Whole-body vibration biodynamics – a critical review: I. Experimental biodynamics

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Abstract: In the framework of whole-body vibration (WBV), biodynamics refers to biomechanical responses of the human body to impressed oscillatory forces or motions. The biodynamic responses of the human body to WBV form an essential basis for an understanding of mechanical-equivalent properties of the body and potential injury mechanisms, developments in frequency-weightings and design tools of systems coupled with the human operator. In this first part, the biodynamic responses obtained experimentally in terms of 'to-the-body' and 'through-the-body' functions, are critically reviewed and discussed to highlight influences of various contributory factors, such as those related to posture, body support, anthropometry and nature of vibration, together with the range of experimental conditions. The reported data invariably show highly complex, nonlinear and coupled effects of the majority of the contributory factors. It is shown that the reported studies often conclude conflicting effects of many factors, such as posture, gender, vibration and support conditions.

Keywords: apparent mass; STHT; seat-to-head transmissibility; absorbed power; driving point mechanical impedance; contributory factors; gender and anthropometric effect; sitting posture; type; magnitude and frequency of vibration.

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1 Introduction

Vehicles (land, air and sea) expose people to mechanical vibration of periodic, random or transient nature. The exposure to whole-body vibration (WBV) is known to be an important occupational risk factor worldwide. Many studies have suggested that prolonged exposure to intense WBV poses increased risk of disorders of the lumbar spine and nervous system. The long-term effects of WBV have been presented in a number of review papers (Seidel and Heide, 1986; Seidel, 1993; Wikström et al., 1994; Ranganathan and Mohan, 1997; Nakashima et al., 2005; Burström et al., 2015). These studies invariably point to adverse effects of long-term vibration exposure on the spine and spine degeneration, with low back pain (LBP) being secondary symptom. Several epidemiological studies have also established a strong association between occupational WBV exposure and LBP, where the focus has been on vehicle drivers, largest population of workers exposed to WBV (Bongers et al., 1988; Bernard, 1997; Bovenzi and Hulshof, 1998; Magnusson and Pope, 1998; Lings and Leboeuf-Yde, 2000; Seidel, 2005; Tiemessen et al., 2008a; Bovenzi, 2009, 2017; Burström et al., 2017).

In 1996, the Comité Européen de Normalisation (CEN) estimated that 4–7% of all employees in USA, Canada and some European countries are occupationally exposed to potentially harmful WBV (Comité Européen de Normalisation, 1996). Considering economic and productivity growth, it is speculated that the occupationally exposed population is growing not only in developed countries but also in developing nations. Occupationally induced LBP is associated with excessive financial costs, and loss of workdays and decreased quality of life (Kuijer et al., 2015). The total cost of LBP in Sweden was estimated to be in the order of 1860 million euro in 2001, where lost productivity accounted for 84% of the total cost (Ekman et al., 2005). Guo et al. (1999) estimated a total of 101.8 million lost workdays attributable to LBP in 1988 in the USA.

While the association between WBV exposure and LBP is not debated, the definite extent of the association cannot be determined, due to contributions of a multitude of co-varying factors. These include heavy lifting, frequent bending and twisting, and unfavourable postures (National Research Council and Institute of Medicine, 2001; Snook, 2004). The static sitting posture coupled with WBV has been suggested as the causal factor for impairment of the cervical spine (Lawrence, 1955; Magnusson and Pope, 1998). The sitting posture and vibration exposure factors such as vibration magnitude, direction, frequencies and duration may vary substantially with the nature of the task and vehicle type. But epidemiological studies are unlikely to yield information on the dose-response effect and potential injury risks and mechanism attributed to WBV alone because the role of other contributory factors were not adequately considered (Seidel and Heide, 1986; Bovenzi and Hulshof, 1998; Lings and Leboeuf-Yde, 2000).

Experimental biodynamic studies have provided substantial knowledge on movement and mechanical properties of the body, the influences of posture and vibration-related variables, resonance frequencies and probable modes of vibration, potential injury mechanisms and frequency-weighting for exposure assessments (Coermann, 1962; Suggs et al., 1969; Mertens, 1978; Fairley and Griffin, 1989, 1990; Hinz et al., 2002; Wang et al., 2004; Mansfield and Maeda, 2005b; Nawayseh and Griffin, 2005a; Rakheja et al., 2006; Toward and Griffin, 2009; Shibata and Maeda, 2010). ISO-5982 (2001) has defined the range of driving-point mechanical impedance (MI) and seat-to-head transmissibility (STHT) characteristics of the seated body exposed to vertical vibration in the 0.5–20 Hz range on the basis of a synthesis of reported data performed by Boileau et al. (1998). The defined ranges are applicable under particular conditions, namely human subjects sitting erect without a back support but with feet supported and exposed to vertical vibration with magnitudes equal to or less than 5 m/s^2, and a body mass in the 49–93 kg range. The German Institute for Standardization (DIN 45676, 1992) has also defined the ranges of biodynamic responses in terms of driving point MI magnitude and phase for three different body masses (55, 75 and 98 kg). The two standardised values, however, show considerable differences. The reported studies on biodynamics have placed a far greater emphasis on the responses to vertical vibration, while far fewer efforts have been made under horizontal vibration, whose magnitudes may be comparable to those of the vertical in many off-road vehicles (Rakheja et al., 2008). Exposure to horizontal forces could induce greater shear stresses in the spine.

In this first part of the paper, the reported studies on experimental biodynamic responses to WBV and biodynamic models of the human body are critically reviewed and discussed to highlight the roles of contributory factors. A review of reported biodynamic models is presented in the second part together with relationships among the different biodynamic measures, and the need for further research.

2 Biodynamic measures and measurement methods

The biodynamic responses of the human body exposed to WBV are expressed by two broad functions:

1. 'to-the-body' response function describing the force-motion relation at the point of entry of vibration or the driving-point, namely, mechanical impedance (MI), apparent mass (AM) and absorbed power
2. 'through-the-body' response function that describes the flow of vibration through the body, such as STHT, foot-to-head transmissibility (FTHT) and body segments vibration transmissibility.

These have been widely used to identify resonance frequencies of the body, so as to quantify the critical frequency ranges of vibration under which greater deflections and thus stresses of the biological system may occur. The frequencies corresponding to peak magnitude response, whether AM, MI, STHT or FTHT, have been commonly referred to as the resonance frequencies of the body.

2.1 'To-the-body' response function

The MI and AM relate the dynamic force developed at the driving-point between the vibrating surface and the body with the velocity and acceleration at the interface, respectively, such that:

$$Z(j\omega) = \frac{F(j\omega)}{v(j\omega)} \text{ and } M(j\omega) = \frac{F(j\omega)}{a(j\omega)} = \frac{F(j\omega)}{v(j\omega)\cdot j\omega} = Z(j\omega)/j\omega \tag{1}$$

where $Z(j\omega)$ and $M(j\omega)$ are complex MI and AM, respectively, corresponding to excitation frequency ω, $F(j\omega)$ is the dynamic force developed at the driving-point, and $v(j\omega)$ and $a(j\omega)$ are the velocity and acceleration, respectively, due to source vibration.

