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THE FATIGUE APPROACH TO VIBRATION AND HEALTH: IS IT A PRACTICAL AND VIABLE WAY OF PREDICTING THE EFFECTS ON PEOPLE?

J. SANDOVER
Jack Sandover Consultant, *Loughborough*, *LE*11 3*LB*, *England*

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The fatigue approach assumes that the vertebral end-plates are the weak link in the spine subjected to shock and vibration, and fail as a result of material fatigue. The theory assumes that end-plate damage leads to degeneration and pain in the lumbar spine. There is evidence for both the damage predicted and the fatigue mode of failure so that the approach may provide a basis for predictive methods for use in epidemiology and standards. An available data set from a variety of heavy vehicles in practical situations was used for predictions of spinal stress and fatigue life. Although there was some disparity between the predictive methods used, the more developed methods indicated fatigue lives that appeared reasonable, taking into account the vehicles tested and our knowledge of spinal degeneration. It is argued that the modelling and fatigue approaches combined offer a basis for estimating the effects of vibration and shock on health. Although the human variables are such that the approach, as yet, only offers rough estimates, it offers a good basis for understanding. The approach indicates that peak values are important and large peaks dominate risk. The method indicates that long term r.m.s. methods probably under-estimate the risk of injury. The BS 6841 *Wb* and ISO 2631 *Wk* weightings have shortcomings when used where peak values are important. A simple model may be more appropriate. The principle can be applied to continuous vibration as well high acceleration events so that one method can be applied universally to continuous vibrations, high acceleration events and mixtures of these. An endurance limit can be hypothesised and, if this limit is sufficiently high, then the need for many measurements can be reduced.

1. INTRODUCTION

Although many would agree that the available data indicate that shock and vibration probably do lead to health problems, epidemiological research has not given us any understanding of the processes by which shock and vibration affect health nor do they offer predictive models to work with. A better background understanding and a better conceptual basis are needed if methods of magnitude assessment and accumulation, and measurements of the health effect are to be sufficiently reliable to facilitate the generation of dose–response relationships. One needs some theoretical concepts or hypotheses before one can hope to proceed (e.g., epidemiological investigations need a range of possible predictive methods to evaluate against found morbidity rates and vibration exposure patterns).

Modelling tissue damage mechanisms and human response to acceleration offers a suitable starting point. It must be cost effective to explore the theoretical issues as far as possible and as early as possible. To produce useful information, a better understanding need not be in full, validated and cross-checked detail but simply a sufficiently sound basis

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for the development of more rewarding epidemiological investigations, targetted engineering solutions and acceptable standards and regulations.

Although cervical pain, gastrointestinal, cardiovascular and renal problems have been noted in many epidemiological investigations of vehicle operators, by far the most prevalent health problem proves to be low-back disorders as demonstrated by reported low-back pain and radiographically observed disorders—see references [1–4].

There seem to be three main hypotheses as to how shock and vibration can lead to back disorders: vibration increases creep effects; vibration leads to an imbalance so that the spine is at risk from high load situations such as lifting and handling; dynamic loading leads to fatigue damage to the vertebral end-plates or reduced nutrition, and this, in turn, leads to increased degeneration. The first two hypotheses appear to have a smaller following in the literature and, as yet, offer no useful basis to relate high acceleration events and degree of damage or risk. On the other hand, the fatigue hypothesis receives considerable support [3, 5–7] and offers the possibility of developing prototype dose response relationships for use in epidemiological and other research.

The aim of this paper is to look at the fatigue approach and current data as applied to high acceleration events in a critical manner with a view to considering the viability of available data, improvements and possible simplifications.

2. METHOD

The investigation comprised a review of available information on tissue fatigue and analysis of available data. The analysis is based on exposure to high acceleration events—see reference [8], which the fatigue approach suggests are the major source of health effects. These were used to estimate the viability of fatigue life predictions in practice and to consider whether simple predictive methods might be used as a practical alternative.

3. TISSUE DAMAGE AND FATIGUE

3.1. DAMAGE AT THE END PLATES

In vitro testing has shown that the endplates are the first to fail under compression [9, 10] due to failure arising from excessive deflection of the underlying trabecular bone [11], whilst disc modelling [12] leads to the same conclusion. There is plenty of evidence of damage to the end-plates and subchondral bone [13–15] with resulting callus tissue and several authors refer to the possibility that shock and vibration lead to such damage and that this can reduce disc nutrition [16]. Some authors suggest that the damage may be fatigue induced and Sandover [17–19] hypothesised that shock and vibration can result in fatigue induced failure at the end-plates or subchondral bone and that this leads to reduced nutrition and degeneration, either directly or via callus formation. This general concept was suggested as early as 1960 by Rosegger and Rosegger [20]. In contrast, it has to be pointed out that Carter *et al*. [21] argued that the risk of fatigue damage is so high that it has to be regarded as a physiologic phenomenon with continuous bone remodelling.

If the fatigue hypothesis is true and the fatigue behaviour at all similar to that of many inorganic materials, then high acceleration events are likely to have a strong influence on spinal health—a small number of high spinal stress events are more likely to lead to fatigue damage than continuous exposure to low level stresses.

Based on the fatigue hypothesis, the relevant issues become the following: the relationship between some measure of seat acceleration and the consequent pressure or force at the endplate; the fatigue processes that lead to damage; the relationship between damage and degeneration, including any repair processes.

At a practical level, the first issue gives information for some form of weighting or modelling process, whilst the other two give information on how to treat and reduce the resulting data.

3.2. ESTIMATES OF END-PLATE STRESSES FROM SEAT ACCELERATION

The lumbar spine is relatively mobile and changes in geometry with level. Thus, stresses are likely to be complex and influenced by complex motions of the seat. Were researchers to investigate all these complexities, progress would be very slow. Because the vertical z accelerations dominate in many vehicles, a sensible way forward would be to concentrate on these. As long as researchers are aware of this simplification, this allows progress in terms of the gathering of useful information and the development of techniques and indices for the primary issues that can be adapted in the future to tackle the other issues.

The functional anatomy of the lumbar spine will influence forces induced in the area by exposure to shock and vibration at the seat. At the same time, the functional anatomy of the lumbar spine is very much posture dependent. Bending in the lumbar spine occurs during exposure to vibration [22], and the pelvis may rock relative to the seat [23], and these may be the source of the resonances at about 5 Hz often observed. Huijgens [24] argued that the spine is stiff (natural frequency several hundred Hertz) in compression whilst in bending, the resonance frequencies can be of the same order as those observed in practice. What one does not know, of course, is if the posture assumed to be good for static loading is also good for dynamic loading—although one could argue that the lordosis of the "good" posture allows more bending, some relief of the transient compressive loading on the discs and less transmission up the spine whereas a more kyphotic posture results in transient loads being transmitted fully up the spine so that the discs and endplate are subject to the full compressive force along the spine. The trunk centre of gravity position also reduces muscle loading in the lordotic posture compared with kyphotic, where the centre of gravity is further forward. A moderately kyphotic or straight spine posture may not be the best, but is probably representative of what occurs in heavy vehicle driving practice. This means that estimates of spinal force low in the lumbar spine based on input force become more acceptable.

3.2.1. *In vitro methods*

Guillon and his colleagues [25, 26] measured intradiscal pressure in cadavers sitting in a car seat whilst vibrated and impacted and one can use these data for a rough estimate of spinal pressures *in vivo*.

3.2.2. *Invasive methods*

Disc pressure. The measurement of disc pressure *in vivo* is extremely invasive and unlikely to be ethically acceptable. Only a few people have carried out such experiments, e.g., Nachemson [27]. From *in vitro* measurements, Nachemson established that the intradiscal pressure is 1·5 times the applied load divided by the disc area. This suggests some pressure concentration over the disc area. Berkson [28] carried out *in vitro* measurements with complex loading and found that the disc pressure was 1·46 times applied load divided by disc area with a pre-load of 400 N. He also found that flexion and lateral bending resulted in significant disc pressure increase so that with a compressive pre-load of 400 N or an additional torque of 10·6 Nm in flexion or lateral bending (at about 5°) both led to a 0·3 MPa increase in pressure.

