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Review Article

On the Relationship between Whole-body Vibration Exposure and Spinal Health Risk

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Abstract: A conceptual framework provides the possibility to identify factors determining the effects of whole-body vibration (WBV) on the spine and the internal stress-strain relationships. Epidemiological studies were critically evaluated with respect to their significance for the derivation of quantitative exposure-effect relationships. The approach of deriving such relationships from a comparison with self-generated accelerations during daily activities was considered as unsuited. Trunk muscle activity and control with apparently identical accelerations of body parts during self-generated and forced motions differ widely. Simple biodynamic models coupled with experimental *in vivo* and *in vitro* data permitted a preliminary deduction of quantitative relationships between WBV and spinal health with the consideration of individual factors and exposure conditions. Examples of anatomy-based verified finite element models and their application are provided. Such models are considered as a very promising instrument. They can be used to assess quantitatively preventive measures and design. Future research needs concern the examination of (1) the nonlinearity of biodynamics, (2) the effects of WBV in x- and y-axes, (3) the strength of the spine for shear, (4) the contact parameters between the seat and man, (5) the significance of postures and muscle activity, and (6) material properties of spinal structures.

Key words: Epidemiology, Posture, Self-generated Vibration, Model, Seat, Anthropometry, Assessment

Introduction

The relationship between long-term whole-body vibration (WBV) exposure and spinal health has been found to vary considerably[1–4]. Several factors were supposed to cause this variability[5, 6] that was reflected by replacing the former Exposure Limit[7] by the health guidance caution zone[8]. The WBV-exposure is defined as the vibration measured at the interfaces between the machine and the operator, i. e. mainly at the driver seat. The spinal health risk arises from a mechanical damage of anatomical structures due to forces acting on these structures (internal load). The internal forces do not depend solely on the WBV-exposure[9, 10]. Therefore, the assessment of the spinal health risk of WBV-exposed workers must not be restricted to a straightforward relationship between WBV-exposure and effect that underlies, e.g., the simplistic derivation of a fixed long-term dose required as prerequisite for the acceptance of degenerative disorders of the spine as an occupational disease[11, 12].

A clear conceptual framework (Fig. 1) can help to clarify the factors determining the effects of WBV on spinal health and contributing to the internal stress-strain relationship during WBV-exposure. The WBV-exposure itself depends on several factors like the driving speed. Even under laboratory conditions, an identical vibration at the base of the same seat generates different WBV-exposures for different subjects due to the between-subject variability of the repercussion of the subjects on suspended seats[13, 14]. The WBV-exposure causes an acceleration of the human body with related dynamic forces acting on the spine. The

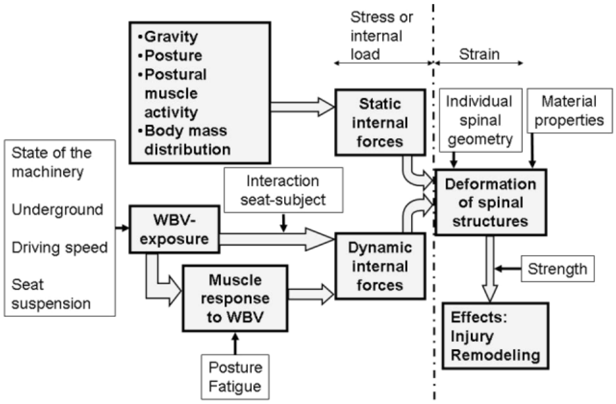

Fig. 1. Conceptual framework of the relationship between whole-body vibration exposure and spinal health.
Main components and relations are marked by grey shade.

significant effects of posture and backrest contact on the transmissibility of WBV to the head[15] suggest the interaction between the subject and the seat to have an effect on the associated dynamic internal forces. Further dynamic internal forces arise from the muscle response with alternating increased and decreased activity[16–19]. Muscles can either exert very high forces on the spine or cause spinal instability by relaxation. Their response to WBV also depends on the posture[20] and muscle fatigue[21]. Sandover[22] draw attention very early to the important role of posture for the development of low back disorders of drivers. Gravity causes the static portion of internal forces with posture, postural muscle activity and body mass distribution as significant variables. Extended and flexed prolonged seated postures can have significantly different effects on the force components in the lumbar spine[23]. Static and dynamic internal forces add up to the stress (internal load) that causes the strain (deformation) of spinal structures. The latter depends on the individual geometry and material properties of spinal structures which determine the stress distribution. The outcome of the strain depends on the strength of spinal structures and their ability to recover from repetitive load. The strength of vertebrae was shown to be a function of their size, mineral content and age[24–26] (cf.[9] for a review).

Using this framework, the paper aims at examining and systematizing the present knowledge on various exposure-effect relationships in order to derive conclusions that can help to improve the risk assessment. Special emphasis will be laid on epidemiological research, experimental studies and modelling. Urgent questions of the evaluation of WBV will be discussed in this context, as the existence of a critical dose, the evaluation of high peak values, the consideration of posture and individual factors, the frequency weighting and effects of WBV in x- and y-axes.

Relationship Associated with Epidemiological Studies

Epidemiological studies have frequently indicated an elevated health risk for the spine in workers exposed for many years to intense WBV. Critical surveys of the literature have been prepared by[1–4, 27, 28]. Bovenci and Hulshof[3] provided a comprehensive table summarising the study and control groups, the magnitude and duration of WBV exposures, study design, methods, prevalence odds ratios (POR) and incidence density ratios (IDR) of numerous papers. The same authors[3] obtained and reported point estimates and 95% confidence intervals of summary POR or summary IDR for different disorders on the basis of the point and interval estimates of POR or IDR presented by the individual cross-sectional or cohort epidemiological studies, respectively. These reviews[1–4, 27, 28] concluded that intense long-term whole-body vibration can adversely affect the spine and can increase the risk of low-back pain, with the latter as a secondary consequence of a primary degenerative change of the vertebrae and/or disks. The lumbar part of the vertebral column was found to be the most frequently affected region, followed by the thoracic region. A high rate of impairments of the cervical part, reported by several authors (cf.[1]), seems to be caused by a fixed unfavourable posture rather than by vibration. Some reports have indicated a significantly higher risk of the dislocation of lumbar disks (cf.[1, 3, 4, 28]). Review papers derived uniformly at the conclusion that exact WBV-exposure-effect relationships could not be obtained by epidemiological studies. This conclusion does not contradict the assumption of the majority of experts that a causal relationship between spinal health and WBV must exist, even though its exact nature is not fully known. Reservation seems to be indicated even if authors claim to have recorded a history of all exposure conditions and examined dose-response relationships as[29]. Considering also the thorough analysis of the original German report[30], some critical points can be mentioned that are more or less applicable also to other epidemiological studies. The record of the WBV-exposure and dose-values by questioning for daily working hours and gathering acceleration values from a data base can surely provide valuable information, but the uncertainty of this kind of information is considerable and has to be taken into account. Workers usually overestimate their working hours. Results of field measurements of WBV-exposure vary to a large extent even for seemingly identical conditions. Different anthropometric characteristics and postures can cause a large variation of the vibration transmission between the floor of the cabin and the seat surface[13]. Hence, the accuracy of such estimates is low and does not justify precise numbers as a "daily reference exposure of $a_{zw(8h)}$=0.6 ms^{-2}" or a long-term dose "D_{VG}=1,414 $m^2/s^4 \times$ days"[29]. Another shortcoming is the restraint to the "Basic evaluation method using weighted root-mean-square acceleration"[8] that is not suited for exposures containing high transients or shocks[8, 31, 32]. As a consequence, possibly different effects of exposures with similar r.m.s.-values, but a different content of high transients, could not be distinguished, although experience and theoretical considerations suggest marked detrimental effects of high peak loads acting on the spine during exposures with high transients[9, 32, 33]. Further reasons for a cautious interpretation of the quantitative results of the study[29] used here as an example arise from the possible causative factors not considered. These were the WBV in x- and y-axes, posture including the use of the backrest, certain anthropometric characteristics, and the age during the exposure. The latter is different from the age during the examination that is usually considered as confounder, but does not necessarily reflect the age-dependent susceptibility during the exposure. The supposed stronger effect of WBV in x- and y-axes is reflected by the multiplying factor k = 1.4, instead of k = 1.0 for z-axis[8], although final evidence for this evaluation is missing. Other studies (cf.[3]) reported only the vector-sum and can, therefore, also not contribute to a better understanding of health-effects related to WBV in different axes. The disregard of a bent forward posture can cause a significant underestimation of the static internal forces contributing to the health risk[9, 20]. The use of the backrest can reduce the load on the spine[15, 34]. In general, differences between subjects have been largely neglected, although selection phenomena suggest they may be of major importance. Particular anthropometric characteristics on a "robust-frail scale" were supposed to be linked with the surface area of lumbar vertebrae and, thus, with the strength of the lumbar spine[35]. The age during the exposure is probably of major importance considering the essential decrease of vertebral strength with increasing age[9, 20, 26, 33]. Hence, identical exposures can be expected to have a rising effect with increasing age, and an exposure of 10 yr duration may be without any harmful effect at an age between 20 and 30, whereas it might have a considerable effect, if it happens at an age between 45 and 55. In summary, the typical shortcomings of otherwise excellent recent epidemiological studies can help to explain missing reliable results with respect to quantitative exposure-effect and/or dose-response relationships.

The changed frequency weighting in ISO 2631-1[8] has to be considered carefully for the interpretation of conclusions derived from earlier studies and review papers as, e.g.,[1, 3, 4, 29]. New values of weighted acceleration in z-axis, e.g., for tractors and wheel-loaders, can be 20 percent lower than those obtained earlier in epidemiological studies with the frequency weighting of ISO 2631/1[7], those for caterpillar trucks can be up to 20 percent higher[36]. Exposure conditions with dominant frequencies below 4 Hz can cause up to 20 percent lower weighted r.m.s.-values than before the changed weighting and might be falsely considered as less hazardous. Hence, daily exposures with weighted r.m.s.-values according to ISO 2631-1[8] between 0.5 and 0.63 ms^{-2} for an exposure time of 8 h in z-axis could have reached the former Exposure Limit 0.63 ms^{-2} of the International Standard 2631/1[7]. Earlier results related to this Exposure Limit should be associated with this range nowadays in order to avoid an underestimation of health risk.