Under random vibration, these measures are generally evaluated using the one-sided power spectral density (PSD) functions, such that:

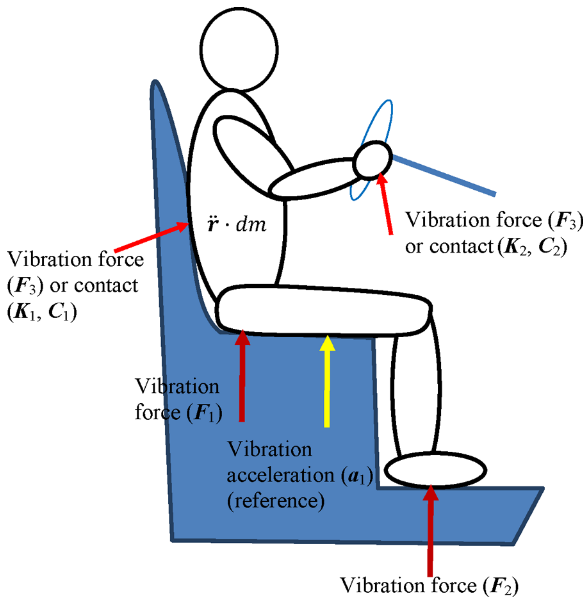
$$Z(j\omega) = \frac{S_{vF}(j\omega)}{S_v(j\omega)} \text{ and } M(j\omega) = \frac{S_{aF}(j\omega)}{S_a(j\omega)} \tag{2}$$

where S_{vF} and S_{aF} are cross spectral densities (CSD) of force and velocity, and force and acceleration, respectively, S_v and S_a are the auto spectral densities of velocity and acceleration, respectively.

The 'to-the-body' functions have also been evaluated from the ratios of auto spectral densities, $M(\omega) = S_F(\omega)/S_a(\omega)$, and root-mean-square (rms) values, $M(\omega) = F_{rms}(\omega)/a_{rms}(\omega)$, where F_{rms} and a_{rms} are rms values of the dynamic force and acceleration corresponding to circular frequency ω (Hinz and Seidel, 1987; Matsumoto and Griffin, 2005). These relations, however, cannot provide reliable AM phase information, while the magnitudes may be comparable to those from equation (2).

Dong et al. (2013) recently introduced an alternate driving-point response function in the formulation of a new vibration theorem, which is termed cross-point apparent mass or cross-point mechanical impedance. Different from the direct apparent mass or mechanical impedance in a direction expressed in Eq.(1), the cross-point response function is calculated using the vibration acceleration or velocity at a selected or reference driving point and the vibration force at a different driving point but in the same direction as that of the reference input vibration direction. For example, while the direct apparent mass at driving point 1, shown in Figure 1, is equal to $\boldsymbol{F}_1/\boldsymbol{a}_1$, the cross-point apparent mass at driving point 2 is equal to $\boldsymbol{F}_2/\boldsymbol{a}_1$, in which $\boldsymbol{a}_1$ is the reference vibration acceleration.

Figure 1 A conceptual model of whole-body response to input vibration (see online version for colours)



A few studies have also reported cross-axis AM responses, to study coupling effects (Nawayseh and Griffin, 2003; Mandapuram et al., 2011). For instance, a seated or standing body exposed to vertical vibration will exhibit motion and biodynamic forces not only in the vertical but also in the horizontal direction. Exposure to horizontal vibration, in a similar manner, causes vertical as well as rotational motions of the body.

Under single- or multiple-axes vibration, the cross-axis AM is defined as the ratio of biodynamic force measured along the non-vibration axis to acceleration:

$$M_q(j\omega) = \frac{S_{a_pF_q}(j\omega)}{S_{a_p}(j\omega)}; \quad p \neq q$$

where $M_q(j\omega)$ is the cross-axis AM along q-axis under vibration along the axis p, F_q is force along the cross-axis q and is a_p is the acceleration due to vibration along axis p.

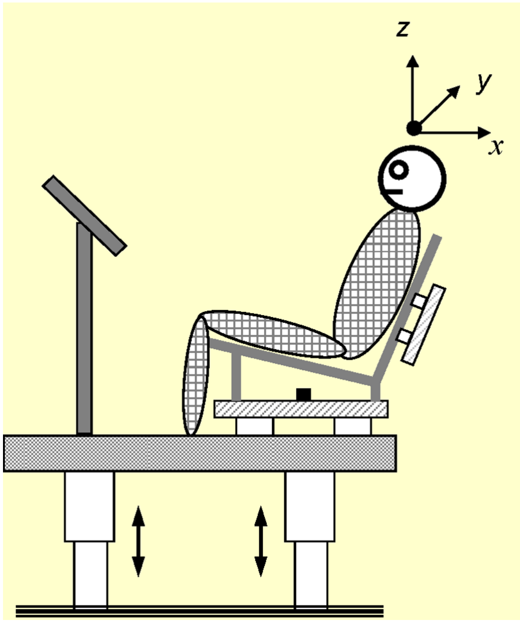
The direct and cross-axis biodynamic responses have been generally evaluated using the H_1 frequency response function estimator. Mandapuram et al. (2011) used *Hv* estimator and concluded that a better understanding of the seated human body response to uncorrelated multi-axis WBV could be developed using the *Hv* estimator.

Biodynamic responses have generally been measured in the laboratory under controlled vibration and postural conditions. The human subject is positioned on a rigid platform or a rigid seat assuming the desired posture, while the total dynamic force is measured beneath the platform under controlled vibration. The platform or seat structure is designed such that its natural frequency is well above the range of vibration frequency to ensure negligible contribution of the structure modes, and to characterise the decoupled responses of the body alone. Figure 2 illustrates a typical measurement setup for the seated body exposed to vertical WBV. The inertia force due to the mass of the platform or seat structure is subtracted in order to derive the biodynamic force developed by the body alone at the driving-point. The degree of correlation between the vibration signal and the measured force is also monitored in the experiments via the coherence function, γ:

$$\gamma^2(\omega) = \frac{\left|S_{aF}(j\omega)\right|^2}{S_a(j\omega)S_F(j\omega)} \tag{3}$$

While earlier studies generally reported the response in terms of MI (Coermann, 1962; Edwards and Lange, 1964; Vogt et al., 1968; Vykukal, 1968; Suggs et al., 1969; Miwa, 1975), the later studies have mostly used AM due to its ease of measurement and analysis. The AM magnitude at very low frequencies resembles the body mass supported by the platform. Unlike the MI, the AM responses exhibit wide variability in low frequency magnitudes attributed to variations in the body mass. The AM magnitude is thus frequently normalised with respect to the magnitude at a low frequency, which ranges from 0.5 to 2 Hz and is considered to represent the static mass on the vibrating platform. Holmlund et al. (2000) normalised the MI magnitude responses of seated subjects exposed to vertical vibration with respect to that at 2 Hz. Subashi et al. (2009) normalised the AM response of seated subjects exposed to fore-aft and lateral vibration with respect to magnitude at 4 Hz. These normalisation factors may not correspond to the static mass supported by the seat. Unlike vertical vibration, where the body behaves similarly to a rigid mass at very low frequencies (below 2 Hz), the seated body exhibits resonances near 0.7 Hz under horizontal vibration (Fairley and Griffin, 1990; Mandapuram et al., 2005). It is thus quite difficult to identify static mass supported by the platform from the measured AM or MI responses. Consequently, a few studies have normalised the measured horizontal biodynamic responses with respect to the standing body mass (Holmlund and Lundström, 1998; Mansfield and Lundström, 1999).

Figure 2 A typical laboratory set up for measurement of 'to-the-body' and 'through-the-body' biodynamic response functions of the seated body under vertical WBV (see online version for colours)



Alternatively, a number of studies have performed normalisation with respect to the static mass on the vibrating platform (Mansfield and Griffin, 2000; Wang et al., 2004; Mansfield et al., 2006; Hinz et al., 2006; Shibata and Maeda, 2010; Mandapuram et al., 2011; Toward and Griffin, 2011). For the seated posture, static mass depends greatly on the seat geometry, back support, thigh contact, sitting posture, etc. For a no back support upright sitting condition, Lundström et al. (1998) reported that 77% and 76% of the total body mass of female and male subjects rests on the flat seat pan. Wang et al. (2004) reported that increasing the floor to pan height from 410 to 510 mm increases the body mass on the pan from 73.4% to 85% for the no back support and from 77% to 88.7% for an inclined back support. This is attributed to greater thigh contact with the pan of a higher seat. For an automotive seat geometry (inclined pan and backrest), Rakheja et al. (2002) reported that 76.6% and 73.3% of the total body mass is supported by the pan, while sitting with hands in lap and hands on a steering wheel (SW); the corresponding mean proportions of the total mass supported by the inclined back support were measured as 30.4% and 28.1%. The normalisation factor thus needs to be determined for the particular seat geometry and sitting posture.

