thus the tendency which is instituted by the last revision of ISO 2631-1, namely the increase of the weighting of high vibration frequencies, is removed by the force-related weighting (Fritz et al., 2003). On the other hand it seems that the revision of the standard does not enable a better assessment of the health risk under vibration. So Hinz et al. (1999) found in their measurements that the correlation between the peak values of the vibration-induced compressive forces and the non-weighted peak acceleration does hardly differ from the correlation between the force and the weighted acceleration. Furthermore, they stated that in contrast to actual research the revised weighting procedure enables higher vibration intensities for the particularly endangered operators (e.g. drivers of wheel-tractors or -loaders) before the health guidance limit is exceeded.

For an improved assessment of the health risk it is not sufficient to recommend force-related weighting factors resulting in new characteristic vibration values. Additionally, a force-related guidance value has to be derived. If the characteristic value computed for a measured vibration exceeds such guidance value a health risk of the operator is likely. Deriving the guidance value the following conditions have to be regarded:

- the compressive forces acting in the motion segments of the spine are composed by a temporal constant (static) and a cyclic force part;
- the cyclic force parts should not exceed the endurance strength of the motion segments.

It can be assumed that the time course of the vibration acceleration represents a random signal. Consequently the time course of the cyclic part of the compressive force can be considered as random too. As for a random signal the rms value equals the standard deviation, the span of the whole signal range, in the special case the difference between the minimal and maximal cyclic part of the compressive force, may be approximated by the eight-fold rms value

$$F_{\max} - F_{\min} = 8F_{\mathrm{rms}}.$$

Assuming a Gaussian distribution the probability of the compressive force exceeding this range is 0.006%.

To assess the ratio between the fatigue strength of the motion segments and the time variant compressive forces Seidel et al. (1997) suggested the following method which was derived from the Goodman relation (Fischer, 1976) and results of Brinckmann et al. (1988). The resultant strength is calculated as the sum of the two quotients of the cyclic force range (difference between minimal and maximal force) over the endurance strength and of the static force part over the ultimate compressive strength. In order to avoid a mechanical overload of the motion segment the resultant strength should be lower than unity. According to Seidel et al., the endurance strength amounts to 20% of the ultimate. The corresponding daily exposure amounts to 8 h without long-lasting breaks.

Regarding the great differences between the individual strength values it seems to be reasonable to compute the resultant strength using the guidance values recommended by Jäger and Luttmann (1996). By these values the individual properties, i.e. the age-, and the gender-induced variations are taken into account. E.g. the guidance value of the compressive strength of the lumbar spine results in 2300 N for male persons older than 60 years. The corresponding endurance strength is 460 N. For females this strength is reached nearly 10 years earlier.

From the relationships mentioned above it follows for the assessment of the compressive forces which do probably not lead to a health risk

$$1 \geqslant \frac{F_{\max} - F_{\min}}{460\,\mathrm{N}} + \frac{F_{\mathrm{static}}}{2300\,\mathrm{N}}$$

and inserting the rms value of the compressive force

$$1 \geqslant \frac{8F_{\mathrm{rms}}}{460\,\mathrm{N}} + \frac{F_{\mathrm{static}}}{2300\,\mathrm{N}}.$$

During sitting without vibration the compressive force in the motion segment L3–L4 is computed to

$$F_{\mathrm{static}} = 575\,\mathrm{N}.$$

Then for the rms value the following relation has to hold

$$F_{\mathrm{rms}} \leqslant 43.1\,\mathrm{N}.$$

Dividing this value by the mass supported by the seat (53.3 kg) the guidance value for the force-related characteristic vibration values for male operators older than 60 years amounts to

$$a_{\mathrm{fe}} = 0.81\,\mathrm{ms}^{-2}.$$

The operators of wheel-loader belong to the group which are often affected by vibration induced disorders of the lumbar spine. On the wheel-loader seats Christ (1988) measured the vertical accelerations which resulted in a weighted acceleration of $a_{\mathrm{we}} = 1.09\,\mathrm{ms}^{-2}$ for a daily exposure of 8 h according to the second edition of ISO 2631-1 (1997). The median frequency was nearly 5 Hz. Considering the relationship in Fig. 5 and Eq. (2) the force-related characteristic vibration value is $a_{\mathrm{fe}} = 1.17\,\mathrm{ms}^{-2}$. For bulldozer and helicopters the corresponding values are listed in Table 2. These values show that wheel-loader operators older than 60 years are experiencing an enhanced health risk by the vibrations whereas for younger operators the risk is lower. The characteristic vibration value a_{fee} increases with the age-related decrease of the strength of the lumbar spine. Furthermore, comparison of the data in Table 2 shows that the force-related weighting results in an assessment of health risk corresponding more to the first edition of ISO 2631-1 (1985) than to the second edition (1997). The corresponding differences between the two editions of the standard are given by change of the weighting factors namely the reduction of the factors at low frequencies and the enhancement at high frequencies in the new edition.

Table 2
Characteristic vibration values of earth-moving machinery and helicopter with low and high median frequencies

	Wheel-loader	Bulldozer	Helicopter
ISO 2631-1 (1985) $a_{\mathrm{we}}(\mathrm{ms}^{-2})$	1.25	0.85	0.2–0.6
ISO 2631-1 (1997) $a_{\mathrm{we}}(\mathrm{ms}^{-2})$	1.09	0.96	0.45
Median frequency (Hz)	5	8.4	20
Force-related value $a_{\mathrm{fe}}(\mathrm{ms}^{-2})$	1.17	0.83	0.26

Guidance values for a daily exposure of 8 h, above these values health risks are likely: $a_{\mathrm{we}} = 0.80\,\mathrm{ms}^{-2}$; $a_{\mathrm{fe}} = 0.81\,\mathrm{ms}^{-2}$.

Under shock containing vibrations the peak values of the seat acceleration can be greater than the plus/minus fourfold root-mean-square value of the vibration signal. In this case for the risk assessment it is necessary to compute the time course of the compressive forces induced by such vibrations. The time courses can be simulated by means of the biomechanical model or by the complex multiplication of the acceleration spectra with the transfer function. The simulation by means of the model is also necessary if the influence of the posture and of the body mass and height upon the compressive forces should be regarded. Then for the risk the maximum force values have to be compared with the strength of the spine.

5. Conclusions

The biomechanical model was developed in order to simulate the forces which act in the joints of the legs and the spine and which can hardly be measured. The ratio between the compressive forces and the vibration acceleration were described by transfer functions. From these functions new weighting factors were derived which enable the computation of force-related characteristic vibration values. The comparison of these force-related factors with the weighting factors listed in ISO 2631-1 showed that low frequency vibration probably results in a greater health risk than it will be estimated by means of the second edition of the standard. This relationship is also figured by the formula, that enables the computation of the force-related characteristic vibration value (a_{fe}) based on the weighted acceleration (a_{we}, rms). Combined with the recommended guidance value the force-related characteristic vibration values can be used for the prevention of spine disorders and in the case of suspicion of occupational disease.

Furthermore, regarding the compressive spine forces uniform criterions for the assessment of the health risk induced by lifting and carrying activities and by whole-body vibration can be developed—e.g. to assess this combination in compliance with the work load requirements set by the German regulation on occupational diseases and to support the prevention.

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