Vertebral pins. Another approach that is more acceptable is to insert pins into the vertebral spinous processes and use these for measurements of motion. This still requires local anaesthesia and one wonders if subject behaviour, muscle tone, etc., are the same as in normal circumstances. However, the method results in a good connection between a transducer and the vertebral body so that, as long as the full motion in all directions is considered, then a good measure of intervertebral motion should be possible. This should facilitate good estimates of intradiscal pressure although the *in vivo* stiffness of the disc may differ from the *in vitro* data available and pressure distribution may need to be taken into account.

Christ and Dupuis [29] filmed the motion of pins located in L4, L2 and T12 during exposure to sinusoidal motion of 10 mm peak to peak at the seat using one subject. Sandover and Dupuis [22] reanalyzed the data of Christ and Dupuis. They used the spatial motion of the pins to estimate motion at the vertebral centroids assuming a centroid to skin surface distance of 65 mm. They found significant angular motion both of vertebrae in space and relative to each other at frequencies up to about 5 Hz. Above 5 Hz, these angular motions decreased rapidly with frequency. They found that the maximum vertical motion of the centroid, the maximum angular motion of the vertebrae and the maximum relative angular motion between vertebral pairs all occurred at 4 Hz. The values were approx. 20 mm, 6° and $4{\cdot}5^\circ$ peak to peak respectively for L4 and L4 $\sim$ L2. They found relative displacements (compression) between L2 and L4 of approx. 2 mm peak to peak at 3 and 4 Hz, 1 mm peak to peak at 4·5 Hz and 0·9 mm peak to peak at 5 Hz. These are equivalent to 1·6, 0·9, 0·35 and 0·26 mm peak to peak per m/s^2 r.m.s. seat acceleration respectively. However, they considered that their error level was of the order of 1 mm for displacement measurements so that calculation of compressive forces and pressures from these data would be unreliable, although they do give ballpark figures. The relative angular motion between adjacent vertebrae in the L2 to L4 region was 1° peak to peak per m/s^2 r.m.s. seat acceleration at 4 Hz and $0{\cdot}6^\circ$ and $0{\cdot}45^\circ$ at 4·5 and 5 Hz respectively.

Pope *et al*. [30] used a special strain gauge device to measure all relative motions between pins inserted in L3 and L4 (1 subject) or L4 and L5 (2 subjects). The subjects were exposed to sinusoidal vibrations of 5 and 8 Hz at approximately 0·5, 1·0 and 1·5 m/s^2 r.m.s. They found translational and rotational motion to be greater at 5 Hz than at 8 Hz. They also found greater motion if the back was supported (by the arms) compared with unsupported whilst flexing at 20°. They argued that this demonstrated that muscle forces had a stabilizing role. Measured motions of relevance here were: relative axial motion at 8 Hz of 0·07, 0·29 and 0·08 mm peak to peak (mean 0·15), and at 5 Hz of 0·18, 0·52 and 0·78 mm peak to peak (mean 0·49).

3.2.3. *Non-invasive methods*

Input force. The force between the body and its supporting surface is clearly an important function as well as being easy to measure. If the point of interest is close to the interface, then the input force is more likely to reflect internal stresses than at more remote points. However, some bending low in the spine and rocking of the pelvis are probable and, together with muscle forces will modify the force pattern in the lumbar spine. The apparent mass function translates seat acceleration to input force.

Skin mounted accelerometers. The use of skin mounted accelerometers to estimate vertebral motion is an attractive non-invasive approach and was used early in vibration research [31, 32]. However, there are drawbacks. The skin moves relative to the underlying bone so that Pope *et al*. [33] showed that the relative motion between a pin fixed to L3 and a motion transducer fixed to the skin was greater than the motion of the pin during exposure to vibration.

Hinz *et al*. [34] and Smeathers [35] developed methods of estimating vertebral acceleration from surface acceleration by means of an assumed single degree of freedom model of the tissue motion over the bone. The parameters of this model were obtained by plucking the accelerometer and observing the transient motion of the accelerometer on a stationary spine. Kitazaki and Griffin [36] argued that this method of estimating natural frequencies and damping ratios from time domain analysis is difficult because of the high damping involved and used a slightly different approach based on the half power points of the frequency response function. They evaluated their correction method by changing the accelerometer mass and showing that, after correction, the different mass systems produced the same results. They observed large differences between individuals and sites. They argued that the method was suitable for frequencies below about 35 Hz (for T3). Their illustrations demonstrate increased apparent transmissibility at frequencies above about 10 Hz if correction is not used.

One problem with correction methods is the reduced response of a single degree of freedom system at higher frequencies and low accuracy in estimation of response at these frequencies also. The resultant correction system has high gain and low accuracy at high frequencies so that the resulting estimate of underlying bone acceleration can indicate acceleration transmissibilities that are much higher than seat to head transmission and apparent mass spectra would suggest possible. Morrison *et al*. [37] sought to overcome this with a two part correction factor, one for low and one for high frequencies, although some of their data still indicate short duration transient accelerations that some find suspect.

The data of Sandover and Dupuis [22] can be used to consider errors arising from relative angular motion of adjacent vertebrae. Assuming that the centroid accelerometer distance is 70 mm and that the L4 $\sim$ L3 relative angular motion is $2{\cdot}25^{\circ}$, calculations will show that at 4 Hz, the angular motion will give rise to an increase in measured linear acceleration of $4{\cdot}6\ \mathrm{m/s^2}$ above what should be $12{\cdot}6\ \mathrm{m/s^2}$ i.e. 37%. In the same way, the relative angular acceleration at L4 $\sim$ L3 will lead to an over-estimate of relative displacements between vertebrae of about 3 mm pp under the given vibration conditions.

The author would argue that the use of skin mounted accelerometers to estimate vertebral acceleration has to be viewed with caution until more satisfactory evaluations are forthcoming. One needs a comparison such as that of Lafortune *et al*. [38] (who compared bone and skin mounted accelerometers on the tibia and developed transfer functions for individuals) for such evaluation, although the relative bending motions of vertebrae mean that, even with a good correction method, there could be errors in acceleration estimates. One has to bear in mind if spinal stresses are the ultimate target, then relative motion between vertebrae, usually of the order of only 1 mm or less, have to be obtained from double integration of two sets of vertebral acceleration estimates, a recipe for error. The possibility of surface travelling waves influencing skin mounted accelerometers cannot be discounted.

3.2.4. *Non-invasive methods coupled with modelling*

There are several non-invasive measurements that give some information on the stresses that may occur in the lumbar spine. Coupled with modelling of the spine, this offers a useful approach to estimation of stresses at the veretebral motion segment.

The force at the seat—buttock interface already mentioned is clearly an important measure, especially if stresses in the lower spine are of interest. However, muscle forces and intra-abdominal pressure may lead to additional stresses as the body attempts to stabilize the trunk. A simple balancing of forces and moments about a disc centroid is often used to estimate the stress at the endplate [7, 17].

Muscle activity can be estimated from the electromiogram (emg). However, estimates of muscle force from the emg require calibration for each test with each individual. If the muscle arrangements (attachments, depth and directions) are complex, then estimates of muscle force are complex. McGill and Norman [39] pointed out that estimates of load arising from muscle action depend on a variety of factors such as which muscle groups are assumed to act, their lines of action and moment arms at the vertebrae. Their argued that a more realistic model could, in some circumstances, reduce estimated compressive loads at L4/L5 by up to 35% and that any simple model should use an "equivalent" moment arm of 75 mm rather than the 50 mm often used. On the other hand, Wilder *et al*. [40] found that the true balance point was posterior to the disc centroid by about 10% of the disc width (i.e., about 5 mm) and this would reduce the muscle moment arm. They also found that the load history affected the balance point location.