The AM or MI functions characterise the biodynamic response or properties of the human body exposed to vibration, but cannot be applied for quantifying the vibration exposure (intensity and exposure duration). The acceleration due to source vibration, on the other hand, is considered to represent the vibration hazard. An alternate measure, the vibration power absorbed (VPA) by the exposed body, that combines both the vibration hazard and the biodynamic properties has been proposed to assess the effects of WBV. The measure is derived from the dynamic force and velocity at the driving point, and relates to dissipation of energy attributed to the relative motion of the visco-elastic tissues, muscles and skeletal system, which under prolonged exposure could lead to physical damage in the musculoskeletal system (Lee and Pradko, 1968). Mathematically, the absorbed power can be computed from the vibration induced stress and strain rate, which constitute the essential mechanical stimuli leading to biological responses and adaptation (Anderson and Boughflower, 1978; Dong et al., 2005). Moreover, unlike the AM or MI, the absorbed power can be used to estimate cumulative energy dissipated by the exposed body over a given duration and can thus facilitate assessment of effects of exposure duration apart from the intensity of vibration.

The average vibration power P_{avg} transferred to the exposed human body over the duration T has been derived using a direct method and an indirect method based upon the AM or MI (Wang et al., 2006a). In the direct method, P_{avg} is evaluated from:

$$P_{\text{avg}} = \frac{1}{T}\int_0^T v(t)F(t)\,\mathrm{d}t \tag{4}$$

The indirect method of analysis requires the transformation to the frequency domain. The relationships between P_{avg} and MI/AM responses have been derived as (Wang et al., 2006a):

$$P_{\text{avg}} = \infty\int_0 \mathrm{Re}\left[Z(j\omega)\right]S_v(\omega)\,\mathrm{d}\omega; \quad P_{avg} = \int_0^\infty \frac{\mathrm{Im}\left[M^*(j\omega)\right]}{\omega}S_a(\omega)\,\mathrm{d}\omega \tag{5}$$

where Re and Im are real and imaginary components, respectively, and $M^*(j\omega)$ is conjugate of the complex AM.

Different forms of normalised power have also been reported, such as power normalised by the body mass (W/kg), power density normalised by acceleration PSD, and that by the product of acceleration spectral density and the body mass. It has been suggested that the normalisation with respect to acceleration spectrum helps to smoothen the small magnitude oscillations in the power response (Mansfield and Griffin, 1998).

2.2 'Through-the-body' response function

The 'through-the-body' response function is used to characterise transmission of vibration to a particular location of the body, and is defined as the ratio of acceleration due to transmitted vibration to that of the source vibration:

$$T_i(j\omega) = \frac{a_{Ti}(j\omega)}{a(j\omega)} \tag{6}$$

where $T_i(j\omega)$ is the 'through-the-body' transfer function or vibration transmissibility of the location along direction i and $a_{Ti}(j\omega)$ is the acceleration response measured at a particular location on the body along direction i ($i = x, y, z$). Since the reported studies are generally performed under vibration along a single axis, the direction subscript is not assigned to the input acceleration $a(j\omega)$ measured at the vibrating platform. More appropriately, the transfer function is derived from the CSD of the output and input vibration, using equation (2), such that:

$$T_i(j\omega) = \frac{S_{aa_{Ti}}(j\omega)}{S_a(j\omega)} \tag{7}$$

where $S_{aa_{Ti}}$ is the CSD of accelerations due to transmitted and source vibration. The 'through-the-body' biodynamic responses are also expressed by vibration transmitted along two or more axes, even though the input vibration occurs along a single axis. Therefore, similar to the cross-axis AM, the cross-axis vibration transmissibility may be expressed as:

$$T_q(j\omega) = \frac{S_{a_q a_p}(j\omega)}{S_{a_p}(j\omega)} \tag{8}$$

where $T_q(j\omega)$ is the direct-axis vibration transmissibility of a body segment, when vibration is applied along the q-axis ($p = q$), and cross-axis transmissibility, when $p \neq q$.

The 'through-the-body' response can yield a better understanding of potential adverse effects of WBV. However, relatively fewer studies have measured the vibration transmissibility using skin-mounted miniature accelerometers to different locations of the vertebrae (L1, L3, L5, T1, T5, T6, T10, T11, T12, C7) and pelvis (Hinz and Seidel, 1987; Hinz et al., 1988; Zimmermann and Cook, 1997; Kitazaki and Griffin, 1998; Matsumoto and Griffin, 1998a, 2000; Pranesh et al., 2010). The relative movement of the skin and the tissue over a bone, together with the mass of the accelerometer and the type of mounting, however, can alter the acceleration responses (Pope et al., 1986). The accelerations measured at the skin surface are thus frequently corrected for the contribution of skin movement. Kitazaki and Griffin (1995) proposed a correction method by considering localised skin and the accelerometer as a single-degree-of-freedom (DOF) dynamic system, which permitted estimation of response at the spinous process from skin surface acceleration. The reported data on vibration transmitted to various locations of the lumbar and thoracic spine, however, show extremely large differences in both the peak values and the corresponding frequencies (Panjabi et al., 1986; Pope et al., 1986; Hinz and Seidel, 1987; Sandover and Dupuis, 1987; Magnusson et al., 1993; Matsumoto and Griffin, 1998a; Mansfield and Griffin, 2000; Pranesh et al., 2010). Such variabilities are partly attributed to wide differences in measurement systems, methods employed in the studies, and contribution due to skin movement.

The vibration transmitted to the head of the standing or seated body has been measured using different methods, namely bite bar, scalp- or helmet-mounted accelerometers. Paddan and Griffin (1988a, 1988b, 1993) developed a six-axis bite-bar to measure vibration of the head along all the translational and rotational axes, which relies upon the teeth to grip a rigid bite-bar and poses considerable complexity in precise alignment of the bite bar with the chosen coordinate system. The authors also proposed the use of a dental mould to reduce discomfort sensation of the test subjects. Woodman (1995, 1996) and Smith (2000) used helmet-mounted accelerometers to measure STHT responses. The relatively larger mass and mass moment of inertia of the helmet together with its relative motion with respect to the head can alter the head vibration and contribute to greater variability. Wang et al. (2006b) developed a light-weight head strap with a tension adjuster to measure the head vibration along the three translational axes.

2.3 Relationships between biodynamic functions and psychophysical responses to vibration

While it is very difficult to identify the frequency dependence of a vibration health effect and to establish a reliable dose-response relationship for risk assessment of vibration exposure, the vibration psychophysical responses have been used as an important basis to establish the standard for measurement and risk assessment of the human vibration exposure (ISO-2631-1, 1997). A few studies have experimentally examined correlations between the subjective judgements of WBV and 'to-the-body' response magnitudes with somewhat contradictory findings. Matsumoto and Griffin (2005) investigated the objective biodynamic responses in terms of AM and MI. A poor correlation of subjective estimations was observed with AM magnitude normalised with respect to the magnitude at 5 Hz under continuous sinusoidal vibration, while a better correlation was obtained with normalised MI; and an opposite trend was noted under transient excitations. Mansfield et al. (2000) also explored correlations between the subjective sensations of vertical vibration with VPA, vibration dose value (VDV), and W_b- and W_k-weighted accelerations (as defined in BS-6841 (1987) and ISO-2631-1 (1997), respectively), apart from the other objective measures. The study concluded the greatest correlation of subjective responses and the VPA, compared to those with the range of objective measures considered.