The acceptance of degenerative disorders of the spine as an occupational disease is debated, some studies seem to contradict a causal relationship between WBV-exposure and spine injury. Brinckmann *et al.*[37] examined radiographic views of two cohorts exposed to WBV with respect to vertebral height, sagittal plane displacement and disc height. Their conclusion sounds: "Whole-body vibration has no measurable detrimental effect on vertebral height, sagittal plane displacement or disc height if the machine operators' seats are damped and peak acceleration thus stays below certain limits. It follows that primary overload damage to vertebrae or to discs is not responsible for the high prevalence of back problems in labour forces with sustained exposure to whole-body vibration. Whole-body vibration and shock loading, experienced if earthmoving machines or rock drills are unsprung and operators' seats are unsprung as well, result in a significant decrease in the height of lumbar discs. The comparison of the results from two vibration-exposed cohorts (damped seats vs. unsprung seats on unsprung machines) impressively demonstrates the effect of workplace redesign on the decrease of the health risk of the labour force." There are, however, many reasons - partially mentioned by the authors themselves - that question these statements. There is no information on the selection criteria of the radiographs of the exposed workers. Social conditions of both groups were very different, the selection bias was supposed to be larger in the group with heavier WBV-exposure. The WBV-exposure within the cohort allegedly worked with "unsprung" seats was very inhomogeneous and partially coupled with a constant twisted posture and/or manual material handling with different loads between 30 and 50 kg. More than 20 percent of this cohort performed heavy physical work for about 8 yr before WBV-exposure began. The exposure conditions changed drastically over time from primitive “unsprung” seats to air-damped seats. “Unsprung” seats were used by only some of the exposed workers and for maximally 4 yr within the possible period of 17 yr. Two very different types of machines were used by this group - wheel loaders and drilling machines. There is no information on the distribution of workers with respect to the different WBV-exposures. The other cohort with “damped seats” also used two very different types of machines with “pressure tyres” and “caterpillar chains”, probably linked with different loads. No information is provided on the machine type distribution within the cohort. Clinical symptoms were not known. Considering the well-known discrepancy between radiographs and complaints, the significance of the results for low back pain remains open. The average differences between age of the control and cohorts with “unsprung” and “damped” seats amounted to 6 and 14 yr, respectively. These differences may be very important, because the authors mentioned the limited accuracy of their method to correct the age dependency for the parameters that were used as indicators for overload injuries. Osteophytes were ignored although they can be considered as a sign of degenerative disc disease and were reported to be significantly more frequent and pronounced with tractor drivers[2, 38]. Two studies contradict the result of missing health effects on “Earthmoving machine operators, open-cut brown-coal mine, Köln, Germany”, i. e. the cohort using “damped” seats. Müsch[39] found a significant increase (prevalence of 82.8%) of degenerative changes in radiographs of earthmoving machine operators of the same enterprise compared with a control (prevalence 63.3%). For these workers, Köhne *et al.*[40] reported a higher frequency of the lumbar syndrome together with earlier and more frequent degenerative changes in radiographs. Although Brinckmann *et al.*[37] use a tempting approach to quantify deleterious effects of WBV, the far-reaching conclusions may be misleading. Another study compared very small groups of 18 drivers and co-drivers with 14 men in the control group, the latter on average 12 yr older[41]. External loads by WBV and physical activity were poorly quantified. The average lifetime driving of 492,000 km in the control group suggests a considerable WBV-exposure, additionally combined with a somewhat higher physical load than in the driver group. The age adjustment of lumbar degenerative findings by a linear model seems to be very questionable, since the average age difference between the two groups was large. The small groups and the unknown selection bias raise additional serious doubts. Therefore, the conclusion of the authors that “even extreme vehicular vibration as associated with rally driving does not appear to have significant effects on disc degeneration as measured from MRIs” is hardly acceptable as a generally valid result. In spite of these critical points, such studies deserve attention, because they might help to clarify the reasons that help to sustain high WBV-loads without serious damage. The career of rally drivers began at a mean age of 22 yr[41], an age that coincides with the maximum age-related strength of the spine. The physical fitness of drivers may have been higher than in the control group. The posture of rally drivers, the shape of the seat and the fixation of the drivers to the seat and backrest are probably factors that cause specific, possibly beneficial, conditions for the transmission of WBV that must not be neglected. The phenomenon “selection of the fittest” (cf.[29]) deserves attention especially in sports and may have contributed to the unexpected result of this study.

The existing studies did not permit the substantiation of a no-adverse-effect level (i.e., safe limit) so as to reliably prevent diseases of the spine[4, 5]. There are no clear data showing whether the effects of whole-body vibration on the spine depend on gender. Specific diagnostic features are not known which would permit a reliable diagnosis of the disorder as an outcome of exposure to WBV. A high prevalence of degenerative spinal disorders in non-exposed populations hinders the assumption of a predominantly occupational aetiology in individuals exposed to whole-body vibration. The current state of knowledge does not permit the designation of a minimal intensity and/or a minimal duration of whole-body vibration as a prerequisite for the recognition of an occupational disease. Urgent questions mentioned before - a critical dose, the evaluation of high peak values, the significance of posture and individual factors, the frequency weighting and effects of WBV in x- and y-axes - could not be answered by epidemiological studies. This disappointing situation becomes understandable when the data available in epidemiological studies are compared with Fig. 1. Usually an incomplete description of WBV-exposure is combined with more or less well-defined health effects, while the information shown in the other boxes of Fig. 1 is missing. Future epidemiological studies may try to provide this information, but the necessary effort to obtain a complete data set for a long period according to the conceptual framework (Fig. 1) would be inconceivably high.

Relationship Associated with Self-Generated Accelerations and the Significance of Muscle Control

Up to now, there are some attempts to declare exposures to whole-body vibration or shocks as conditions without risk by referring to comparisons with self-generated accelerations. A recent example was reported by Girvan and Serina[42]. They claimed that “In particular, acceleration measurements can be useful to assess lumbar loading on subjects exposed to sudden vertical motions. A biomechanical analysis can then be performed to assess whether the claimed lumbar inter-vertebral disc injuries resulted from a particular incident.” Because axial compression was considered as a significant factor in causing lumbar injuries, they made comparative measurements of z-axis (vertical) accelerations for two conditions by an accelerometer mounted in an ISO 7096-1982 seat pad and placed between the operator’s buttocks and the seat. The conditions were: (1) Operating a fork lift in which a wheel dropped through a wooden floor, operating a railroad tie remover/inserter machine, driving over railroad tracks and speed bumps; (2) Sitting down into a seat as an example for a typical activity of daily living. Sitting down in the operator’s seat produced higher peak vertical accelerations than the machine operation. Using the accelerations measured, the authors calculated the forces allegedly applied to the spine by multiplying the measured accelerations by the effective mass, i.e., the upper body mass. The authors claimed that “this analysis technique can be a useful tool in determining whether a lumbar inter-vertebral disc was injured in a particular incident. If the measured loads in a particular incident are comparable to or less than the loading experienced during activities of daily living, there is no biomechanical rationale for how the given incident could have caused the lumbar inter-vertebral disc injury.” This approach might look reasonable, but it ignores existing experimental data and knowledge. Accelerations caused by “activities of daily living” like walking, running, jumping can reach an extent[22, 43–46] that could not be tolerated, if these accelerations were produced by WBV[47]. Running and jogging cause accelerations up to 30 ms^{-2} at the head. The majority of people would not tolerate WBV-exposures that would be required to produce such head accelerations, which can be endured by runners without any back problem for prolonged durations. The question arises how these different effects of seemingly identical accelerations can be explained.

Self-generated accelerations require usually a standing posture, whereas WBV acts mainly on sitting persons. Seated postures cause a changed position of the pelvis and lumbar spine[48] chiefly by an elimination of the lumbar lordosis which in its turn changes the conditions for acceleration transmission. Even with relaxed standing, deep back muscles remain active with minor postural changes[49, 50]. The author is not aware of similar results for sitting postures. Except from m. trapezius, pars descendens, the missing or extremely low activity of superficial back and multifidus muscles suggest inactivity with relaxed sitting[18]. The deep short back muscles like m. multifidus and mm. rotatores may be very important as sensors for the feedback in motion control, because they contain between 2 and 6 times more muscle spindles than the long back muscles. Bending forward of the upper part of the body while standing caused a silence of their activity. This result may be important for the relaxed or bent forward sitting posture, too. The postural myoelectric activity pattern of back muscles are different during relaxed, erect and bent forward sitting[18]. Dunk and Callaghan[51] observed a flexion-relaxation reaction during sitting, i.e. a sudden onset of myoelectric silence in the erector spinae during seated forward flexion, more pronounced in the thoracic than in the lumbar part. The common theory explains this reaction with the passive tissues being stretched to a point where they can support the moment imposed on the low back. Flexion relaxation occurred during seated flexion at an average angle of 46.6% of maximum flexion, during standing at a much bigger relative flexion angle of 84.1%. The flexion of the spine occurring during relaxed sitting and more pronounced during bent forward sitting could be important for the WBV-induced injury for several reasons. The low muscle activity means a greater laxity of the lumbar spine due to a reduced muscular stabilization. The significance of co-activation of agonists and antagonists for the stabilization of the spine was stressed by[52, 53] and confirmed for the trunk even near the point of an ideal equilibrium[54]. Possibly, back muscle relaxation during a relaxed seated posture impedes also the mechanism described as “follower load”[55, 56] or “wrapping compression loading”[57]. This mechanism is supposed to reduce the strain by compression, and muscles around the spine are believed to underlie its function. It is closely connected with the muscular stabilization of the spine. Experiments with unexpected external loadings during different degrees of trunk muscle activity speak in favour of an enhanced “effective spine stability”, if the muscle activity prior to the external load is higher[58]. Based on animal studies, Solomonov[59] assumed a prolonged static flexion of the lumbar spine to elicit a complex transient neuromuscular disorder that may last up to several days and consists of four components: An exponential decrease in muscle activity during flexion, spasms superimposed on the decreasing EMG, initial hyperexcitability and a delayed hyperexcitability during rest. Summarizing, the geometry of the spine and the function of deep back muscles during seated postures are supposed to constitute unfavourable conditions for the effects of accelerations on the spine.