The intra-abdominal pressure clearly plays a role, although pressure levels may be small compared with those arising from coughing, etc. It is the only internal force that might reduce compressive loads on the spine. However, intra-abdominal pressure increase requires increased abdominal muscular activity and both McGill and Norman [41] and Nachemson *et al*. [42] (both cited in reference [43]) argue that increased intra-abdominal pressure leads to a net increase in spinal compression.

Seidel *et al*. [7] carried out a comprehensive and detailed investigation of the biodynamic behaviour of 36 subjects (young and fit 20 to 24 year old military recruits) exposed to shocks. They used extensive anthropometric measurements and dynamic posture measurements to estimate centre of gravity positions, muscle lever arms, etc., during exposure to impacts of subjects grouped from anthropometric measurements as having "frail", "intermediate" or "robust" body types. They considered the effects of posture and body type on the resulting load estimates. Seidel *et al*. used two basic models for their estimates of spinal load. One (Model 1) centred on the measured apparent mass, the other (Model 2) made use of spinal accelerations estimated from surface measurements. Most of their results relate to the use of Model 1. They used three postures, a bent forward posture, an upright posture and a "driving" posture.

They presented their results in terms of linear regression equations relating peak accelerations measured at the seat and the resulting stress peak at points in the lumbar spine. They found posture to be a very important variable whilst body type had some, though smaller, effect also. An important variable proved to be the static load although this might be less important in some practical situations where peak seat accelerations are very high.

Although they acknowledged that their weighting filter which emulated the ISO 2631: 1985 curves could lead to over-amplification of peak values, they used peak weighted seat accelerations for most of their work. It is interesting to note that unweighting seat accelerations, where included, gave higher correlations's in most of their tables. Because Model 1 depends on the subject apparent mass, a weighting related to the apparent mass may have given even higher correlations's.

Seidel *et al*. found that, although dynamic reaction forces at the seat increased from frail to robust subjects, the effect of decreased disc diameter resulted in frail subjects having higher disc pressures. However, body type is a much less important variable than posture (which will, in practice, vary considerably, even with one individual over the day). Consequently, the most pessimistic regression equation for disc pressure (frail subject) coupled with the most sensible posture (driving) were used for the investigation below. To compare with other authors, the equation for L5/S1, predicted force plus 1 *SD* is used here as the most pessimistic.

3.2.5. *Comparison of force estimates from the above data*

Estimates of spinal stress arising from typical high acceleration events are now compared. The estimates are intended to be qualitative only so that a rough comparison can be made between a variety of methods. A more detailed analysis of the Seidel *et al*. [7] method as applied to a range of field measured high acceleration events is presented later. The following assumptions are made unless otherwise indicated.

The events consisted of oscillatory events at 2 to 3 Hz with peak values up to about 10 m/s^2 and short pulses with peak values of 20 m/s^2 whose energy is concentrated between 10 and 20 Hz. It is further assumed that such events occur 100 times per day.

The spine configuration shows some kyphosis so that the L5/S1 interface is only at a small angle to the horizontal.

The system behaves linearly so that vibration data can be extrapolated to impact data.

Fatigue and failure data can be extrapolated from low to high strain rate conditions. (These last two are both contentious assumptions).

End-plate area is 1800 mm^2; [44], 1790 mm^2 (*SD* 300) for males.

Vertebral motion segment axial stiffness is between 1 and 4 MN/m [17].

The weight of the body above L5/S1 is 60% of total body weight [45] and total body weight is 75 kg.

During sitting, the lumbar spine supports the upper body mass so that the load in the lumbar spine is about 450 N. Apparent mass data indicate that at 3 and 10–15 Hz respectively, this force will be about 20% higher and 50% lower: 540 and 225 N respectively. At 10 and 20 m/s^2, these will be approximately doubled and tripled so that the estimates for peak load at L5/S1 will be 1080 N for the 10 m/s^2 oscillations and 675 N for the 20 m/s^2 pulses.

One can show from the Guillon *et al*. [25, 26] data that the resulting spinal pressures during sinusoidal motion are about 0·03 MPa per m/s^2 at 4 Hz and 0·015 MPa per m/s^2 at 10–12 Hz. The equivalent spinal forces would be 36 and 18 N per m/s^2 respectively, upon using 1800 mm^2 disc area. Thus both the oscillations and the pulses would lead to peak loads of about 360 N.

In the Pope *et al*. [30] study, the relative motion at 8 Hz, 0·98 m/s^2 r.m.s. was on average 0·15 mm peak to peak and, at 5 Hz, 0·49 mm. Upon using the above assumptions on motion segment stiffness, these would lead to between 1700 and 6900 N for the oscillations and between 1080 and 4330 N for the pulses (both overestimated slightly as different frequencies are involved).

The *SD*1 (upwards) and the *HSD*2 (upwards) data of Hinz *et al*. [46] have most energy at 3 and 8 Hz respectively and their peak values were 3·27 m/s^2 (range: 3·15–3·43) and 2·63 m/s^2 (2·48–3·00) respectively and these are used for comparison. Their predicted spinal compressive forces for these were respectively 328·6 N (202·4–724·4) and 482·2 N (382·0–549·7). This would result in spinal forces of 1004·9 N (590·1–2300·0) for the oscillatory events and 3667·0 N (2546·7–4433·1) for the pulses. The *SD*3 data has a flatter power spectrum but it is a maximum in the range 9–12 Hz which may be more representative of the sharp peaks. In this case, their peak values were 1·33 m/s^2 (1·21–1·43) and their predicted compressive forces 442·0 N (164·7–755·8). This would result in spinal forces for the pulses of 6646·6–(2303·5–12492·6).

Robinson *et al*. [47] exposed subjects to high acceleration events consistently of the first oscillations of a decaying 5 Hz sinusoidal pulse. They measured seat acceleration, muscle activity, intra-abdominal pressure and skin accelerations at L4 and T3 to model the compressive forces at L3/L4. For a 9·81 m/s^2 peak seat acceleration, they estimated the dynamic spinal compression to be 4000 N. For shocks up to 40 m/s^2, they estimated

TABLE 1

Estimates of high acceleration event load in the lumbar spine—values in N

Source	Oscillatory events 3 Hz, 10 m/s² peak	Pulse events 10–20 Hz, 20 m/s² peak
Apparent mass data	1080	675
Guillon *et al*.	360	360
Hinz *et al*.	1005	3670 or 6650
Pope *et al*.	1700–6900	1080–4330
Robinson *et al*. (1)	4000	–
Robinson *et al*. (2)	1000	2000

extremely high forces—beyond *in vitro* vertebral strength values. Using a more recent model (Robinson, personal communication) their data indicate forces of approximately 1000 N for 10 m/s^2, 4 Hz oscillatory events and 2000 N for 20 m/s^2, 10–20 Hz events. The above data are summarized in Table 1.

As the estimates are based on a very simple analysis of available data, it is suprising that they are generally of the same order. As one might expect, the apparent mass data and cadaveric pressure data give lower estimates as they do not include the effects of muscle activity. The Pope *et al*. data are reduced in value by the weak information on motion segment stiffness—better estimates could well bring them into line with the others. Clearly, *in vivo* data are important for such estimates.