3 Characterisation of whole-body vibration

At the workplace, a human operator generally assumes two forms of postures: standing (ship workers and operators of high speed crafts) and sitting (vehicle drivers). The WBV environment is invariably characterised in terms of acceleration due to ease of its measurement and its direct relevance with force or stress and human sensation to vibration. The assessment methods are also based on the frequency-weighted acceleration (ISO-2631-1, 1997; BS-6841, 1987). Table 1 summarises the ranges of frequency-weighted magnitudes of WBV of different vehicles along the three translational axes (a_{wx}, a_{wy} and a_{wz}), where the source of data is indicated by the lead author. Although a large number of off-road and industrial vehicles impose comprehensive magnitudes of vibration along the roll and pitch axes (Boileau and Rakheja, 2000), only minimal data seem to exist. The studies generally report the mean or range of WBV magnitudes measured at the driver-seat or cabin floor for generic vehicle types (such as tractor, earthmoving machinery, forklift truck, etc.), while vehicle size, power, wheel base, types and condition of tyres, and the nature of task and speed are not always elaborated; all of which greatly affect the nature of the WBV. A number of studies, however, have shown that the exposure levels are greatly dependent upon the tasks performed and terrain conditions (Village et al., 1989; Boileau et al., 2002; Kumar, 2004; Rehn et al., 2005; Newell et al., 2006) and the vehicle size (Maeda and Morioka, 1998; Boileau and Rakheja, 2000).

The results summarised in Table 1 clearly show that the magnitudes of fore-aft (x) and lateral (y) vibration at the driver's seat of off-road tractors, highway trucks, loaders, dumpers, forestry machines, excavators, snowmobiles, helicopters and port cranes could be either comparable or exceed the magnitudes of vertical vibration. Exposure to such large magnitudes of horizontal vibration could cause greater shear forces in the lumbar spine (Fritz, 2005). The exposure may thus impose relatively higher risks due to low shear strength of the lumbar. However, the biodynamic responses of the standing and seated body to horizontal vibration have been reported in far fewer studies than those under vertical vibration.

Unlike the intense variability in the magnitudes of vertical and horizontal vibration, relatively smaller variations occur at the predominant frequencies of vibration of most vehicles. The ranges of frequencies of dominant vibration of most wheeled road- and off-road vehicles are generally up to 20 and 10 Hz, respectively, in the vertical and horizontal directions, while those of tracked vehicles tend to dominate up to 30 Hz. Considering that the magnitudes of biodynamic responses, particularly, the 'to-the-body' function diminish at higher frequencies, it would perhaps be appropriate to characterise the responses up to 20 Hz and 10 Hz under vertical and horizontal WBV, respectively. Owing to the strong dependence of the biodynamic responses on the magnitudes of vibration and wide variations in the WBV levels at the workplace, it would be extremely complex to define and synthesise representative WBV spectra in the laboratory for characterising the biodynamic responses. Consequently, the vast majority have reported the responses to broad-band random or sinusoidal vibration.

Table 1 Ranges of reported WBV levels of different vehicles

	Weighted acceleration (m/s^2)			
Vehicle	a_{wx}	a_{wy}	a_{wz}	*Source(s)*
Forklift trucks				Bovenzi et al. (2002, 2006), Okunribido et al. (2006), Boileau and Rakheja (2000), Costa and Arezes (2009)
Diesel	0.40–0.49	0.06–0.38	0.12–1.20	
Electric	0.40	0.56–0.59	1.52–1.63	
City busses	0.03–0.45	0.05–0.47	0.10–1.01	Bovenzi and Zadani (1992), Bovenzi and Hulshof (1998), Bovenzi et al. (2006), Okunribido et al. (2006, 2007), Blood et al. (2010)
Taxis	0.13–0.23	0.13–0.18	0.30–0.34	Funakoshi et al. (2004)
Motorcycle	0.15–0.44	0.11–0.17	0.30–0.93	Chen et al. (2009)
Highway trucks and combinations	0.16–0.83	0.18–0.87	0.02–1.1	Cann et al. (2004), Tiemessen et al. (2008b), Bovenzi et al. (2006), Okunribido et al. (2006), Smets et al. (2010)
Mining trucks	0.13–0.57	0.14–0.48	0.40–1.52	Kumar (2004)
Pick-up trucks (2 and 4 wheels drive)	0.17–0.48	0.23–0.31	0.46–1.08	Salmoni et al. (2008)
Agricultural tractors (various tasks)	0.07–1.12	0.11–1.40	0.16–1.35	Scarlett et al. (2007), Servadio et al. (2007), Okunribido et al. (2006), Futatsuka et al. (1998), Kumar et al. (2001), Marsili et al. (2002)
Underground mining dumpers – loaded and unloaded	0.63–1.50	0.54–0.84	0.87–2.50	Village et al. (1989)
Cranes – mobile and overhead	0.07–0.66	0.11–0.67	0.22–0.52	Tiemessen et al. (2008b), Bovenzi et al. (2002, 2006)
Garbage trucks	0.08–1.49	0.12–1.98	0.21–2.45	Maeda and Morioko (1998), Bovenzi et al. (2006)

Table 1 Ranges of reported WBV levels of different vehicles (continued)

Vehicle	*Weighted acceleration* (m/s^2) a_{wx}	a_{wy}	a_{wz}	*Source(s)*
Construction machines				Newell et al. (2006), Tiemessen et al. (2008b), Okunribido et al. (2006), Bovenzi et al. (2006), Eger et al. (2008)
Loaders-wheeled	0.21–1.40	0.22–1.30	0.29–1.26	
Loaders-tracked	0.65–1.12	0.34–0.76	0.51–0.96	
Dumpers	0.51–1.12	0.30–0.78	0.46–1.18	
Excavators	0.24–0.52	0.20–0.26	0.30–0.52	
Forestry machines				
Skidders	0.54–0.86	0.49–1.42	0.72–1.15	Cation et al. (2008), Golsse (1989), Golsse and Hope (1987)
Forwarder	0.64–0.75	0.80–1.52	0.39–0.68	Mansfield et al. (2002), Sherwin et al. (2004)
Snow vehicles				Boileau et al. (2002)
Side–walk ploughs	0.35–1.03	0.20–0.86	0.81–2.23	
Snowmobiles	0.43–1.00	0.50–1.00	0.30–1.00	Rehn et al. (2005)
Groomers	0.15–0.36	0.15–0.36	0.40–1.10	
Armoured vehicles				
• wheeled	0.07–0.21	0.04–0.20	0.26–0.66	Nakashima et al. (2005)
• tracked	0.12–0.35	0.13–0.57	0.57–0.89	
Aircraft – landing	0.20–0.30	0.10–0.40	0.60–0.90	Burström et al. (2006)
High speed crafts	0.35–2.15	0.22–1.41	1.12–4.31	Rakheja and Boileau (2006)
Helicopter	0.54	0.54	0.44	Okunribido et al. (2006)