The frequency of self-generated rhythmic whole-body motion does not exceed 4 Hz. The number of easily tolerated load cycles during WBV can be much higher due to (1) the higher possible frequencies and (2) lower energy expenditure that does not limit the load duration as the high energy consumption during, e.g., jogging. With increasing running speed, growing shock attenuation within the human body was observed[60]. This result suggests a highly effective control mechanism which does not take place with WBV. In this context, the various extent and control of muscle activity seem to be the decisive factors that might explain the different risks accompanying similar accelerations of body parts. Self-generated locomotion goes along with a pronounced back muscle activity that is significantly higher than that observed with similar accelerations of body parts during WBV in z-axis[43]. If WBV-exposed subjects are sitting relaxed in a bent position, the postural and vibration-synchronous activity of back muscles can disappear almost completely[16, 20]. The higher activity during locomotion is precisely timed[50]. Figure 2, top, illustrates this timing during walking with different speeds. Two kinds of activity are obvious. Several muscles (m.i.t., m.l.t., m.m. in Fig. 2) exhibited two bursts of activities during the ipsilateral heel contact, others (m.rr., m.q.l., m.r.a., m.o.e. in Fig. 2) showed uniformly one burst with both, the ipsi- and contra-lateral contacts. It can be estimated that the muscle activation starts around 200 ms before the heel contact. Similar findings were obtained by later studies[61–63]. Considering the electro-mechanical delay[64, 65], the increased compressive loading of the spine during and shortly after heel contact goes along with its muscular stabilization and the decompression is linked with muscle relaxation. The direct comparison of the EMG-activities during natural movement and WBV-exposure (Fig. 2, bottom) suggested a feed-forward control during self-generated motion in contrast to a reflex-like response during WBV-exposure[64, 66]. The back muscle reaction with two bursts during jumping resembles that during the ipsi-lateral heel contact (Fig. 2, bottom left), whereas a similar acceleration frequency of WBV caused only one burst in the same subject (Fig. 2, bottom right). Similar one-burst muscle responses were observed under WBV-exposure with 2 Hz[16, 45]. This frequency equals about the acceleration

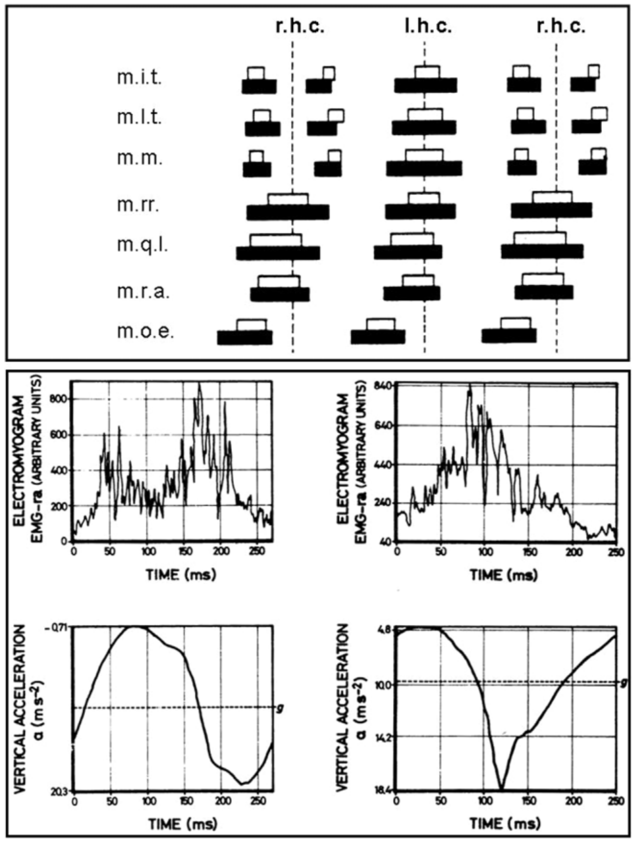

Fig. 2. Timing of myoelectric trunk muscle activity.
Top: During walking with 4.4 km/h (white boxes) and 5.3 km/h (black boxes), EMGs acquired by fine wire electrodes from the right side of the body[50]. r.h.c. = right heel contact, l.h.c. = left heel contact, m.i.t. = m. ileocostalis thoracis, m.l.t. = m. longissimus thoracis, m.m. = m. multifidus, m.rr. = mm. rotatores, m.q.l. = m. quadratus lumborum, m.r.a. = m. rectus abdominis, m.o.e. = m. obliquus externus. Bottom: During jumping on two feet with a frequency of 3.7 Hz (left) and during sinusoidal WBV in z-axis with 4 Hz, 3 ms^{-2} rms (right). EMG-ra = rectified averaged EMG of m. erector spinae at the T12 level acquired with surface electrodes[66], a = averaged acceleration measured above the spinous process T5.

frequency during walking. The comparison of the timing of the myoelectric activity exhibits a difference. During walking, muscles relax before and after the minimal acceleration (uppermost displacement of the body), whereas during WBV the myoelectric activity begins to exceed the postural activation level at the moment of minimal acceleration and increases afterwards.

One crucial question concerns the muscular stabilization or destabilization of the spine during high transients. Figure 3 shows the average myoelectric reactions to transient displacements together with the force at the interface between the subject and the seat[17]. The muscle response depends on the direction of acceleration and the preceding time history

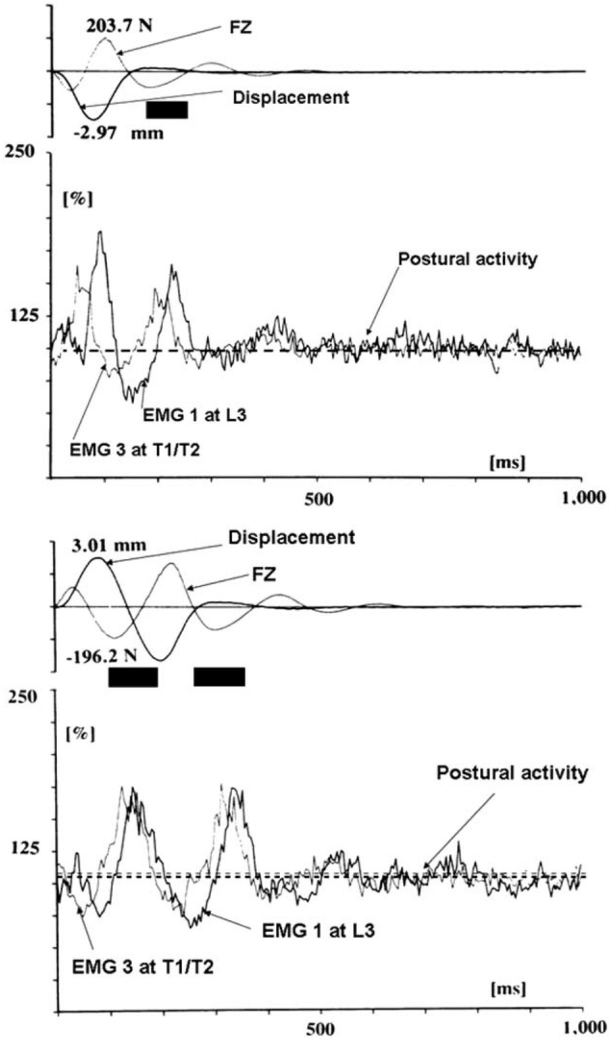

Fig. 3. Back muscle response to transients.
EMGs acquired from the m. erector spinae muscles at the level T1/T2 (EMG1, thin line) and L3 (EMG3, thick line) of 5 subjects were rectified, normalized with respect to the postural activity, and averaged (30 transients for each subject). The amplitudes of the first extreme values of the displacements and the second ones of the force FZ at the interface between the subject and the seat are given by numbers. Top: Half-sinusoidal transient beginning with a downward motion. Bottom: Sinusoidal transient beginning with an upward motion. The black bars symbolize the interval with a lumbar back muscle force falling below the static postural force.