3.3. FATIGUE PROCESSES AT THE END-PLATE AND SUBCHONDRAL BONE

In engineering materials, fatigue is related to the yield of material at atomic levels leading to micro cracks which grow and eventually lead to failure. The fatigue behaviour of materials such as metals is commonly expressed in terms of an "S–N" or "Wöhler" curve as a logarithmic relationship between the cyclic stress (σ), static strength (σ_u) and the number of cycles to failure (N) so that $N = (\sigma/\sigma_u)^s$. Very often, the S–N curve is, to all extents and purposes, asymptotic at some value of cyclic stress so that the material will not fracture after an infinite number of cycles. This level of cyclic stress is often called the threshold value or endurance limit.

The summation of a number of different stresses is often assumed to follow the Palmgren–Miner hypothesis so that $\Sigma_i\, n_i/N_i = 1$ where n_i and N_i relate to the number of cycles at stress level i and N the number of cycles to failure at that stress level. In practice, failure often occurs at summed ratios of the order of 0·5 rather than 1.

The main issues then become the following: Does the target material behave in this way? What is its static strength? What is the value of exponent s?

3.3.1. *Fatigue behaviour of vertebral bone*

In the case of biological tissues, bone certainly appears to have a typical fatigue behaviour [48], and the exponent s is large so that a small number of large stress cycles are more important than a large number of small stress cycles.

The fatigue behaviour of bone has been investigated extensively [21, 48–50]. Lafferty [48] found that the exponent s was about $-9{\cdot}95$ for a range of materials (human and animal) and type of loading (though not including compression). He found that behaviour was cycling frequency independent below 30 Hz. However, Caler and Carter [50] found that frequency had a strong influence when comparing 0·02 and 2·0 Hz.

Although a number of people have investigated the fatigue behaviour of the vertebral motion segments (see; *e.g*., references [10, 51, 52]), two investigations stand out.

TABLE 2

Probability of failure (%) after exposure to specified number of cycles, after Brinckmann et al. [54]

Normalized stress as % static strength	Number of cycles 10	100	500	1000	5000
60–70	10	55	80	95	100
50–60	0	40	65	80	90
40–50	0	25	45	60	70
30–40	0	0	10	20	25
20–30	0	0	0	0	10

Hansson *et al*. [53] exposed 17 cadaveric motion segments to a 0·5 Hz sinusoidal dynamic compressive loading regime. The segments were cycled to failure, which included an audible popping and/or a sudden increase in axial deformation. Their observed failure damage was Schmorl's node and central end-plate fracture—i.e. primarily related to failure of the cancellous bone beneath the end-plate. Crush or burst fractures were observed in two specimens that failed in the first cycle. Using bone mineral content to predict static strength, they related cycles to failure to percentage of static strength and determined an exponent s of $-13{\cdot}54$ with a correlation coefficient of 0·70. However, if one eliminates the two specimens that failed in the first cycle, the correlation reduces significantly and a much larger exponent (e.g. -8) might easily fit the data.

Brinckmann *et al*. [54] carried out an extensive study of 70 motion segments exposed to a 2 s rise time triangular (0·25 Hz) compressive regime. They used adjacent vertebrae to estimate static strength [44]. Failure was recorded when a step occurred in the deformation–time curve. This usually resulted in extrusion of bone marrow. Most of the fractured specimens had damage at the end-plate. Their relationship, normalised cyclic stress versus cycles to failure, probably shows less scatter than the data of Hansson *et al*. Their data indicate lower fatigue strengths than those of Hansson *et al*. and they attributed this partly due to the fact that Hansson *et al*. tested at room temperature which would normally result in much higher fatigue strengths than at body temperature. They did not attempt to fit an analytical function such as the Wöhler relationship to the data "due to the lack of a theoretical model". They found that at loads less than 30% static strength, failure was rare and they argued that a normalised stress of 30% could be regarded as an endurance limit for *in vivo* exposure. They produced the table of probability of failure shown in Table 2.

Their investigation was related primarily to activities such as lifting and handling and they considered that their data represented the lower limits of fatigue strength as their rise time was low. They pointed out that damage due to vibration should be investigated separately as fatigue strength increases with shorter rise time [55, 56].

3.3.2. *Static strength*

The literature indicates that there is significant variability in static strength and that a variety of factors influence static strength. Mital *et al*. [43] mentioned age, gender, body weight, spinal level, spinal components, loading, posture, physical activity, type of specimen and whether one measures trabecular bone or the cortical shell of vertebrae. However, the experiments of Brinckmann *et al*. [44] suggest a simpler, more logical approach—that static strength depends primarily on disc/vertebral area and bone density.

Disc area increases caudally and bone density varies individually but tends to decrease with age and is influenced by gender.

Jäger *et al*. [57] investigated some 13 data sets from the literature. They gave the range of strengths varying between 3 kN to about 13 kN with a mean and Standard Deviation (*SD*) of 5·81 kN (*SD* 2·58) for male spines (3·97 and 1·50 respectively for female). They presented the following equation to predict compressive strength from various factors:

$$\text{Compressive strength (kN)} = (7{\cdot}26 + 1{\cdot}88G) - (0{\cdot}494 + 0{\cdot}468G)A + (0{\cdot}042 + 0{\cdot}106G)C - 0{\cdot}145L - 0{\cdot}749S.$$

Here G is 0 female, 1 male; A is decade of age; C is area of cross-section in cm^2; L represents lumbar level (0 for the L5/S1 disc to 10 for the T12/L1 disc) and S is 1 for vertebra, 0 for disc strength. This indicates 2 kN difference between male and female.

Brinckmann *et al*. [44] also investigated the static strength of vertebrae and found that it could be predicted accurately with knowledge of endplate area and bone density, the relationship being

$$\text{Compressive strength (kN)} = 0{\cdot}32 + 0{\cdot}00308 \times \text{bone density (mg/ml } K_2\text{HPO}_4) \times \text{endplate area (cm}^2).$$

The correlation coefficient was 0·80 and standard error of the estimate of 1·06 kN. They presented similar equations with differentiation between male and female, under and over 50 years and for levels T10–L1 and L2–L5.

From the Brinckmann *et al*. data, one can obtain the following. Mean area L5/S1, male, is approximately 20 cm^2 (L4/5 18 cm^2) and mean bone density, male, is approximately 100 mg/ml K_2HPO_4. Hence predicted compressive strength is 6·5 kN for L5/S1 and 5·9 kN for L4/5 (Standard Error of mean about 1 kN: −1 kN inactive person, +1 kN active person). This compares with Jaeger for fourth decade male, L5/S1, 20 cm^2 of 7·5 kN, L4/5, 18 cm^2 of 6·6 kN.

Seidel *et al*. [7] carried out an extensive review of available data on static strength of vertebral endplates and of the fatigue behaviour of bony materials. Although Brinckmann *et al*. [42] demonstrated that the static strength of the endplates could be predicted from measures of endplate area and bone density, Seidel *et al*. argued that bone density is not easy to measure and used these data and those of Hansson *et al*. [13] to develop relationships to take age into account. They argued that the static strength of the end-plates was (for age 50 years) 3·21 MPa with 5% and 95% values of 1·89 and 5·06 MPa respectively. For a 20 cm^2 disc area, these are equivalent to 6·4, 3·8 and 10·1 kN respectively.

3.3.3. *The exponent s*

As already noted, Hansson *et al*. [53] estimated the exponent to be −13·54 but Brinckmann *et al*. [54] did not consider their data suitable for expression in the normal logarithmic way.

Seidel *et al*. [7] examined these data and those of Lafferty *et al*. [58] and Michel *et al*. [59] for bone in some detail. They argued that the published data did not take into account the influence of the static loading on specimens and introduced an "equivalent cyclic stress" which is the sum of the "normalized cyclic stress" and the "normalized static stress". They argued that the fatigue limit is usually about 0·2 or 0·3 times the static strength and then reworked the data (using 0·2) and calculated that the exponent for the Lafferty data should be −7·47 instead of the published value of −9·66. Even after their correction method, the Hansson *et al*. and the Brinckmann *et al*. data showed considerable scatter and they considered that the Michel exponent of $-8{\cdot}1$ offers the most sensible starting point.