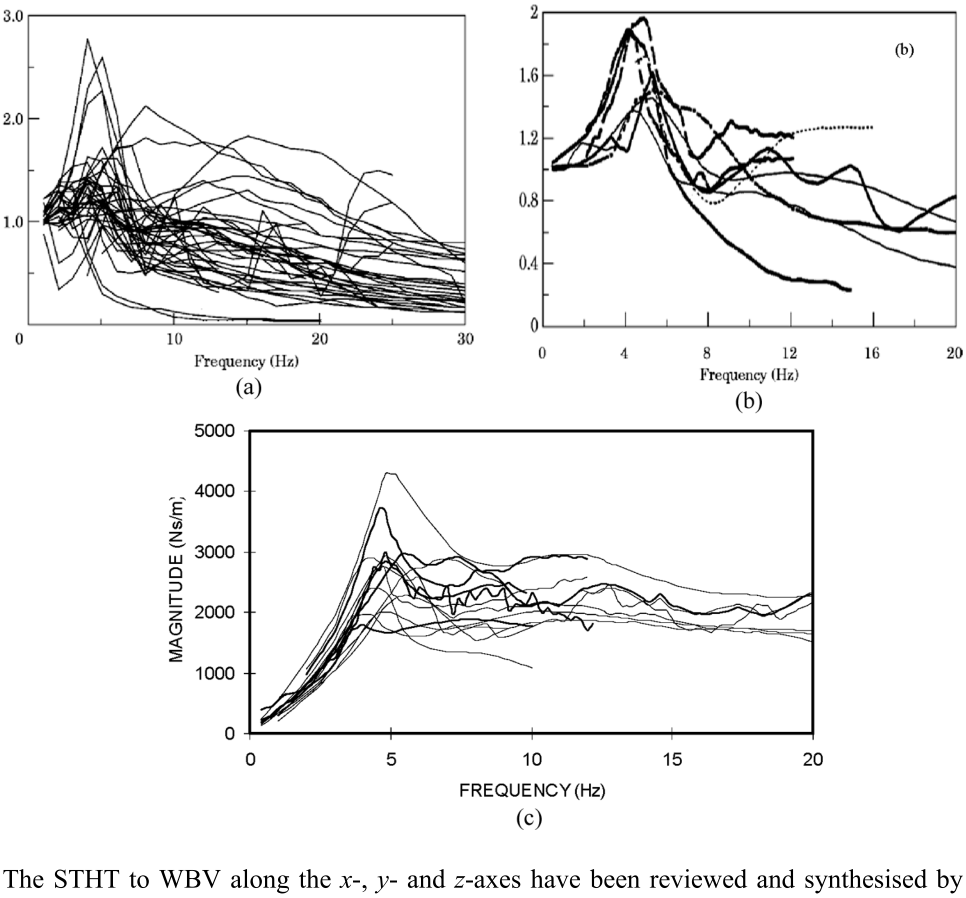
4 Biodynamic responses of the seated body to WBV

The majority of studies have considered a body seated on a rigid seat platform in order to obtain the decoupled biodynamic responses to WBV. A number of studies have also measured the responses of the body seated on cushioned seats with a variety of harnesses, which are known to alter the biodynamic properties substantially (Fairley and Griffin, 1986; Nishiyama et al., 2000; Fleury and Mistrot, 2006; Stein et al., 2007, 2008, 2009; Toward and Griffin, 2009; Dewangan et al., 2013a, 2015). Despite the comprehensive magnitudes of WBV along the *x*- and *y*-axes (Table 1), the 'to-the-body' responses to such vibration have been reported in very few studies, while only minimal efforts have been made with respect to 'through-the-body' responses. Lewis and Griffin (1980) measured vertical and pitch head vibration under *x*- and *z*-axes seat vibration, and lateral and roll head vibration under *y*-axis seat vibration. The study concluded that very little horizontal vibration is transmitted to the head. Paddan and Griffin (1988b), on the other hand, observed higher head vibration along the fore-aft, vertical and pitch axis under *x*-axis seat vibration, when seated with a back support, compared to those with an unsupported back. A few studies have also reported 'to-the-body responses' under simultaneously applied dual (Mandapuram et al., 2010; Hinz et al., 2006b; Mansfield and

Maeda, 2006, 2007) and three-axis vibration (Hinz et al., 2006, 2010; Mansfield and Maeda, 2006, 2007).

Wang et al. (2006b) reported coefficients of variation (COV) in excess of 30% in the cross-axis fore-aft head vibration of the body seated without a back support under vertical vibration. Mandapuram et al. (2010) reported peak COV values in 21–40% and 22–75% ranges in seat pan and backrest AM magnitudes. The MI data under horizontal vibration, reported by Holmlund and Lundström (1998), revealed COV on the order of 30% in magnitude and in excess of 100% in phase. The inter-subject variability decreased when a back support was used. Greater variability between responses of the subjects was observed by Pranesh et al. (2010) around the primary resonant frequency (4–7 Hz) for both the vertical and horizontal vibration transmissibility magnitudes at most locations in the trunk and head. Additionally, the dispersions in this frequency range were generally higher for responses in the horizontal axis.

Figure 3 Variations among the reported biodynamic responses: (a) STHT magnitude (Paddan and Griffin, 1998); (b) STHT magnitude (Boileau et al., 1998) and (c) MI magnitude (Boileau et al., 1998)



The STHT to WBV along the *x*-, *y*- and *z*-axes have been reviewed and synthesised by Paddan and Griffin (1998). Rakheja et al. (2010) performed the synthesis of the data, which included AM of the body seated with and without a back support while exposed to *x*-, *y*- and *z*-axes and those of the standing body to *z*-axis, and STHT of the seated body.

The comparisons of the reported results revealed excessive discrepancies among them, which is partly attributed to inclusion of data acquired under a wide range of experimental conditions, namely: subjects seated on rigid, and cushioned seats; subjects seated in car or ejection seats with and without harnesses and belts; and those reporting the transmissibility data at a few discrete frequencies or under vibration within a limited low or high frequency range. The ranges of experimental conditions used in studies reporting seated body biodynamic responses have been summarised by Rakheja et al. (2010). As an example, Figure 3 illustrates variations among the reported vertical STHT and MI magnitude responses of the seated body exposed to vertical WBV (Paddan and Griffin, 1998; Boileau et al., 1998). Wide differences in experimental conditions employed in different studies have been illustrated in a data synthesis study by Rakheja et al. (2010). Consequently, considerable differences among the reported properties would be expected.

The reported studies on experimental biodynamics generally focus on the mechanical properties, namely, the resonance frequencies and damping as determined from magnitude responses over the selected frequency range. Both, the 'to-the-body' and 'through-the-body' responses to vertical vibration exhibit a consistent principal resonance in the 4–7 Hz range, often referred to as the whole-body vertical mode. The studies reporting vibration transmitted to the spine and pelvis (Panjabi et al., 1986; Magnusson et al., 1993; Zimmermann and Cook, 1997; Matsumoto and Griffin, 2000; Mansfield and Griffin, 2000; Pranesh et al., 2010) and abdominal pressure (White et al., 1962; Sandover, 1978) also show this principal mode in the 4–6 Hz range. A number of measured data have also shown the presence of relatively less clear secondary vibration modes in the 8–15 Hz range, which was not always evident for all the subjects (Coermann, 1962; Vogt et al., 1968; Vykukal, 1968; Fairley and Griffin, 1989; Mansfield et al., 2001; Rakheja et al., 2002; Wang et al., 2004). Peaks in the cross-axis STHT responses to vertical vibration near 2 Hz have also been identified and attributed to pitch motion of the upper body (Hinz et al., 2002; Wang et al., 2006b).