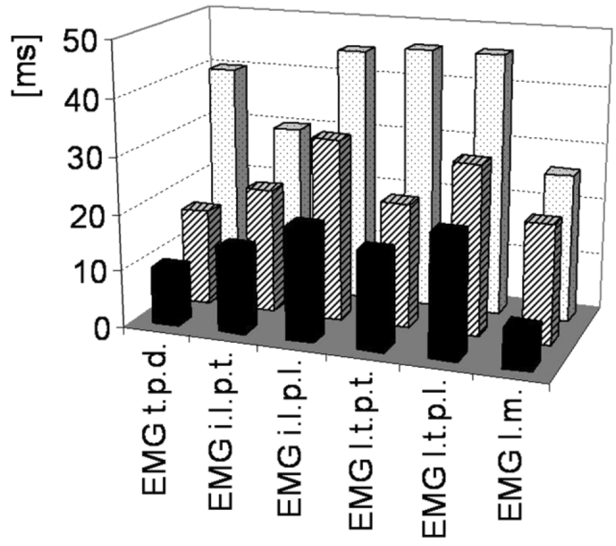

Fig. 4. Median values (8 subjects, 3 postures, 10 transients per peak-to-peak acceleration magnitude) of time differences in ms between the maximum EMG-activity of back muscles and the maximum force measured at the the interface between the seat and the subject.
t.p.d. – m. trapezius pars descendens, i.l.p.t. – m. ileocostalis lumborum pars thoracis, i.l.p.l. – m. ileocostalis lumborum pars lumborum, l.t.p.t. – m. longissimus thoracis pars thoracis, l.t.p.l. – m. longissimus thoracis pars lumborum, l.m. – lumbar multifidus muscle. White colums with dots: 3.5 ms^{-2} peak-to-peak acceleration, hatched colums: 7.2 ms^{-2} peak-to-peak acceleration, black columns – 11.1 ms^{-2} peak-to-peak acceleration.

of transient events. The sudden downward motion is followed by an activity increase of the EMG-activity (Fig. 3, top), the sudden upward motion induces at first a decrease (Fig. 3, bottom). The maximum force FZ after the lowermost displacement coincides with the maximum lumbar EMG (Fig. 3, top), but, due to the electromechanical delay the maximum muscle force can be expected about 50 ms later[64, 65]. The same moment in the sinusoidal transient, the maximum FZ after the lowermost displacement, goes along with a decreasing lumbar EMG-activity (Fig. 3, bottom). These examples demonstrate also that an appropriate muscular stabilization cannot be expected. The predicted muscle force during the maximum FZ after the downward motion at the beginning remains at the static postural level (Fig. 3, top), the increasing FZ during the downward motion coincides with a predicted relaxation (Fig. 3, bottom). Another experiment[18] examined the response to transients with a variable magnitude (from 3.5 to 11.1 ms^{-2} peak-to-peak) that were inserted into a sinusoidal "background"-WBV. Figure 4 demonstrates the median values of the time between the occurrence of the maximum myoelectric activity and the maximum force FZ at the interface between the seat and the subject. Considering the electro-mechanical delay[64], 50 ms would be the optimum difference causing a coincidence of the maximum muscle force and predicted compression. Values near this optimum were observed for low peak accelerations only, with higher peak values the differences become much smaller, i.e., the maximum predicted acceleration-related lumbar disk compression occurs without an increased muscular stabilization. Additionally, high muscle "excess" forces are predicted later when a higher stabilization is not required[18]. In the same experiments, the predicted reflex-like mechanical relaxation of back muscles during the forward bent posture coincided with the expected maximum decompression of disks and the subsequent growing compression, thus suggesting the possibility of high acceleration-related shear loads without muscular protection. High transients caused a great subjectively experienced spinal strain resembling the situation during locomotion, when the feed-forward control mechanism of muscles fails. Such situation can happen, e.g., if someone unexpectedly treads in a hole in darkness or one unforeseen stair at the end of a staircase leads to a high transient acceleration of the body. Back muscle fatigue is a factor that could modify the mechanical muscle response to transients. Muscle fatigue can be caused by a constant seated posture of drivers that may require a low, but constant activation level[67]. A comprehensive review of back-muscle fatigue and its possible significance for WBV-associated injury was presented by Seidel[21]. The seat acceleration-EMG relationships with random WBV were described by Blüthner *et al.*[19].

The incongruity between acceleration-related forces and muscle forces during WBV-exposure may be one reason for deformations of spinal structures that are higher compared with deformations occurring during natural motion. Up to now, the comparison with self-generated accelerations and the examination of muscle control cannot answer questions with respect to a critical dose of WBV, the role of individual factors, frequency weighting and effects of WBV in x- and y-axes. However, it can help to understand the significance of different postures and to evaluate high acceleration peak values of WBV with the muscular stabilization of the lumbar spine as a crucial parameter. The different control of muscle function and clearly different tolerances prohibit the use of the direct comparison between self-generated accelerations and accelerations induced by WBV or shocks as an instrument for the assessment of the relationship between WBV-exposure and spinal health risk.

Relationship Associated with *In Vivo* and *In Vitro* Experimental Data Coupled with Simple Biomechanical Modelling

The first attempt to join *in vivo* experimental biodynamic and EMG-data together with anthropometric characteristics and a simple biomechanical model in order to predict the compressive forces caused by vertical z-axis WBV and acting on the lumbar disc L3/4 was reported by our study[68]. The authors concluded that WBV-related muscle activity cannot be considered as an optimal protecting mechanism. The predicted low extent of decompression was supposed to be an additional handicap for the nutrition of the disc. Two subsequent studies added results with respect to angular motions and shear forces[69] and predicted disc compression during transient WBV[70]. A comprehensive German experimental study with 36 subjects was finished in 1995[20, 71]. Acceleration and force measurements, motion analysis, anthropometric characteristics, electromyography and psychophysics were used for the assessment of stresses in the lumbar spine due to vertical z-axis WBV containing shocks. Simple biomechanical models were used to calculate compressive and shear forces acting on lumbar discs. A comprehensive analysis of the static and repetitive dynamic strength of human vertebrae and bone, for various frequencies and boundary conditions, provided the basis of a theoretical method for the evaluation of repetitive compressive loads on the lumbar spine. This method considered peak accelerations, posture and anthropometric characteristics in order to predict internal static and dynamic loads in relation to the predicted individual strength and Miner's hypothesis. These results were published later in a condensed form[9], critically assessed and applied to practical situations[72]. They enabled the derivation of a critical dose of WBV with the quantitative consideration of the robustness of the skeleton and kind of sitting posture, whereas they could not contribute to elucidate frequency weighting and effects of WBV in x- and y-axes. No reliable conclusion with respect to the assessment of shear loads was possible.

A similar approach was chosen by a Canadian research group[73–75]. Their results were obtained with a hard seat and have formed the basis of the current ISO 2631-5[33]. A critical analysis and discussion of these results, their interpretation and implementation has been recently published[76]. One main result was a method for the prediction of the acceleration of the spine by means of a neural network called "dynamic response model" which simulates a transformation of the seat acceleration into predicted bone acceleration. This method constituted one out of three different methods for the same measure "bone acceleration"[73–75]. Different predicted bone accelerations for the same input acceleration at the seat were deliberately used for the derivation of regressions and comparisons. A fundamental misinterpretation of the magnitude and "frequency of shocks" tested in the experiments was obvious from the figures published[73] due to considering peaks of high-frequency parts (e.g., around 30–40 Hz) of the transient accelerations as the peak accelerations of shocks described as low-frequency shocks (e.g., about 4 Hz). This inaccuracy possibly caused an overestimation of the shock amplitude up to 50% error.

Since the high-frequency peak accelerations are efficiently damped by soft tissue between the seat and skeleton, the use of an overestimated low-frequent shock amplitude could lead, consequently, to an underestimation of the relationship between the real low-frequency shock amplitude and the effect on the spine. This error, the highly questionable procedure to calculate bone accelerations, and the use of only one subject for the derivation of the relation between seat acceleration and predicted intra-spinal force cause serious doubts (for more details cf.[76]).

Common shortcomings of both trials[71, 73, 74] to join experimental data and modelling are the missing reflection of human anatomy by the simple biomechanical models as well as the missing verification of the latter.

Relationship Associated with Model Predictions

This chapter deals with model predictions that are made with more complex models not coupled with experimental data apart from use for the verification of models. Their main purpose is the calculation of forces acting on spinal structures under defined WBV-exposure conditions. Seidel and Griffin[77] reviewed briefly modelling the response of the spine to WBV and repeated shock. Hofman *et al.*[78] analysed recently the state of the art in this area. As Pankoke *et al.*[79], they distinguish two groups of models. Phenomenological models intend to represent special dynamic properties of the real structure, e.g. the living human. These dynamic properties can be expressed by different characteristic functions like the mechanical impedance or the seat to head transmissibility, which both are characteristic functions in the frequency domain. These characteristics can be obtained *in vivo* by experimental procedures. Thus, phenomenological models may be verified in a direct manner with *in vivo* measurements, e.g. of the mean impedance and the mean seat to head transmissibility of a group of persons or a special percentile group.

The development of a phenomenological model starts with the choice of the characteristic function the model shall represent. The basic structure of the model is then derived by an analysis of the chosen characteristics. Different system identification techniques may be used to define this basic structure. In general, the basic structure of the model will not look like a human body at all. Once the basic structure is defined by the number of masses, the number of degrees of freedom (DOFs) and the force transmitting connections between the masses, the system parameters of this basic structure will have to be identified by different parameter identification techniques. The model, which is described completely by its basic structure and identified parameters, will now represent the characteristics it was designed for. A phenomenological model which is designed to represent the mechanical impedance of a sitting man is suitable for a numerical simulation of the interaction between a driver's seat and the driver in dynamic numerical models of a vehicle. Since in most cases the basic structure of the model is a quite simple lumped parameter structure, it is possible to develop and to construct a physical representation of this model, which is then able to perform an experimental simulation of the interaction between a driver's seat and the driver without the need to use living test drivers. From such a model, it is not possible to obtain other information than the information it was designed for, in the case described the mechanical impedance. Thus, a phenomenological model can be only used to reproduce known information. New information about formerly unknown facts like spinal loads cannot be produced. This is the main difference between phenomenological and the second group of models, the anatomy-based models.

The intention of anatomy-based models is to produce still unknown information, e.g. the determination of spinal loads from a given base excitation at the seat. Since this new information cannot be determined by invasive measurements for ethical reasons, it is impossible to verify anatomy-based models directly. The structure of an anatomy-based model shall reflect the real anatomy as well as possible within the intended application of the model. The parameters of the model are not obtained from the examination of the whole system, but from the examination of the properties of biological materials forming its structures. In the ideal case, the complete model is constructed with the right parameters and reflects the mechanical properties correctly. The geometrical shape of an anatomical model has to look more or less similar — depending on the model purpose and complexity — to the geometrical shape of the real structure, the human body. Since the anatomical model can represent body parts that are not accessible by an experimental setup *in vivo*, e.g., the vertebral discs, a prediction of formerly unknown information as the forces and moments in these vertebral discs can be made. The verification of model properties which cannot be measured requires an indirect comparison of the model results that depend on these properties, with measurements. Such a model can be built using different computational methods. The most important methods are (1) the method of finite elements (FE), building a model with structural and/or discrete elements, from which, e.g., strains, stresses, deformations and forces can be derived, and (2) the method of multi-body dynamics, constructing a model from discrete masses, rotary inertias, springs and dampers. From such a model, only deformations and forces can be derived, strains and stresses are not accessible.