Seidel *et al*. [7] pointed out that the concept of a threshold value meant that a single fatigue approach might be applied to both long term vibration and repeated shocks. The threshold value may be sufficiently high to allow one to disregard most of the long term vibration time history. They calculated the threshold to be 4 to 6 m/s^2 *Wb* weighted peak for weaker spines and 11 to 13 m/s^2 for average spines.

3.4. DEGENERATION AND REPAIR

Collagen (such as in the intervertebral disc) is slow to repair whereas bony tissue with a good blood supply repairs quickly. Brinckmann *et al*. [44] argued that the turnover rates for bone are such that repair will be small in less than two weeks.

However, although damage such as micro fractures to the endplates and sub-chondral bone repair, this may be with callus. This is less permeable so that the result is reduced nutrient supply to the nucleus and eventual degeneration.

In the case of endplate damage, it is assumed that, over time, the reduced nutrition results in degeneration—the source of back problems. Because the nucleus has poor nutrient supply routes, there is little spare capacity so that each instance of reduced nutrition route capacity will have a degenerative effect that is small to start with but increases with time. Some form of integration of damage to predict degeneration of spinal tissues seems plausible.

4. ESTIMATES OF FATIGUE LIFE IN PRACTICE

The high acceleration events obtained by Sandover [8] were used to estimate fatigue life under normal working conditions using the methods of Seidel *et al*. [7] and a simple apparent mass approach. For the former, the BS 6841 [60] *Wg* weighted algorithm was used to simulate the ISO 2631: 1985 [61] weighting by using a modification of the filter algorithm published by Lewis [62]. For the latter, a single degree of freedom model with an undamped natural frequency of 4 Hz and damping ratio 0·25 was used to simulate an apparent mass function.

For each event, the peak values were obtained. Seidel *et al*. [7] have argued that there is a threshold, below which health effects are unlikely. Accordingly, the peak detection algorithm was designed to ignore all peaks less than $+/-5$ m/s^2.

The data set was used for estimates of stresses at L5/S1 by using a simple apparent mass approach and methods of Seidel *et al*. [7] and for estimates of fatigue life based on Brinckmann *et al*. [44]. The data set was also used to estimate fatigue life directly by using Seidel *et al*. [7] methods.

4.1. FATIGUE LIFE FROM FORCE AND END-PLATE PRESSURE ESTIMATES

For each high acceleration event the force at L5/S1 was estimated from the product of the negative peak acceleration value after treatment by the apparent mass model and the measured mean sitting weight of Seidel *et al*. subjects (about 50 kg). As the peak to peak value is probably more relevant for fatigue estimates, the peak to peak value of the apparent mass model output was also used.

Seidel *et al*. gave two regression equations for estimates of spinal force–one from the unweighted negative peak acceleration (a_{min}) and one for the ISO 2631: 1985 [61] weighted value (a_{wmin}). For L5/S1, driving posture, these are

$$\text{force (N)} = 63{\cdot}55 \times a_{min}(\text{m/s}^2) - 491.82,$$

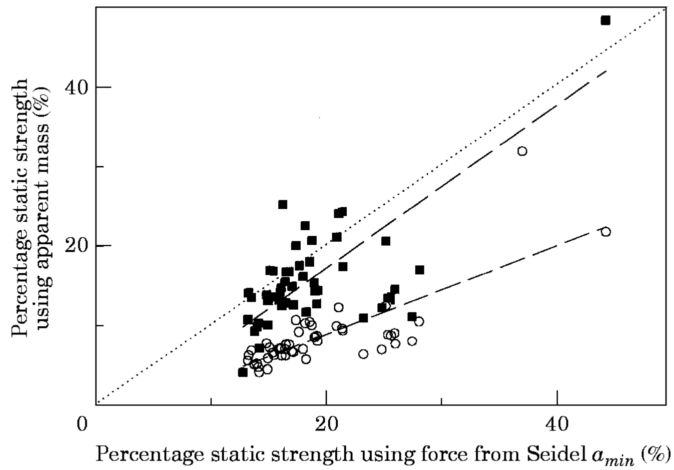

Figure 1. Comparison of percentage static strength estimates obtained by using the apparent mass approach and one Seidel *et al*. method. ○, App mass min; ■, app mass peak to peak; ——, regression line.

and

$$\text{force (N)} = 51{\cdot}50 \times a_{wmin}(\text{m/s}^2) - 531.65,$$

(these are mean values; the values plus one *SD* are clearly higher).

They also gave regression equations to estimate absolute cyclic stress (see section 4.2 below). The equations for a 40 year old male in the driving posture are

frail body type, *mean plus SD*:

$$\text{pressure (MPa)}$$
$$= -0{\cdot}054994 \times a_{wmin}(\text{m/s}^2) + 0{\cdot}151009 \text{ (including static stress component)},$$

robust body type, *mean*:

$$\text{pressure (MPa)}$$
$$= -0{\cdot}048092 \times a_{wmin}(\text{m/s}^2) + 0{\cdot}014776 \text{ (including static stress component)},$$

These give a high–low range of cyclic pressure values and were combined with high–low disc area estimates of 20 cm^2 and 18 cm^2 respectively to give high and low estimates of force.

All the above force estimates were divided by 6500 to give an estimate of applied stress in proportion to a static strength of 6500 N (see section on static strength above) to allow use of the probability of failure predictions of Brinckmann *et al*. [54].

Figure 1 shows that estimates of percentage static strength from the apparent mass model response minimum are generally lower than those obtained from the Seidel *et al*. force prediction methods. However, predictions of percentage static strength from the apparent mass model response peak to peak values are of similar magnitude to the predictions from the Seidel *et al*. method. Of course, this figure also reflects the relationships between predicted spinal forces.

4.2. FATIGUE LIFE PREDICTION FROM SEIDEL *ET AL.* [7]

Seidel *et al*. produced a system to estimate fatigue life which can be summarized as follows: calculate age normalised and posture related static stress; calculate age normalized, posture and body type related cyclic stress from a_{wmin}; thence calculate the equivalent cyclic stress Y_{eqi}; calculate of the number of cycles N_i at which failure is likely to occur for an equivalent stress Y_{eqi} by using the formula $\log(N) = -8{\cdot}140544 \log(Y_{eq02}) + 5{\cdot}804797$; sum the quotients $1/N_i$. Health risks are unlikely for $\Sigma 1/N_i < 1$.

The relevant equivalent cyclic stress regression equation for L5/S1 of a 40 year old male of intermediate body type in the driving posture (using the mean plus *SD*) reduces to:

$$Y_{eq} = -0{\cdot}1712788 \times a_{wmin}(\mathrm{m/s^2}) + 0{\cdot}4354865$$

(plus 0·211907 for the static stress component)

(acceleration convention used, negative accelerations denote upwards displacement).

The predicted cycles to failure can then be obtained from the above logarithmic relationship. This was calculated for each high acceleration event using a single minimum acceleration value and the results are given in Table 3.

The Seidel *et al*. method leads to predicted cycles to failure varying from under 10 to over 10 000—see Figure 2.

Upon using available rates of occurrence observed in the original tape recorded data, the exposure times to failure varied from about 80 min–500 h. However, two vehicles (see below) account for all the very short exposure times and, apart from these, the times varied from 6–500 h. A time to failure of 100 h would be roughly equivalent to two weeks exposure, the time considered by Brinckmann *et al*. [54] to be the minimum tissue repair time and therefore likely to lead to long term degeneration. Quite apart from the two particular vehicles, several cases exceeded this level.