A few researchers have attempted to identify the precise deflection mode associated with the principal frequency, although little consensus exists. Hagena et al. (1985) hypothesised that the primary and secondary resonances are associated with vertical motions of the entire body and the spinal column, respectively. Sandover and Dupuis (1987) associated the principal frequency with bending mode of the lumbar spine caused by pelvis pitch. Coermann (1962) reported peak relative motions of the pelvis near 5 Hz and 9 Hz, while Mansfield and Griffin (2000) and Matsumoto and Griffin (1998a) noted peak seat-to-pelvis transmissibility near 4 Hz and in the 7–10 Hz range, respectively. Zimmermann and Cook (1997) reported peak pelvic motion in the 4.5–6 Hz range, depending upon the mean or static pelvic orientation. Hinz et al. (1988) suggested flexion and extension mode of the spine accompanied by vertical motion of the entire body near 4.5 Hz. It was further suggested that movement of the upper body (above L3–L4) was primarily responsible for the lumbar spine bending, while the pelvis rotation may be the secondary effect. Kitazaki and Griffin (1998) extracted modal properties of the seated body and identified several modes below 10 Hz. For the upright normal posture, the principal whole-body mode at 4.9 Hz was associated with vertical motions of the head, spinal column and pelvis due to axial and shear deformations of the buttock tissue, in-phase viscera and bending modes of the upper thoracic and cervical spine. The lumbar and lower thoracic spine bending mode coupled with vertical motion of the head was identified at 5.6 Hz. The study did not report pelvic rotation corresponding to both the modes, which was reported by Hagena et al. (1985), Sandover and Dupuis (1987), Zimmermann and Cook (1997) and Hinz et al. (1988). Kitazaki and Griffin (1998) also identified the pelvic pitching modes at 8.1 and 8.7 Hz, and the second visceral mode at 9.3 Hz. It was suggested that the modes at 8.1, 8.7 and 9.3 Hz correspond to secondary resonances observed in the AM responses. Mansfield and Griffin (2000) observed fore-aft and vertical resonances of the viscera around 6 Hz, which was not attributed to the principal resonance in AM due to small mass of the viscera. The spine and pelvis vibration transmissibility peaked near 4 Hz, while a second larger magnitude peak was observed in the 8–10 Hz range. The deflection modes and the corresponding frequencies in both the studies were observed to vary with changes in the sitting posture.

The resonance frequencies of the body in the horizontal directions have also been identified from the measured biodynamic responses, although greater discrepancies seem to exist. From the AM responses to fore-aft and lateral vibration, Fairley and Griffin (1990) observed primary resonance frequencies near 0.7 Hz in both directions, when seated without a back support, with secondary modes near 2.5 and 2 Hz along the x- and y-axes, respectively. Mandapuram et al. (2005, 2010) also observed the same primary resonance, while the secondary modes in x-axis occurred at 2.8 and 4.75 Hz, and near 2 Hz under y-axis. Stein et al. (2009) observed primary resonance near 0.75 Hz, and a secondary peak near 2.75 Hz under y-axis vibration. Lee and Pradko (1968) identified these frequencies as 1.3 Hz under x-axis, and 0.6 and 1.8 Hz in the y-axis. Holmlund and Lundström (1998), on the other hand, observed MI magnitude peaks in the 3–5 Hz range under x-axis vibration for the erect sitting posture, and near 3 and 6–7 Hz range for the relaxed posture. The peaks in the y-axis MI magnitude occurred at 2 and 6 Hz for both sitting postures. Measurements in this study, however, were conducted under vibration in the 1.13–80 Hz range, and consequently the low frequency vibration modes could not observed. Mansfield and Maeda (2007) measured the AM under individual and multiple axis vibration in the 1–20 Hz range, and reported median resonance frequencies of less than 1.0 and near 1.75 Hz corresponding to x- and y-axes vibration, respectively. The first mode was attributed to pitching and swaying of the upper body under x- and y-axes vibration, while the secondary modes were believed to be associated with horizontal motions of the musculoskeletal structure. In contrast to the above results, Rahmatalla and DeShaw (2011) observed resonance of the head near 7 Hz under fore-aft vibration.

The above studies have consistently shown that addition of a backrest causes the fore-aft and lateral modes to mostly converge to a single mode, with resonance frequencies being 3.5 and 1.5 Hz under x- and y-axes, respectively (Fairley and Griffin, 1990), 2.7–5.4 Hz and 0.9–2.1 Hz (Mandapuram et al., 2005), and 4 and 2 Hz (Mansfield and Maeda, 2007). However, Mandapuram et al. (2010) observed two peaks at 1.25 and 4.5 Hz under x-axis, and 0.88 and 2.25 Hz under y-axis vibration.

The STHT responses to horizontal vibration revealed peaks around frequencies that are considerably different from those observed from the AM responses. The STHT magnitude peaks near 3 and 1.5 Hz under both x-and y-axes vibration, when sitting without a back support, were reported by Paddan and Griffin (1988b). The responses measured with a back support showed an additional peak in the fore-aft STHT near 8 Hz, with the primary peak shifting towards 2 Hz, while the effect of back support on the frequency of lateral STHT was minimal. Hinz et al. (2010) found primary resonance at 1 Hz under x- and y-axes vibration, while sitting without a back support. Three resonances were evident at the back–backrest interfaces of seated subjects exposed to fore-and-aft vibration (Abdul Jalil and Griffin, 2008). First resonance was around 2 Hz, while the secondary resonance occurred between 4 and 5 Hz or 5 and 8 Hz depending on the back support orientation. A third resonance was observed around 7 Hz, although only for some of the subjects.

The reported data suggests even greater differences in the magnitudes of ‘to-the-body’ and ‘through-the-body’ response magnitudes. These differences are attributable to nonlinear dependence of the biodynamic responses on various factors, namely, posture, muscle tension and nature of vibration (frequency, magnitude and direction). These factors may be grouped under subject anthropometry, sitting posture, nature of WBV and support conditions, as summarised in Table 2. The data reported in various studies are reviewed and discussed in the following sections to gain insight into the effects of particular factors. This however is quite complex in many instances, where the coupled effects of a number of factors exist.

Table 2 Grouping of factor affecting the biodynamic responses of the seated body exposed to WBV

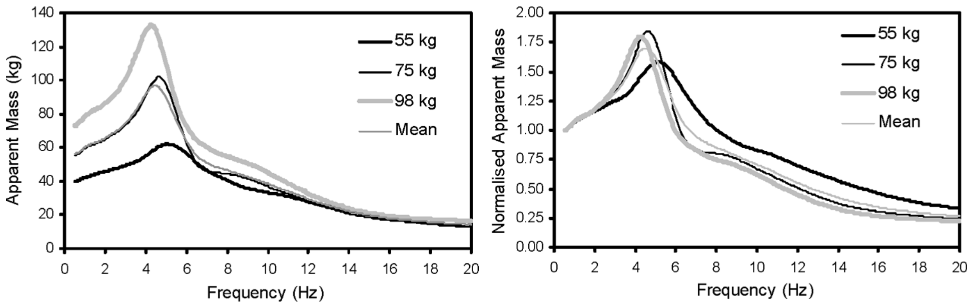
Anthropometry	*Sitting posture*	*Vibration*	*Support*
Body mass	Sitting erect	Type (sine, random, shock)	Back support
Body fat	Sitting slouched	Direction	Back rest orientation
Height	Muscle tension	Intensity	Pan orientation
Contact area on vibrating surface	Feet support	Frequency	Seat height
Gender	Hands support		
	Thigh support		
	Twisted upper body		
	Pelvic orientation		

4.1 Effect of body mass

The ‘to-the-body’ biodynamic responses of the seated body are greatly influenced by the body mass. A larger body mass causes greater contact area and more uniform contact force between the thighs and the seat pan, which could considerably alter the ‘to-the-body’ biodynamic responses (Nawayseh and Griffin, 2005a). Mertens (1978) suggested that variations in body mass and age yield only small variations in MI responses of subjects exposed to vertical vibration under increasing gravity in a centrifuge. The vast majority, however, have shown large scatter in the AM magnitude response, particularly in the low frequency range, attributed to body mass variations.