The models used in[80–82] do not reflect the anatomy of the lumbar spine. Therefore, these models are less suited for the prediction of forces within this structure. The procedure for the prediction of WBV-induced forces in the lumbar spine described in[33] is based on an poorly described model[74] which is neither based on anatomy nor verified[76]. The recommended[33] extent of the reduction of vertebral strength with age is disputable, because it relies also on static strength data that were obtained with a high load rate of 3 kN/s. A more conservative dependence was recommended by Seidel *et al.*[9]. Buck *et al.*[83] developed a plane dynamic FE-model of sitting man. The model was based on an anatomic representation of the lower lumbar spine (L3 to L5). This lumbar spine model was incorporated into a dynamic model of the upper torso with neck, head and arms as well as a model of the body caudal to the lumbar spine with pelvis and legs. Additionally, a simple dynamic representation of the viscera was used. All parameters of the model were determined according to human anatomy and carefully verified with respect to mechanical properties of biological materials. Pankoke *et al.*[79] developed a three-dimensional anatomy-based FE-model of sitting man adjustable to body height, body mass and posture using the experience with a highly detailed dynamic anatomical FE-model of the whole human body[84] and with the plane dynamic FE-model[83]. Part of the information was taken from a more detailed model of sitting man[84]. The model was validated by experimental results of an experimental study[71] using the driving point impedance and the transfer-functions from seat to head in vertical and horizontal directions as well as the histogram of the seat-force as integral measure in the time domain. Seidel *et al.*[10] used it for the prediction of forces acting on the spine during whole-body vibration. One might argue that FE-models require unacceptable computing time. A possible solution was proposed and applied to field measurements of ride-pad acceleration signals in[10]. In order to simplify the computation, a set of transfer functions from seat acceleration to the WBV-induced internal forces acting on the spine was calculated by FE-models and used for an efficient prediction of spinal loads. The adjustability of the FE-model[79] permitted predictions of compressive and shear stress due to WBV-exposures combined with different seated postures and for persons with a representative variability of body mass and body height. Seidel *et al.*[10, 85] concluded that the variability of the spinal loads for a given whole-body vibration suggests a ratio between the minimum and maximum internal loads of about 1:2, when a normal range of several biological factors is considered with the model calculations. These results supported the extent of the Health-Guidance Caution Zone[8] that had been fixed without a sound basis. A two-dimensional FE-model enabled the simulation of the human responses on a common driver seat[86]. Typical properties of the seat were considered by including the seat's transfer function. The verification of the human interaction with the seat was done using data from measurements[87]. Hinz *et al.*[15, 34] used this model for the prediction of the internal spinal load by considering individual vibration inputs to the buttocks and to the back on a suspended seat. The peak transmissibilities between the accelerations at the seat base and the compressive forces at L5/S1 were highest for the seat without the backrest during a forward bending posture. The authors concluded that the backrest contact and posture should not be neglected in the assessment of

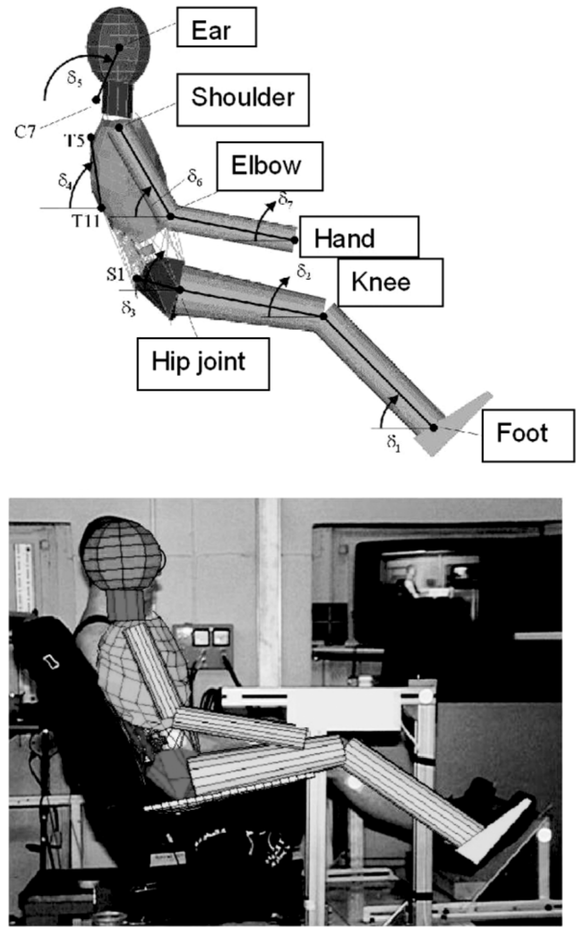

Fig. 5. Schematic drawing of the extended model with angles describing the sitting posture (top) and a photo of subject seated in a car seat with a superimposed FE-model adjusted to the posture (bottom).

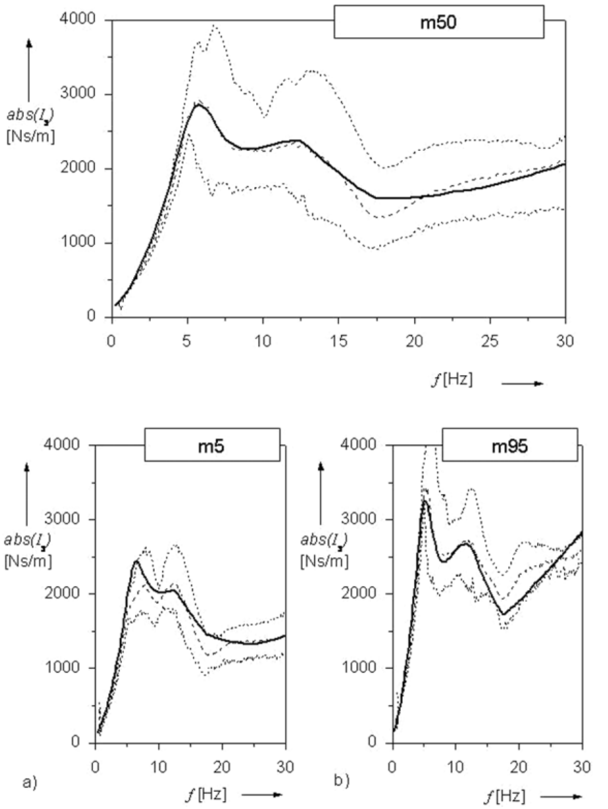

Fig. 6. Verification of the FE-model[78] by measurements of the impedance with a car-typical posture and using a backrest[88] for subjects representing three percentiles of the population m5 (height 1.76 m, body mass 58.5 kg), m50 (height 1.74 m, body mass 74.6 kg), m95 (height 1.81 m, body mass 93.9 kg).

---: range of measurements, —: FE-model calculations.

health risk caused by WBV.

The FE-model[79] was recently extended towards a much more sophisticated model with three essential improvements[78]. (1) A better possibility for the adjustment of the model that is relevant for individual biodynamics. The extended model will be adjustable to additional individual anthropometric data - the standing height, sitting height, chest breadth, chest depth, pelvis breadth and circumference at the waist. These characteristics are assumed to be related to inertial properties, individual health risk and mass distribution. (2) A more detailed reflection of trunk muscles. The extended model will represent the following muscles with the number of fascicles on each side given in brackets: m. rectus abdominis [1], m. obliquus externus [7], m. obliquus internus [5], m. multifidus [10], m. iliocostalis [5], m. longissimus [9], m. quadratus lumborum [5], m. latissimus dorsi [4], m. psoas major [5]. (3) An improved adaptable reflection of the sitting postures by introducing the possibility to characterise the posture by a set of seven angles (Fig. 5). The extended model[78] was verified for different seating conditions and postures. Figure 6 exemplifies the verification of the model for a posture adopted in car seats. For this verification, Hofman *et al.*[78] could rely on a representative data set obtained for different percentiles of the population in the z-axis[88, 89]. The agreement with m50 and m95 is good, the calculation for m5 resulted in a too large first peak. The verification for the dynamic behaviour on a hard seat in x-, y- and z-axes was made by utilizing various data published in the literature[78]. Seidel *et al.*[90] applied these models in order to compare the frequency-dependent load on the spine for different postures and representative anthropometric characteristics with the current frequency weightings in[8]. They used transfer functions from the seat acceleration to the intra-spinal forces in order to describe the exposure-effect relationship. Figure 7 illustrates the considerable variability for the transfer functions from the z-axis acceleration of the seat without backrest to compression. The effect on the spine is predicted to depend on both, the anthropometric characteristics and the posture. The frequency dependence of the transfer function with the bent forward posture reflects the special dynamic behaviour causing a shift of the resonance towards lower frequencies due to a change from a vertical vibration mode near 5 Hz to a bending mode near 2.5 Hz. The results of model calculations are compared with the w_k-frequency weighting[8] in Fig. 8. When the transfer functions are normalized with respect to their maximum set to 1, they can be well compared with the frequency weighting. In Fig. 8, the normalized magnitudes of transfer functions were averaged across 3 percentiles of anthropometric characteristics (m5 – 1.70 m height, 63.6 kg body mass; m50 – 1.74 m, 79 kg; m95 – 1.79 m, 97 kg, cf.[85] for more details), 3 discs (T12/L1, L3/L4, L5/S1) and a variable number of postures. The comparison with the w_k-weighting[8] suggests a certain underestimation of the health effect for frequencies below and an overestimation for frequencies above 5 Hz. Bent forward postures seem to cause an increased risk that is not considered by the current frequency weighting. The comparison of the normalized transfer function from WBV in the x-axis to shear forces with the w_d-weighting[8] suggested a possible overestimation of frequencies above 3 Hz in the x-axis by[8].