Two vehicles lead to very short fatigue life predictions. One was a "terminal tractor" (T12–T24) which was driven around a track with various obstacles and for two obstacles this resulted in very high accelerations which might not be encountered frequently. The high acceleration events of the other vehicle (D2–D8) all included very short, high acceleration transients of about 25–50 ms duration. These were usually tracked faithfully by the weighting filter and led to very high values of a_{wmin}.

The spinal force predictions coupled with the Brinckmann *et al*. [54] approach to fatigue produced only a few instances where the predicted value was greater than the endurance criterion of 30% static strength (all from the "terminal tractor" data). This might suggest that the Seidel *et al*. fatigue life predictions are too short. However, the variability in bone density and disc area between people is such that the 30% criterion is likely to be exceeded in many cases.

Clearly, many variables are involved in such predictions, for instance, which disc area and bone density has one chosen for predicted static strengths, has one used a mean value or mean +/− one *SD* to include some consideration of worst cases? However, it is encouraging that the various predictions are of the same order, despite the fact that, in many cases, the events were different in character to those on which the Seidel *et al*. regression equations were based.

4.3. CAN ONE SIMPLIFY MEASUREMENTS FOR FATIGUE LIFE PREDICTION?

The peak to peak value is normally used in fatigue life investigations with a periodic stimulus. However, in practice, the stress time history is usually complex so that automatic

TABLE 3

Estimates of fatigue life for each event (for details see text); the vehicles, conditions and normal vibration measures are described fully in reference [8]

		Ratio estimated force to static strength as a percentage						
File name	Seidel cycles to failure	From Seidel a_{min}	From Seidel a_{wmin}	From Seidel slender 20 cm	From Seidel robust 18 cm	From App Mass min	From App Mass pk to pk	VDV $ms^{-1.75}$
F1	—	—	—	—	—	—	—	2·0
F2	—	—	—	—	—	—	—	2.1
B1	336	16·3	16·9	23·2	15·0	13·0	25·2	4·1
B2	5626	17·4	13·5	15·9	9·3	10·6	20·0	3·1
B3	17213	13·3	12·4	13·6	7·5	6·2	14·1	2·3
B4	—	—	—	—	—	4·3	10·0	3·0
B5	—	13·5	—	—	—	6·9	13·5	2·5
B6	—	14·9	—	—	—	7·7	13·9	2·2
J1	—	—	—	—	—	—	—	1·2
J2	—	—	—	—	—	—	—	1·3
J3	—	18·0	—	—	—	7·0	16·2	3·8
J4	14547	16·5	12·5	14·0	7·7	7·6	16·7	3·7
J5	—	14·0	—	—	—	5·2	9·9	1·8
J6	12769	16·8	12·7	14·2	7·9	7·6	16·8	3·0
J7	—	—	—	—	—	—	—	2·2
J8	3902	21·1	13·8	16·7	9·9	12·2	24·0	4·2
J9	—	12·8	—	—	—	—	—	2·4
J10	—	16·5	—	—	—	7·0	15·5	2·7
J11	—	15·9	—	—	—	7·0	13·2	2·7
J12	—	—	—	—	—	—	—	1·1
J13	—	—	—	—	—	—	—	1·2
J14	13805	15·1	12·6	14·1	7·8	7·2	16·9	3·7
J15	—	15·5	—	—	—	6·2	13·6	3·0
J16	16908	17·1	12·4	13·7	7·5	6·7	14·9	3·0
J17	3704	18·2	13·9	16·8	10·0	10·2	22·5	4·0
J18	10675	17·7	12·8	14·6	8·2	9·2	17·5	3·3
J19	8740	18·8	13·0	15·0	8·5	10·0	20·6	3·3
J20	9653	15·4	12·9	14·8	8·4	6·5	16·8	3·2
D1	3704	17·2	13·9	16·8	10·0	6·6	12·6	3·1
D2	83	27·5	19·1	27·9	18·7	8·0	11·0	5·5
D3	417	23·3	16·6	22·6	14·5	6·3	10·9	4·2
D4	173	26·0	17·9	25·3	16·7	9·0	14·5	5·9
D5	93	25·7	18·9	27·5	18·4	8·7	13·5	5·8
D6	169	24·9	17·9	25·4	16·8	6·9	12·2	5·1
D7	95	26·1	18·8	27·4	18·3	7·6	13·1	7·0
D8	88	25·4	19·0	27·7	18·5	8·8	13·4	5·5
T1	4205	18·6	13·8	16·6	9·8	10·4	18·0	4·8
T2	784	18·3	15·7	20·8	13·1	5·7	11·7	4·4
T3	570	21·4	16·1	21·7	13·8	9·6	24·2	6·9
T4	2353	19·1	14·4	17·9	10·9	8·5	14·3	3·9
T5	—	14·2	—	—	—	4·1	7·1	2·2
T6	229	19·0	17·4	24·4	16·0	8·4	15·3	5·7
T7	2542	19·2	14·3	17·7	10·7	8·7	12·7	4·6
T8	956	20·9	15·5	20·2	12·7	9·9	21·0	4·3
T9	—	—	—	—	—	5·7	13·9	2·5
T10	—	19·3	—	—	—	8·0	14·4	2·6
T11	7426	15·0	13·2	15·3	8·8	5·8	13·1	2·6

TABLE 3 (*Continued*)

		Ratio estimated force to static strength as a percentage						
File name	Seidel cycles to failure	From Seidel a_{min}	From Seidel a_{wmin}	From Seidel slender 20 cm	From Seidel robust 18 cm	From App Mass min	From App Mass pk to pk	VDV $ms^{-1\cdot75}$
M1	—	16·2	—	—	—	6·2	12·5	3·9
M2	—	14·9	—	—	—	4·4	10·0	3·4
M3	—	—	—	—	—	—	—	1·9
M4	—	16·1	—	—	—	—	—	2·2
M5	16757	15·1	12·4	13·7	7·5	—	—	3·0
M6	—	12·9	—	—	—	—	—	3·9
M7	—	—	—	—	—	—	—	3·1
M8	—	—	—	—	—	—	—	1·4
M9	—	—	—	—	—	—	—	2·2
T12	2420	16·1	14·4	17·8	10·8	7·1	14·7	3·4
T13	—	12·8	—	—	—	4·1	4·1	2·7
T14	—	—	—	—	—	—	—	1·0
T15	—	13·8	—	—	—	5·1	9·3	2·8
T16	3221	16·5	14·0	17·2	10·3	6·2	12·8	3·2
T17	16029	13·2	12·5	13·8	7·6	5·6	10·8	2·4
T18	62	28·2	19·6	29·0	19·6	10·4	16·9	6·6
T19	13	44·4	22·7	35·6	24·8	21·6	48·1	9·9
T20	—	14·1	—	—	—	4·7	10·3	2·6
T21	130	21·5	18·3	26·3	17·5	9·3	17·4	5·4
T22	2	37·1	27·1	45·1	32·2	31·7	59·5	13·4
T23	62	25·2	19·6	29·0	19·6	12·4	20·5	6·0
T24	2789	16·0	14·2	17·5	10 5·	7·1	14·2	4·3

identification of the relevant peaks and troughs is difficult, although the problem is less when the acceleration data have been treated with a low damped model such as that used to simulate the apparent mass. In this investigation, a travelling "window" was used to minimise the effects of minor changes that would produce a spurious peak and accelerations less than 5 m/s^2 were ignored but the algorithm still identified some "peaks"

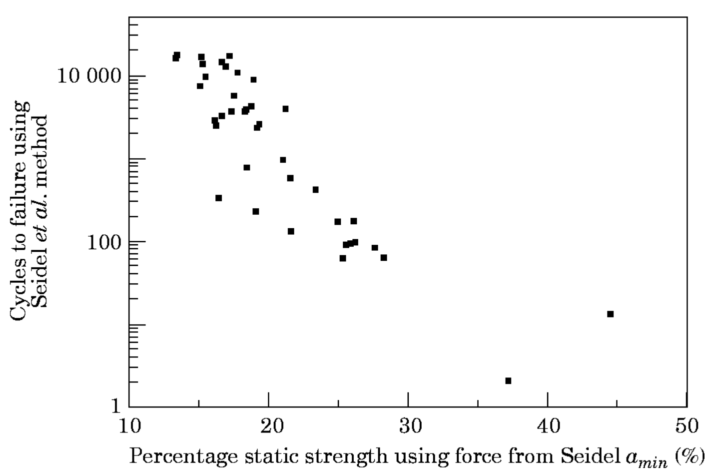

Figure 2. Fatigue life predictions for the full data set.