Fairley and Griffin (1989) measured AM responses of 60 seated subjects, including men, women and children, exposed to vertical vibration. The measured data revealed large scatter due to large variations in the subject masses. Such scatter in the data have been widely observed (Rakheja et al., 2002; Nawayseh and Griffin, 2004; Wang et al., 2004; Mansfield and Maeda, 2006). A large number of studies have presented normalised AM responses, which greatly reduce the data scatter. Wang et al. (2004) showed that normalisation cannot eliminate the important effect of body mass on the biodynamic responses. Moreover, it could alter the essential trends in the magnitude response, which may make interpretations more demanding (Patra et al., 2008). As an example, Figure 4 illustrates the vertical AM magnitudes of subjects with body masses in the vicinity of 55, 75 and 98 kg together with the normalised magnitudes. The AM magnitudes vary substantially at low frequencies and in the vicinity of the primary resonance (4.4–5.3 Hz), while they seem to converge at frequencies above 10 Hz. The normalised values greatly suppress the variations at low frequencies and in the vicinity of the primary resonance, but emphasise the magnitude differences at higher frequencies. The lower body mass (55 kg) results in largest normalised magnitude at frequencies above 6 Hz, while the absolute AM response suggests largest magnitude of the 98 kg subjects. Giacomin (2004) showed that the normalised AM magnitude of infants is comparable to that of the adults. The mean primary peak frequency (6.25 Hz), however, was observed to be higher than those reported for adult subjects under vertical vibration, which was attributed to supine or semi-supine posture of the infants.

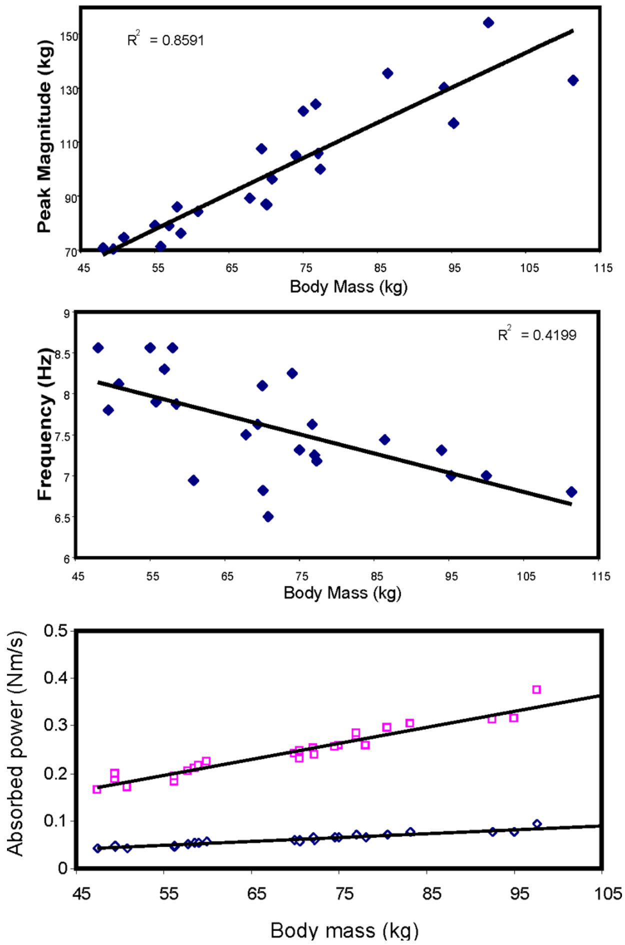
Figure 4 Effect of normalisation on the vertical apparent mass magnitude of subjects seated without a back support under 1 m/s^2 rms random vibration (Patra et al., 2008)



The vertical AM or MI magnitude is significantly positively correlated with body mass at frequencies up to and slightly above the primary resonance, while the frequency corresponding to the peak AM or MI magnitude is weakly negatively correlated with the body mass (Donati and Bonthoux, 1983; Wang et al., 2004; Nawayseh and Griffin, 2005b). Figures 5(a) and (b), as an example, illustrate correlations of peak AM magnitude and corresponding frequency with the body mass (Rakheja et al., 2002). Wang et al. (2004) further investigated correlations of AM magnitudes at different frequencies up to 12 Hz with the body mass for different body support conditions involving vertical and inclined back supports, flat and inclined seat pan, and two hands positions (on lap and on SW). The study concluded linear dependence of AM magnitude on the body mass at all the selected frequencies, irrespective of the body supports and hands position. The AM magnitude above 12 Hz was found to be less sensitive to body mass. The correlation between the AM phase response and the body mass and anthropometry could not be noted, although it has been attempted in a single study (Donati and Bonthoux, 1983).

The power absorbed by the body, attributed to vertical WBV, has also been positively correlated with the body mass (Lundström et al., 1998; Dewangan et al., 2018). Wang et al. (2006a) investigated relations between vertical vibration and VPA with a number of anthropometric variables for 12 different body support conditions through linear regression analyses. The study concluded positive linear correlations between VPA and body mass (Figure 5(c)). A poor correlation with sitting height was observed, although the standing height was a significant factor ($p < 0.0001$). An excellent positive correlation with body mass index (BMI), however, was demonstrated, irrespective of the seat support. The lowest correlation was obtained with body fat. Dewangan et al. (2013c) presented effects of various anthropometric parameters on AM responses, and concluded that the AM responses are strongly coupled with body mass and a number of anthropometric dimensions such as body fat and hip circumference. The peak AM magnitude was positively correlated ($r^2 > 0.7$) with body mass, BMI, body fat mass and hip circumference. However, the correlations were moderate with the lean body mass and percent body fat, and poor with the stature and seat-pan contact area. Such correlations between the 'to-the-body' responses to horizontal vibration have not been explored. Nawayseh and Griffin (2005b), and Rakheja et al. (2006), however, showed similar degree of positive correlation between the body mass and peak cross-axis fore-aft AM magnitude, measured under vertical vibration.

Figure 5 Correlations with the body mass: (a) peak vertical AM magnitude and (b) frequency of peak AM deduced from the responses of 24 subjects sitting in automotive posture with hands on thighs (Rakheja et al., 2002) and (c) absorbed power under vertical vibration (Wang et al., 2006a) (see online version for colours)



The influences of anthropometry on the 'through-the-body' response have been addressed in a few studies reporting STHT under vertical vibration. Donati and Bonthoux (1983) observed higher vertical vibration transmissibility to the thorax of taller subjects up to

4 Hz, while a negative correlation was observed between transmissibility magnitudes and the body mass at higher frequencies (8 and 10 Hz). This is opposite to that observed for the AM. Griffin and Whitham (1978) found negative correlation between STHT magnitude and the body size of adult male subjects, particularly the mass and hip size, at 16 Hz, while the correlation coefficients ranged from 0.3 to 0.4.

Furthermore, the vast majority of the studies report mean or median biodynamic responses of subjects with considerably different masses. The mean or median values thus do not illustrate the effects of body mass on peak magnitudes and the corresponding frequencies (Figure 4), nor do they represent the properties of subjects of particular masses. Moreover, the mean and median responses do not permit for analyses of individual contributory factors, which are strongly coupled with body mass effects, while they tend to suppress the secondary peaks in the responses. A few studies have attempted to isolate body mass effect from other factors by either considering subject population within a predetermined narrow body mass range or by grouping the data under different mass groups. MI responses of 37 subjects seated erect without a back support and exposed to vertical vibration were grouped in 4 different mass groups (<60, 60–70, 70–80 and >80 kg) to study the body mass effects (Seidel, 1996, cited in Boileau et al., 1998). The impedance data clearly showed slightly lower frequencies corresponding to peak magnitudes with increasing mass groups. The differences in these frequencies, however, vanished when the data was converted to AM, suggesting negligible effect of body mass on the primary resonant frequency (Wu et al., 1998). In a similar manner, Rakheja et al. (2002, 2006) and Wang et al. (2004) presented vertical AM responses in the same four body mass ranges. Rakheja et al. (2006) also grouped the fore-aft cross axis AM response magnitudes in four mass groups. The AM responses were grouped by Hinz et al. (2004) in accordance with the body weight percentile. The results of all these studies confirm that the peak AM magnitude increases with the body mass, while the corresponding frequency decreases, as reported by Patra et al. (2008) (Figure 4).