Coming back to the questions put in the introduction, the predictions of anatomy-based verified models together with *in-vitro* strength data of the spine can help to find answers. Not one critical dose for all conditions, but several algorithms for the determination of critical doses that vary in dependence on posture, age, and anthropometric conditions could be

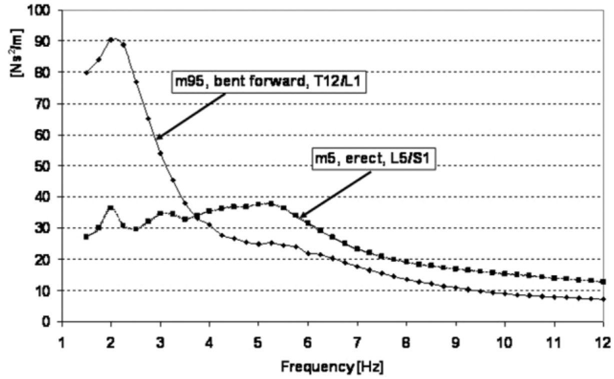

Fig. 7. Variability of the magnitudes of transfer functions from z-axis seat accelerations to predicted compressive forces acting on discs T12/L1 and L5/S1 of subjects belonging to the 95th (m95) or 5th (m5) percentiles (details see text) with a bent forward and erect seated posture on a hard seat without backrest.

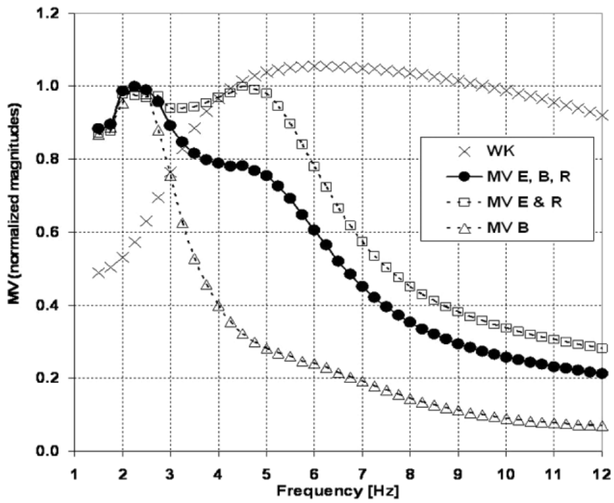

Fig. 8. Averaged normalized (maximum of the average curve set to 1) magnitudes of transfer functions from z-axis seat accelerations to predicted compressive forces.

Averaging across 3 percentiles of anthropometric characteristics (cf. text for details), 3 discs (T12/L1, L3/L4, L5/S1), and a variable number of postures (E: erect, B: bent forward, R: relaxed).

tentatively derived. The algorithms include the evaluation of high peak values and can quantify the significance of posture and individual factors. A uniform frequency weighting for health effects on the spine may not be appropriate for different postures and anatomic locations. There are, however, still important gaps of knowledge and more effort is required to overcome the present limitations and shortcomings of models.

Conclusions and Future Research Needs

The combination of anatomy-based finite element models with *in vitro* data seems to be the best suited approach to establish quantitative exposure-effect relationships. Model predictions can help to prepare political decisions by an estimate of the effects of preventive measures. Manufacturers could use model predictions for an assessment of different designs. Up to now, major uncertainties result from the nonlinearity of human biodynamics, insufficient knowledge on the effects of WBV in x- and y-axes, the unknown strength of the spine for shear loads, the influence of the coupling-characteristics between the seat and man, the significance of postures and muscle activity, and missing reliable data on material properties of the spinal structures. Recent results[91)] indicate a pronounced dependence of the stiffness of the lumbar motion segment on the static preload by compression. Similar results will be crucial for the decision to which extent FE-models shall reflect non-linearities with different magnitudes of WBV and signs of acceleration and/or dynamic internal forces. Rützel *et al.*[92)] published first results on modelling the contact parameters between the buttocks and the seat with verification via pressure distributions[93, 94)]. This kind of research will help to adapt FE-models much better to the conditions of real life, i.e., to the interaction with seat cushions and backrest. Very preliminary data with respect to the determination of the apparent mass of subjects sitting on soft seats[95)] indicate also an important direction of future research. The systematic examination of between-subject differences and gender-related differences[14, 93, 94, 96)] will contribute to an improved adjustment of models to individual conditions and, finally, to the prediction of individual exposure-effect relationships. The future prediction of the individual health risk will require the consideration of the individual geometry and material properties of spinal structures. Hofman *et al.*[78)] developed a submodel, embedded in the global model that permits the calculation of the stress distribution and local strain caused by WBV. Future development of this kind of research might help to detect pathognomic locations of WBV-induced injury and dangerous exposure conditions. Non-invasive studies will be necessary in order to clarify the significance of trunk muscle activity and biodynamic behaviour that is closely related to practical conditions, e. g., behaviour during uni-axial and multi-axial WBV-exposures. Well-designed and carefully directed epidemiological studies could verify predictions of models and experimental research duly considering concomitant factors like posture, anthropometric properties and age during the WBV-exposure.

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References

1) Seidel H, Heide R (1986) Long-term effects of whole-body vibration — a critical survey of literature. Int Arch Occup Environm Health **58**, 1–26.
2) Seidel H (1993) Selected health risks caused by long-term whole body vibration. Am J Ind Med **23**, 589–604.
3) Bovenzi M, Hulshof CTJ (1999) An updated review of epidemiologic studies on the relationship between exposure to whole-body vibration and low back pain. Int Arch Occup Environ Health **72**, 351–65.
4) Lings S, Leboeuf-Yde C (2000) Whole-body vibration and low back pain: a systematic, critical review of the epidemiological literature 1992–1999. Int Arch Occup Environ Health **73**, 290–7.
5) Seidel H, Griffin MJ (1998) Whole-body vibration. In: Encyclopaedia of occupational health and safety. 4th ed. eds. by Stellman JM, McCann M, Warshaw L, Brabant C, 50.2 – 50.7, Geneva ILO.
6) Wilder DG, Pope MH (1996) Epidemiological and aetiological aspects of low back pain in vibration environments — an update. Clin Biomech **11**, 61–73.
7) ISO 2631/1-1985(E) (1985) International standard evaluation of human exposure to whole-body vibration — Part 1: general requirements. 1st ed., International Organization for Standardization, Geneva.
8) ISO 2631-1 (1997) International standard. Mechanical vibration and shock — Evaluation of human exposure to whole-body vibration — Part 1: General requirements. 2nd ed. International Organization for Standardization, Geneva.
9) Seidel H, Blüthner R, Hinz B, Schust M (1998) On the health risk of the lumbar spine due to whole-body vibration — theoretical approach, experimental data an evaluation of whole-body vibration. J Sound Vib **215**, 723–41.
10) Seidel H, Blüthner R, Hinz B (2001) Application of finite-element models to predict forces acting on the lumbar spine during whole-body vibration. Clin Biomech **16** (Suppl 1), S57–63.
11) Dupuis H, Hartung E (1994) Arbeitstechnische voraussetzungen der Berufskrankheit Nr. 2110. Die BG, **5**, 346–9.
12) Schäfer K, Hartung E (1999) Mainz-Dortmunder Dosismodell (MDD) zur Beurteilung der Belastung der Lendenwirbelsäule durch Heben oder Tragen schwerer Lasten oder durch Tätigkeiten in extremer Rumpfbeugehaltung bei Verdacht auf Berufskrankheit Nr. 2108, teil 3, Vorschlag zur Beurteilung der arbeitstechnischen Voraussetzungen im Berufskrankheiten-Feststellungsverfahren bei kombinierter Belastung mit Ganzkörper-Schwingungen. Arbeitsmed Sozialmed Umweltmed **34**, 143–7.
13) Hinz B, Menzel G, Blüthner R, Seidel H (1998) Laboratory testing of operator seat vibration with 37 subjects — critical comment on ISO/DIS 7096. J Sound Vib **215**, 977–88.
14) Hinz B, Menzel G, Seidel H, Blüthner R (2001) Zur Abhängigkeit der Sitzübertragungsfunktionen und -faktoren von Versuchsbedingungen und Frequenzbewertungen. Z Arb Wiss **55**, 1–6.
15) Hinz B, Seidel H, Menzel G, Blüthner R (2002) Effects related to random whole-body vibration and posture on a suspended seat with and without backrest. J Sound Vib **253**, 265–82.
16) Seidel H (1988) Myoelectric reactions to ultra-low frequency and low-frequency whole body vibration. Eur J Appl Physiol **57**, 558–62.
17) Blüthner R, Hinz B, Menzel G, Seidel H (1993) Back muscle response to transient whole-body vibration. Int J Ind Ergonomics **12**, 49–59.
18) Blüthner R, Seidel H, Hinz B, Schust M (1998) Timing of back muscles during whole-body vibration with transients — its significance for the internal spinal load. J Low Freq Noise, Vib Active Control **17**, 215–26.
19) Blüthner R, Seidel H, Hinz B (2002) Myoelectric response of back muscles to vertical random whole-body vibration with different magnitudes at different postures. J Sound Vib **253**, 37–56.
20) Seidel H, Blüthner R, Hinz B, Schust M (1997) Stresses in the lumbar spine due to whole-body vibration containing shocks. Experimental interdisciplinary study—anthropometry, biodynamics, biomechanical model, psychophysics and electromyography. The project formed part of the collaborative project on whole body vibration III; (Final Report). Schriftenreihe der Bundesanstalt für Arbeitsschutz und Arbeitsmedizin Dortmund/Berlin, Forschung Fb 777, 197, Wirtschaftsverlag NW, Bremerhaven.
21) Seidel H (1995) Back-muscle fatigue and whole-body