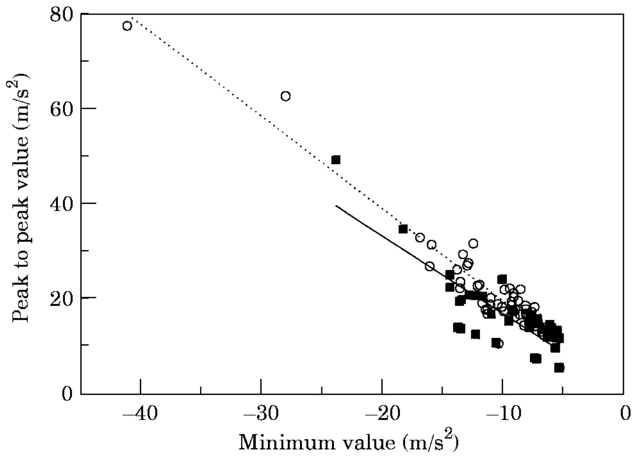

Figure 3. Minimum acceleration as a predictor of peak to peak acceleration for BSI *Wg* (ISO 2631—1985) treated events and apparent mass model treated events. ○, apparent mass; . . . . . , regression line, $r^2 = 0{\cdot}92$; ■, BSI *Wg*/ISO 2631—1985; ——, regression line, $r^2=0{\cdot}71$.

that appeared doubtful. The identification of the major peak–trough or trough–peak pairs to obtain peak to peak values needed human assistance to avoid anomalies. The "rainflow" count method [63] identifies and uses all successive peaks and troughs and may offer a useful method of overcoming this problem in practice.

A simple peak counting method was used here. To simplify analysis, only the largest peak and trough in each event were considered in the estimates of fatigue life. This was based on an assumption that because values to the power 7 or 8 would be used, the effects of the largest peaks and troughs would dominate. In the event, it was found that although the largest dominated in many events, the assumption could not be relied upon. This indicates underestimation of risk in some cases.

Although a reliable algorithm to obtain the peak to peak value for adjacent positive and negative peaks is difficult to realise, Seidel *et al*. were able to obtain a relationship between negative peak values and the peak to peak for their limited range of high acceleration events. An attempt was made to see if this would apply for a wider range of events in this data set. Figure 3 shows that the minimum acceleration value for a high acceleration event and the maximum peak to peak value are related to a certain extent for the ISO 2631: 1985 weighted peaks. However, in the case of high acceleration events weighted with the apparent mass function, the resulting time history is much smoother with the results that the correlation is higher so that the minimum might well provide a useful prediction of the peak to peak value. These data may present a pessimistic view as peaks less than $+/-5$ m/s^2 were rated as zero.

5. DISCUSSION

5.1. DIRECTION

Although vibration in all three orthogonal directions and even rotational components are thought to be important as regards discomfort, there is no useful information on their importance as regards health. If one accepts the concept of end-plate damage then disc/end-plate pressure is the basic stressor. It might be assumed that compression of the spine and vertical *z* direction accelerations only need be considered. However, bending and shear loads can increase disc pressure [28], so that horizontal *x*, *y* and rotational seat accelerations may have significance. Also, vertical vibration can lead to significant bending in the lumbar spine [22]. Meiling *et al*. [64] have argued that one should calculate accelerations relative to the spinal axes rather than assume that the spine is always at right angles to the seat—another complication for field measurements.

Although accelerations in the vertical direction are much greater than in the other directions for many vehicles, this may not be the case for all vehicles. However, restriction to vertical seat accelerations may be a necessary simplification initially and this approach has been followed here.

5.2. SHORTCOMINGS

Several issues have been ignored that may influence the predicted forces and fatigue in the lumbar spine. In the following instances the data may underestimate risk: the data are for male subjects—females with smaller vertebrae and lower bone density may be at greater risk; the models assume that the muscles do no more than balance forces so that the resulting compressive force values may be conservative; the models do not take increases in disc pressure from bending into consideration; the models assume that disc pressure is equally distributed over the disc area—it is most likely that some pressure concentration occurs although the Nachemson 1·5 increase may take pressure concentrations into account; the analysis used a static strength value for L5/S1 of 6·5 kN—however, if one were to consider a reasonably fit person with low bone density and disc area, 6·5 kN may still overestimate the static strength; Seidel *et al*. stated that their procedure is more likely to under-estimate than over-estimate internal loads.

In the following instances the data may over-estimate risk: Seidel uses mean $+1$ *SD* in most cases; high loading rates lead to increased stiffness [65] and increased static strength [55, 56]—the fatigue data of Brinckmann *et al*. [54] are for a triangular stress signal with a rise time of 2 s—their experiments were geared towards activities such as lifting and handling and they pointed out that the data may overestimate fatigue effects for vibration—this is likely to be very important for the short (20 to 50 ms) transients; Seidel *et al*. assumed the endurance limit to be 20% of static strength whereas Brinckmann *et al*. found it to be approximately 30%.

Clearly inherent variability needs to be considered and the Seidel *et al*. [7] approach highlights postural and inter-subject variables. In the case of standards for industrial exposure to shock and vibration, this is an important issue. A standard can include consideration of probability of injury but needs a defined population to do so. For instance, it would seem incorrect to extend the probability of injury to take into account data from cadaver specimens from old, bedbound patients.

5.3. THE HIGH ACCELERATION, SHORT DURATION TRANSIENTS

The peak acceleration values were often related to sharp transients of about 25 to 50 msec duration.

Seidel *et al*. [7] made it clear that one cannot assume their results to apply to conditions not similar to those used. Their seat acceleration transients contained energy mostly in the 2–4 Hz region with very little at 20 Hz and this would cover the oscillatory events of many vehicles. However, the sharp transients are a case where the data cannot be expected to apply. The apparent mass reflects the seat interface force and would indicate significant attenuation at 10–20 Hz. However, the ISO 2631: 1985 weighting at 20 Hz is about the same as for the 2–4 Hz range so that the effect of the sharp transients is likely to be over-estimated by the Seidel approach.

The reduced angular motions at higher frequencies observed by Sandover and Dupuis [22] indicate reduced bending and simpler modelling requirements for these sharp transients.

A delay occurs between an emg signal and the resulting muscle contraction force. This is thought to be about 50 ms. This suggests that in the case of the 20 ms duration transients, the motion may be complete before the muscles come into play so that they will not increase the load on the spine. Thus, the input force might be more acceptable as a measure of the force at L5/S1.

5.4. WEIGHTING METHODS

Estimates of spinal stress need some function to facilitate prediction of disc pressure from vertical accelerations measured at the seat. The methods of Seidel *et al*. [7] and the non-linear approach of Morrison *et al*. [66] may be considered by some too complex.

ISO 2631: 1985 [61] specified a weighting function with unit gain between 4 and 8 Hz. BS 6841 [60] details the *Wb* weighting function for use with vertical seat vibration. This is essentially a filter with band limiting to between 0·4 and 100 Hz. The ISO 2631: 1997 [67] *Wk* filter is similar with only minor variations and most comments regarding *Wb* will apply also to *Wk*. The *Wg* weighting function in BS 6841: 1987 is similar to the ISO 2631: 1985 weighting.