vibration. In: Proceedings of International Conference, Work and Health; Scientific Basis of Progress in the Working Environment (International Conference, 1993, Copenhagen Denmark). eds. by Svane O, Johansen C, 89–94, European Commission Health and Safety at Work, Directorate-General Employment, Industrial Relations and Social Affairs, Brussels.
22) Sandover J (1981) Vibration, posture and low-back disorders of professional drivers. Research report No DHS 402. 141, University of Technology, Dept. of Human sciences, Loughborough.
23) Hedman TP, Geoffrey RF (1997) Mechanical response of the lumbar spine to seated postural loads. Spine **22**, 734–43.
24) Brinckmann P, Biggemann M, Hilweg D (1989) Prediction of the compressive strength of human lumbar vertebrae. Clin Biomech **4** (Suppl 2), S1–27.
25) Hansson T, Roos B, Nachemson A (1980) The bone mineral content and ultimate compressive strength of lumbar vertebrae. Spine **5**, 46–55.
26) Jäger M (2001) Belastung und Belastbarkeit der Lendenwirbelsäule im Berufsalltag. Ein interdisziplinärer Ansatz für eine ergonomische Arbeitsgestaltung. Fortschritt-Berichte VDI Reihe 17 Nr. 208, 302, VDI Verlag, Düsseldorf.
27) Dupuis H, Zerlett G (1986) The effects of whole-body vibration. 162, Springer, Berlin, Heidelberg, New York, Tokyo.
28) Bongers PM, Boshuizen HC (1990) Back disorders and whole-body vibration at work. Thesis, University of Amsterdam, 318.
29) Schwarze S, Notbohm G, Dupuis H, Hartung E (1998) Dose-response relationships between whole-body vibration and lumbar disk disease — a field study on 388 drivers of different vehicles. J Sound Vib **215**, 613–28.
30) Schwarze S, Notbohm G, Hartung E, Dupuis H (1999) Epidemiologische Studie “Ganzkörpervibration” Verbundforschungsvorhaben im Auftrag des Hauptverbandes der gewerblichen Berufsgenossenschaften. Abschlussbericht. 288, Hauptverband der gewerblichen Berufsgenossenschaften (HVBG), Sankt Augustin.
31) Spång K (2001) Multiple shocks. Investigation of methods for assessment of adverse health effects. Report H-14023-r-B, Annexes A-D, 60, Göteborg, Ingemansson Technology AB.
32) Griffin MJ (1998) A comparison of standardized methods for predicting the hazards of whole-body vibration and repeated shocks. J Sound Vib, **215**, 883–914.
33) ISO 2631-5:2004(E) (2004) Mechanical vibration and shock — Evaluation of human exposure to whole-body vibration — Part 5: Method for evaluation of vibration containing multiple shocks. 20, International Organization for Standardization, Geneva.
34) Hinz B, Seidel H, Menzel G, Blüthner R, Gericke L (2002) Biodynamik und Kraftwirkungen in der Wirbelsäule von Vibrationsexponierten auf Fahrersitzen mit und ohne Rückenlehne. Schriftenreihe der Bundesanstalt für Arbeitsschutz und Arbeitsmedizin, Forschung Fb 948, 198, Wirtschaftsverlag NW, Bremerhaven.
35) Hinz B, Seidel H (1995) Anthropometrische Methoden und Quantifizierung der Wirbelsäulenbelastbarkeit für interne mechanische Belastungen. Arbeitsmed Sozialmed Umweltmed **30**, 487–99.
36) Verein Deutscher Ingenieure (2002) VDI 2057 Part 1. Human exposure to mechanical vibrations. Whole-body vibration. 39, Beuth-Verlag GmbH, Berlin.
37) Brinckmann P, Frobin W, Biggemann M, Tillotson M, Burton K (1998) Quantification of overload injuries to thoracolumbar vertebrae and discs in persons exposied to heavy physical exertions or vibration at the workplace. Part II. Occurrence and magnitude of ovelaoad injury in exposed cohort. Clin Biomech **13**, (Suppl 2), S(2)1–36.
38) Wukasch W (1979) Zur Wirkung langzeitiger Ganzkörpervibrationsexposition auf die Wirbelsäule von Traktoristen: eine epidemiologische Studie. Doctoral Thesis, 131, Akademie für Ärztliche Fortbildung der DDR, Berlin.
39) Müsch FH (1987) Lumbale Bandscheibendegeneration bei Erdbaumaschinenfahrern mit langjähriger Ganz-Körper-Vibrationsbelastung. Thesis, 135, Universität Mainz.
40) Köhne G, Zerlett G, Duntze H (1982) Ganzkörperschwingungen auf Erdbaumaschinen: Entwicklung geeigneter Dämpfungssysteme. Schriftenreihe Humanisierung des Arbeitslebens **32**, 366, VDI-Verlag, Düsseldorf.
41) Videman T, Simonen R, Usenius JP, Österman K, Battie MC (2000) The long-term effects of rally driving on spinal pathology. Clin Biomech **15**, 83–6.
42) Girvan D, Serina ER (2002) Biomechanical analysis of acute lumbar intervertebral disc loading. IV. World Congress Biomechanics, Calgary. CD with abstracts.
43) Dielmann E (1983) Experimentelle Untersuchung zur

Belastung der Wirbelsäule durch mechanische Schwingungen bei verschiedenen Fortbewegungsarten. Dissertation, 90, Johannes Gutenberg-Universität. Mainz.
44) Capozzo A (1982) Low frequency self-generated vibration during ambulation in normal men. J Biomech **15**, 599–608.
45) Griffin MJ (1990) Handbook of human vibration. 988, Academic Press, London.
46) Nigg BM (1994) Acceleration. In: Biomechanics of the musculo-skeletal system. eds. by Nigg BM, Herzog W, 237–53, John Wiley & Sons, Chichester, New York.
47) Magid EG, Coermann RR, Ziegenruecker GH (1960) Human tolerance to whole body sinusoidal vibration: short-time, one-minute and three-minute studies. Aerospace Med **31**, 915–24.
48) Chaffin DB, Andersson, GBJ (1991) Occupational Biomechanics. 2nd ed., 518, John Wiley & Sons Inc, New York, Chichester, Brisbane, Toronto, Singapore.
49) Morris JM, Benner G, Lucas DB (1962) An elctromyographic study of the intrinsic muscles of the back in man. J Anat Lond **96**, 509–20.
50) Waters RL, Morris JM (1972) Electrical activity of muscles of the trunk during walking. J Anat **111**, 191–9.
51) Dunk NM, Callaghan JP (2002) Erector spinae flexion relaxation during short duration slumped sitting. IV. World Congress Biomechanics, Calgary. CD with abstracts.
52) Quint U, Wilke HJ, Shirazi-Adl A, Parnianpour M, Löer F, Claes L (1998) Importance of the intersegmental trunk muscles for the stability of the lumbar spine. A biomechanical study in vitro. Spine **23**, 1937–45.
53) McGill SM (1998) Point of view. Spine **23**, 1945.
54) Cholewicki J, Panjabi, MM, Khachatryan A (1997) Stabilizing function of trunk flexor-extensor muscles around a neutral spine posture. Spine **22**, 2207–12.
55) Patwardhan AG, Havey RM, Ghanayem AJ, Diener H, Meade KP, Dunlap B, Hodges SD (2000) Load-carrying capacity of the human cervical spine in compression is increased under a follower load. Spine **25**, 1548–54.
56) Rohlmann A, Neller S, Claes L, Bergmann G, Wilke HJ (2001) Influence of a follower load on intradiscal pressure and intersegmental rotation of the lumbar spine. Spine **26**, E557–61.
57) Shirazi-Adl A, Parnianpour M (2000) Load-bearing and stress analysis of the human spine under a novel wrapping compression loading. Clin Biomech **15**, 718–25.
58) Cholewicki J, Simons A, Radebold A (2000) Effects of external trunk loads on lumbar spine stability. J Biomech **33**, 1377–85.
59) Solomonow M (2002) Models of neuromuscular disorders in the spine. IV. World Congress Biomechanics, Calgary. CD with abstracts.
60) Mercer JA, Vance J, Hreljac A, Hamill J (2002) Relationship between shock attenuation and stride length during running at different velocities. Eur J Appl Physiol **87**, 403–8.
61) Callaghan JP, Patla AE, McGill SM (1999) Low back three-dimensional joint forces, kinematics, and kinetics during walking. Clin Biomech **14**, 203–16.
62) Thorstensson A, Carlson H, Zomlefer MR, Nilsson J (1982) Lumbar back muscle activity in relation to trunk movements during locomotion in man. Acta Physiol Scand **116**, 13–20.
63) Cappozzo A (1984) Compressive loads in the lumbar vertebral column during normal level walking. J Orthop Res **2**, 292–301.
64) Seidel H, Beyer H, Blüthner R, Bräuer D, Hinz B, Menzel G, Weißmüller A (1985) Electromyography in back research — assessment of static and dynamic conditions. In: Biomechanics: Current interdisciplinary research. eds. by Perren SM, Schneider E, 611–6, Martinus Nijhoff Publishers, Dordrecht, Boston, Lancester.
65) Hinz B, Seidel H (1989) On time relation between erector spinae muscle activity and force development during initial isometric stage of back lifts. Clin Biomech **4**, 5–10.
66) Blüthner R, Hinz B, Seidel H, Erdmann U (1988) Zur Schätzung der auf die Lendenwirbelsäule bei Ganzkörpervibration wirkenden Kompressionskraft. In: Der Mensch unter Schwingungsverhalten, Vorträge zum 75. Karl-Marx-Städter Seminar über Festkörpermechanik und Bruchmechanik. Akademie der Wissenschaften der DDR, Institut für Mechanik, Report No. 12, 35–47.
67) Seidel H, Beyer H, Bräuer D (1987) Electromyographic evaluation of back musces fatigue at repeated sustained constant-force contractions with different levels. Eur J Appl Physiol **56**, 592–602.
68) Seidel H, Bluethner R, Hinz B (1986) Effects of sinusoidal whole-body vibration on the lumbar spine: the stress-strain relationship. Int Arch Occup Environ Health **57**, 207–23.
69) Hinz B, Seidel H, Bräuer D, Menzel G, Blüthner R, Erdmann U (1988) Bidimensional accelerations of

lumbar vertebrae and estimation of internal spinal load during sinusoidal vertical whole-body vibration: a pilot study. Clin Biomech **3**, 241–8.