The weighting method has been used extensively for predictions of the magnitude of human responses to vibration. The method owes much to the weighting approach for acoustics, e.g., dB(A). The original concept was based on sinusoidal vibration exposure data and the aim was simply to take into account variations in human response at different sinusoidal frequencies. The approach then developed from octave band analysis of field data to the direct weighting of field accelerations prior to obtaining a single value estimation of effect via r.m.s. (and later *VDV*). This approach is probably viable in acoustics where both health and auditory responses relate to the behaviour of a parallel set of relatively lightly damped biological receptor systems.

However, in the case of whole body vibration and health, one is dealing with mechanical systems that are not wholly in parallel and the prime factor is probably the behaviour of a relatively simple system in the lumbar area. Any method used for converting from seat acceleration to spinal stress should therefore reflect, or at least not contradict, the behaviour of such a system. Even though the weightings were originally developed from sinusoidal vibration exposure, they could be viable for current standards, despite unusual phase behaviour, because current methods of summation are based on power (acceleration squared). However, if peak measuring methods become used then these weighting methods may not be viable.

Payne *et al*. [68] showed that the *Wb* weighting function behaves unusually with single transients. Its response reaches a maximum *before* the input maximum and has a "kick down" when the input pulse finishes. Some of this behaviour may be due to the phase behaviour of the high-pass part of *Wb*.

The apparent mass function allows one to obtain the input forces to the sitting person from measurements of the seat acceleration. Because the area of interest, the lumbar spine, is close to the seat/human interface, these input forces might serve as an estimate of forces in the lumbar spine. Available data suggest that a simple, relatively lightly damped model would be appropriate to simulate the apparent mass. Fairley and Griffin [69] suggested a single-degree-of-freedom model with a natural frequency of 5 Hz and 0·475 critical damping, although their comparison with the results of 60 subjects suggests that less damping would be more appropriate. They also found that the mean resonance frequency decreased from 6 to 4 Hz as the signal magnitude increased from 0·25 to 2 m/s^2 r.m.s. so that one would expect, for the acceleration levels occurring in high acceleration events, that the natural frequency should be less than about 4 Hz. Sandover [70] observed a slightly smaller change with stimulus magnitude with resonance frequencies moving from about 5 to 4 Hz as the stimulus increased from 1 to 3 m/s^2 r.m.s. His data would suggest a damping factor of about 0·25. Cole [71] investigated the apparent mass of subjects exposed to impact. His smallest impacts were of about 60 ms duration with peak values between about 30 and 50 m/s^2—more directly equivalent to the high acceleration events observed here. His data suggest a resonance frequency of between 3 and 5 Hz with damping similar to that of Sandover. Thus, apparent mass data suggest a model natural frequency of about 4 Hz and a damping ratio of about 0·25 if a simple model were used and this has been attempted here. Although it might not seem appropriate to use a single model for a variety of people, both Fairley and Griffin [69] and Sandover [70] showed that, once sitting weight had been taken into account, the apparent mass spectra for a variety of subjects are not widely different.

Concern over the use of the *Wb* weighting for peak measurements prompted a simple comparison of the responses, to the events investigated here, of *Wb*, *Wg*, an apparent mass model and a single degree of freedom model which has similar frequency characteristics to *Wb*. The frequency characteristics of these are illustrated in Figure 4.

The high attenuation of *Wg* and, especially, *Wb* at 2 to 3 Hz, and the similarity in response of *Wg*, *Wb* and the *Wb* simulating model in tracking most sharp transients were clearly evident. However, in some cases where the sharp transients were superimposed on oscillatory acceleration, the high attenuation of *Wg* and *Wb* at low frequencies resulted in lower peak values than one might expect. This did not occur with the simulating model.

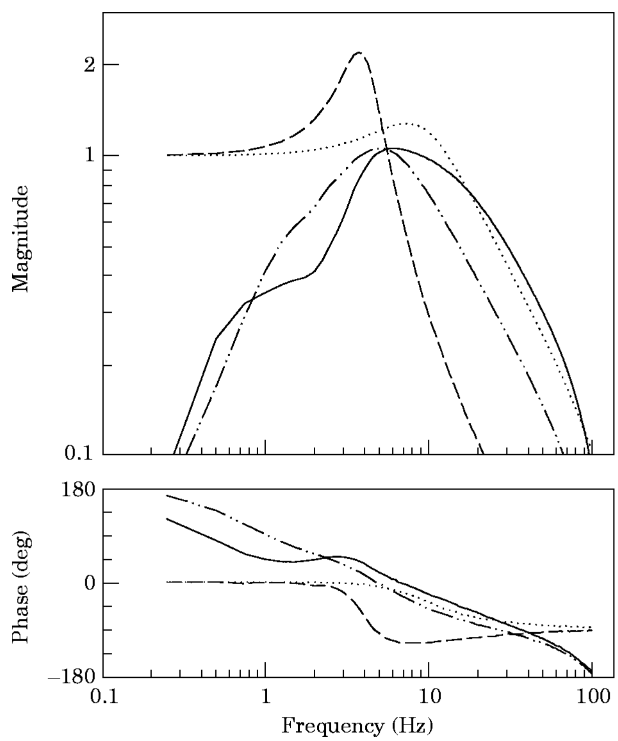

Figure 4. Frequency characteristics of the weightings and models compared as regards response to high acceleration events. ——, BSI *Wb* weighting; . . . . . , *Wb* model; – – –, apparent mass model; — · · — · · —, BSI *Wg* weighting (ISO 2631: 1985).

At times, *Wg* and *Wb* overshot the unweighted response. As one might expect, the lower damped apparent mass model responses appeared as relatively smooth oscillations with little response to the sharp transients.

5.5. RESEARCH NEEDS

Modelling spinal stresses from *in vivo* exposure of subjects in progressing rapidly and becoming more sophisticated. However, the fatigue data for vertebral end plates is confined to two sets of data designed to simulate low frequency activities with low strain rates and where extensive damage occurred. We urgently need fatigue data based on *in vitro* impact accelerations similar to those occurring in practice and using more sensitive failure detection methods.

There is very little useful information on the long term degenerative processes. It is sometimes assumed that tissue repair is a positive process so that, after some time, the effects of fatigue induced damage decrease. However, if callus formation is the main cause of reduced nutrition and degeneration, then repair is not a benign process but leads to cumulative degeneration.

6. CONCLUSIONS

The modelling and fatigue approaches combined offer a basis for estimating the effects of vibration and shock on health. The variables are such that the approach, as yet, only offers rough estimates. However, it gives information and a basis for understanding and, to a certain extent, puts figures to the variables and their effects.

The approach indicates that peak values are important as a large fatigue exponent is probable. Large peaks dominate risk. The method indicates that long term r.m.s. methods probably under-estimate the risk of injury. This may be the reason why epidemiological investigations are not usually able to relate exposure to health effect—exposure data are usually presented as estimates of long term r.m.s. acceleration levels.

The BSI *Wb* and ISO *Wk* weightings have shortcomings when used where peaks values are important. A simple model may be more reliable.

The approach here has been concentrated on high acceleration events. However, the principle also applies to continuous vibration and there is no reason why one method cannot be applied universally to continuous vibrations, high acceleration events and mixtures of these. The fact that an endurance limit can be hypothesized such that, for peak acceleration below a specific value, fatigue failure is unlikely is an important issue in practice. If this limit is sufficiently high, then both complex measurements and consideration of continuous vibration may be unnecessary.

In practice, it is quite unlikely that methods based on hypotheses, laboratory investigations and *in vitro* data will be accepted without question. A possible scenario is that the general approach will be accepted but the actual values used will arise from the application of the approach to epidemiological data and to well known situations.

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