70) Hinz B, Blüthner R, Menzel G, Seidel H (1994) Estimation of internal spinal load during transient whole-body vibration. Clin Biomech **9**, 263–71.

71) Seidel H, Blüthner R, Hinz B, Schust M (1995) Belastung der Lendenwirbelsäule durch stoßhaltige Ganzkörperschwingungen. Experimentelle interdisziplinäre Untersuchung — Anthropometrie, Biodynamik, biomechanisches Modell, Psychophysik und Elektromyographie. (Teilprojekt des Verbundprojekts Ganzkörperschwingungen III). (Schlußbericht), Schriftenreiche der Bundesanstalt für Arbeitsmedizin, Forschung, Fb 01 HK 061, 199, Wirtschaftverlag NW Verlag für neue Wissenschaft GmbH, Bremerhaven.

72) Sandover J (1998) The fatigue approach to vibration and health: is it a practical and viable way of predicting the effects on people? J Sound Vib **215**, 699–721.

73) Cameron B, Morrison J, Robinson D, Vukusic A, Martin S, Roddan G (1995) Development of a standard for the health hazard assessment of mechanical shock and repeated impact in army vehicles: Phase 4, Data collection. B.C. Research Inc., 3650 Wesbrook Mall, Vancouver, B.C. V6S 2L2, Final report prepared for Contract Report No. DAMD17-91-C-1115, U.S. Army Aeromedical Research Laboratory, 122, Vancouver.

74) Morrison JB, Robinson DG, Roddan G, Nicol JJ, Springer MJ-N, Martin SH, Cameron BJ (1997) Development of a standard for the health hazard assessment of mechanical shock and repeated impact in army vehicles: Phase 5, B.C. Research Inc., 3650 Wesbrook Mall, Vancouver, B.C. V6S 2L2, Project No: 6-06-600, Final report, prepared for Contract Number DAMD17-91-C-1115, MCMR-UAC-E, U.S. Army Aeromedical Research Laboratory, 117, Vancouver.

75) Nicol J, Morrison J, Rodan G, Rawicz A (1997) Modelling the dynamic response of the human spine to shock and vibration using a recurrent neural network. Heavy Vehicle Systems, Int J Vehicle Design **4**, 145–65.

76) Seidel H (2004) Wirkung und Beurteilung von Ganzkörper-Schwingungen—biologische Aspekte. In: VDI-Gesellschaft Entwicklung Konstruktion Vertrieb. VDI-Berichte Nr.1821. Humanschwingungen, Tagung, Darmstadt, 1–24, VDI Verlag GmbH, Düsseldorf.

77) Seidel H, Griffin MJ (2001) Modelling the response of the spinal system to whole-body vibration and repeated shock. Clin Biomech **1** (Suppl 1), S3–7.

78) Hofmann J, Pankoke St, Wölfel HP (2003) Individualisierbares Finite-Elemente-Modell des sitzenden Menschen zur Berechnung der Beanspruchungen bei Vibrationsanregung in verschiednen Raumrichtunen und Stoßanregung an einer Reihe von Körperstellen—Ganzkörpermodell und Submodell der unteren Lendenwirbelsäule. Schriftenreihe der Bundesanstalt für Arbeitsschutz und Arbeitsmedizin. Forschung Fb 994. 187, Wirtschaftsverlag NW Verlag für neue Wissenschaft GmbH, Bremerhaven.

79) Pankoke S, Buck B, Woelfel HP (1998) Dynamic FE model of sitting man adjustable to body height, body mass and posture used for calculating internal forces in the lumbar vertebral disks. J Sound Vib **215**, 827–39.

80) Fritz M (2000) Description of the relation between the forces acting in the lumbar spine and whole-body vibrations by means of transfer functions. Clin Biomech **15**, 234–40.

81) Kitazaki S, Griffin MJ (1997) A modal analysis of whole-body vertical vibration, using a finit element model of the human body. J Sound Vib **200**, 83–103.

82) Yoshimura T, Nakai K, Tamaoki G (2004) Modelling of human body for the evaluation of exposure to whole-body vibration. Computational mechanics, WCCM VI in conjunction with APCOM, Beijing, China. Tsinghua University Press & Springer Verlag, 6.

83) Buck B, Pankoke St, Wölfel HP (1997) Lateralsymmetrisches Modell der Lendenwirbelsäule zur Berechnung dynamischer Bandscheinbenkräfte. Schlußbericht. Schriftenreihe der Bundesanstalt für Arbeitsschutz und Arbeitsmedizin, Forschung, Fb 770.183, Wirtschaftsverlag NW Verlag für neue Wissenschaft GmbH, Bremerhaven.

84) Buck B (1997) Ein Schwingungsmodell des sitzenden Menschen mit detaillierter Abbildung der Wirbelsäule und Muskulatur im Lendenbereich. Dissertation, 232, Technische Universität, Darmstadt.

85) Seidel H, Blüthner R, Hinz B (2000) Ermittlung vibrationsbedingter Belastungsverläufe der Lendenwirbelsäule mit Hilfe dynamischer Vielkörpermodellierung. Schriftenreihe der Bundesanstalt für Arbeitsmedizin Berlin, Fb 889, 59, Wirtschaftsverlag NW, Bremerhaven.

86) Pankoke St, Hofmann J, Wölfel HP (2000) Weiterentwicklung eines Modells zur Berechnung von Kräften, die in der Lendenwirbelsäule wirksam werden

von einem 2-dimensionalen Modell zu einem 3-dimensionalen Modell und von einem Modell für einen dynamischen Input zu einem Modell für vier dynamische Inputs bei Nutzung von realen Fahrersitzen. Schriftenreihe der Bundesanstalt für Arbeitsschutz und Arbeitsmedizin, Forschung, Fb 885. 177, Wirtschaftverlag NW Verlag für neue Wissenschaft GmbH, Bremerhaven.

87) Hinz B, Menzel G, Blüthner R, Seidel H, Gericke L, Schust M(1996) Biologische Wirkung von praxisrelevanten Ganzkörperschwingungen unter Berücksichtigung realer Fahrersitzbedingungen (Schlußbericht). Schriftenreihe der Bundesanstalt für Arbeitsmedizin Berlin; Forschung, Fb 10.004. 297, Wirtschaftsverlag NW, Verlag für neue Wissenschaft GmbH, Bremerhaven.

88) Hinz B, Blüthner R, Keitel J, Menzel G, Seidel H, Wagner W (1998) Durchführung von Personenmessungen als Grundlage einer zu entwickelnden Prüfmethode von Pkw-Sitzen als Beitrag für eine Verbesserung der Gestaltung des Fahrerarbeitsplatzes im Pkw. Unpublished research report fort he contract with the technical university darmstadt, 230, Federal Institute for Occupational Safety and Health, Berlin.

89) Hinz B, Seidel H, Menzel G, Gericke L, Blüthner R, Keitel J (2004) Ermittlung der scheinbaren Masse für die an Pkw-Arbeitsplätzen auftretenden Expositionen in fahrzeugunabhängiger Körperhaltung. Z Arb Wiss **58**, 249–64.

90) Seidel H, Hinz B, Hofmann J, Menzel G (2004) Entsprechen die Frequenzbewertungen für das Beanspruchungskriterium Gesundheit nach ISO 2631-1 und VDI 2057 Blatt 1 der Wirkung? In: VDI-Gesellschaft Entwicklung Konstruktion Vertrieb. VDI-Berichte nr.1821. Humanschwingungen, Tagung, Darmstadt, 183–99, VDI Verlag GmbH, Düsseldorf.

91) Gardner-Morse MG, Stokes IAF (2004) Structural behavior of human lumbar spinal motion segments. J Biomech **37**, 205–12.

92) Rützel S, Mischke C, Wölfel HP (2004) Abbildung einer nachgiebigen Körperoberfläche für ein FE-Menschmodell zur Simulation von Kontaktparametern. In: VDI-Gesellschaft Entwicklung Konstruktion Vertrieb. VDI-Berichte nr.1821. Humanschwingungen, Tagung, Darmstadt, 363–75, VDI Verlag GmbH, Düsseldorf.

93) Hinz B, Gericke L, Keitel J, Menzel G, Seidel H (2002) Untersuchungen der Druckverteilungen an der Kontaktstelle Mensch-Sitz in Abhängigkeit von Sitzen, Haltungen und Körpermassen. Z Arb Wiss **56**, 159–77.

94) Hinz B, Seidel H, Menzel G, Keitel J, Gericke L (2003) Laborstudie zu Druckverteilungen an Pkw-Sitzen — Untersuchungen an Kollektiven von Frauen und Männern mit repräsentativer Verteilung von Körpergröße und Körpergewicht. Z Arb Wiss **57**, 169–87.

95) Hinz B, Rützel S, Keitel J, Menzel G, Seidel H (2004) Bestimmung der scheinbaren masse bei nutzung von pkw-sitzen als voraussetzung für eine modellierung des sitzenden menschen — Ergebnisse von frauen und männern. In: VDI-Gesellschaft entwicklung konstruktion vertrieb. VDI-Berichte nr.1821. Humanschwingungen, Tagung, Darmstadt, 57–86, VDI Verlag GmbH, Düsseldorf.

96) Hinz B, Seidel H, Menzel G, Blüthner R (2001) Zur Variabilität der Impedanz in Abhängigkeit von Intensität, Haltung und Körpermasse. Z Arb Wiss **55**, 7–14.