Considering the body mass effect, the biodynamic models of the seated body should not be formulated on the basis of the mean or median responses of subjects with widely varying body mass. Hinz et al. (2001) suggested that the biodynamic models formulated for predicting individual spinal loads should be validated by mean transmissibility derived from repeated measurements on the same individual. Patra et al. (2008) proposed that the reference values for standardisation may be established by considering subject populations in the vicinity of particular body masses. It was further suggested that three body masses (55, 75 and 98 kg) be considered in accordance with the standardised seat test method (ISO-7096, 2000), which may be adequate for model development and for the designs of anthropodynamic manikins for assessment of seats.

4.2 Gender effect

A few studies have investigated gender effect on the biodynamic responses of seated subjects, which appear to be somewhat contradictory. Some studies have concluded the gender effect was mostly insignificant (Mertens, 1978; Griffin et al., 1982; Parsons and Griffin, 1982; Fairley and Griffin, 1989; Rakheja et al., 2002), while others suggest otherwise. Through measurements, Laurent (1996) concluded that vibration transmissibility of a cushion seat is strongly influenced by the gender. Griffin and Whitham (1978) reported negative correlation of STHT magnitude and the body size (weight and hip) for male subjects at 16 Hz, and with body mass and height for the female subjects. Another study by Griffin et al. (1978) revealed higher STHT magnitudes for females than males at frequencies above 5 Hz, and an opposite trend was reported at frequencies up to 4 Hz. The differences were found to be significant and the STHT magnitude of females was nearly twice that of males at some higher frequencies. Wei and Griffin (1998) concluded insignificant gender effect from the statistical analyses of individualised one-dimensional single- and two-DOF model parameters, derived from vertical AM data of male and female subjects. The parameters of the single- and two-DOF models, however, consistently suggested considerably higher primary stiffness and lower damping of the male subjects' models than those of the females.

Lundström and Holmlund (1998) and Holmlund (1999) measured absorbed power of 15 male and 15 female subjects under horizontal and vertical WBV and concluded significant gender effect. The females absorbed more power per kg of seated body mass than the males. Significant gender effect was also reported on the MI response under horizontal vibration in the 2–6.3 and 18–31.5 Hz ranges; the effect, however, diminished under a higher vibration excitation of 1.4 m/s^2 (Holmlund and Lundström, 1998). The normalised MI magnitude for females was observed to be higher than the males, as in case of the absorbed power. The authors suggested that greater power absorption and MI magnitude of females may be attributed to their greater fat to muscle mass proportions, which would yield relatively higher damping, and lower muscle strength capacity. The contributions due to breast supports were also suspected and investigated, while the outcome did not show any effect of the support. Another study on vertical MI suggested a higher and distinct primary peak for male subjects in the 4–5 Hz range than the females, and an opposite trend near the second peak around 10 Hz, and insignificant gender effect on the impedance phase response (Holmlund et al., 2000). Mansfield et al. (2001), on the other hand, showed distinct second peak in vertical normalised AM magnitude around 10 Hz for the male subjects only, and concluded no gender effect on the AM magnitude and the primary resonant frequency. On the basis of observed differences, Lundström et al. (1998) suggested the use of different injury criteria for the two genders. Mansfield and Lundström (1999), however, observed lack of consistent trends in resonant frequencies under lateral and combined fore-aft and lateral vibration, while higher normalised AM magnitudes were observed for females under fore-aft and lateral, and combined fore-aft and lateral vibration at frequencies above 3 Hz.

The above studies have considered male and female subjects of considerably different body masses to arrive at the gender effect. The observations in this case would be most likely coupled with the body mass effect. Wang et al. (2004) investigated gender effect on the vertical AM by extracting responses of male and female subjects of similar body masses from the ensemble of 27 subjects. The results showed slightly higher mean magnitudes above 15 Hz and the presence of a more clear second resonance for the females, as reported by Holmlund and Lundström (2001). An ANCOVA analysis, however, revealed that the gender effect could be observed only at frequencies above 15 Hz, where the AM and the absorbed power are generally very small. Dewangan et al. (2013b, 2013c, 2018) reported a significant gender effect on AM, STHT and absorbed power responses measured with 31 male and 27 female subjects, which was strongly coupled with various anthropometric parameters. Female subjects showed lower primary resonance frequency compared to male subjects. A clear gender effect was established by considering data for male and female subjects of comparable body mass. While the peak magnitudes of both the genders of similar body mass were comparable, female subjects showed higher magnitude near the secondary resonance frequency, which was attributed to relatively greater pelvic and visceral mass of females. Owing to higher body stiffness, the softening tendency with increasing vibration magnitude was more pronounced for male subjects as compared to the female subjects. Furthermore, the softening behaviour of both the genders was coupled with the sitting condition.

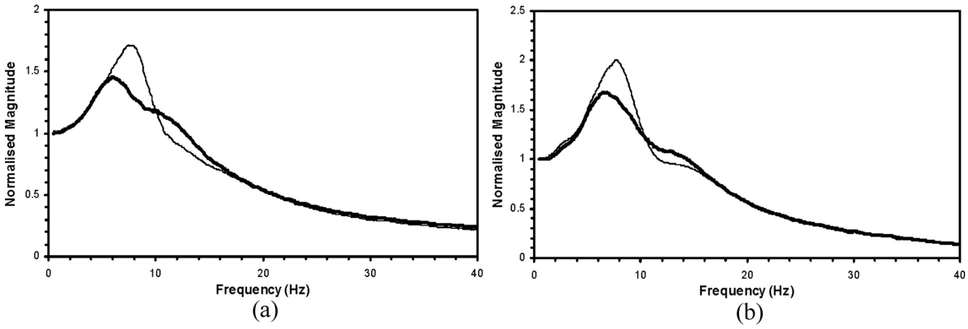
4.3 Effect of hand supports

The biodynamic responses have been mostly reported for the body seated with the hands on the lap or on the thighs or arms folded across the chest, which cannot be considered representative of vehicle driving postures. ISO-5982 and DIN-45676 also define the vertical MI and STHT reference values of for hands on lap or thighs. Only a few studies have measured responses of the seated subjects with hands on a bar or a steering wheel (SW) (Hinz et al., 2002; Rakheja et al., 2002; Wang et al., 2004, 2006a; Stein et al., 2006, 2009; Patra et al., 2008; Mandapuram et al., 2010, 2011; Pranesh et al., 2010; Toward and Griffin, 2010). Some of these have also investigated relative effects of the hands position. The vibrating SW would represent another source of vibration into the body or an additional driving-point. Furthermore, the hands on the SW or a bar reduce the proportion of body weight on the seat pan. It may also cause stiffening of the body, and alter the pelvic orientation, which has been shown to influence the biodynamic responses (Pope et al., 1990; Zimmermann and Cook, 1997; Nishiyama et al., 2000).

Wang et al. (2004) reported that sitting with an inclined backrest coupled with hands on the SW yields a more pronounced secondary mode near 10 Hz, while the effect is insignificant above 12 Hz. A relatively smaller effect of the hands position, however, was observed on vibration transmitted to the spine and head by Pranesh et al. (2010), while the hands support resulted in higher peak vibration magnitude at C7 and L5. The hands support, however, showed a strong influence on the fore-aft response at C7 in the absence of a backrest. Patra et al. (2008), on the other hand, noted only minor differences due to hands position on the vertical AM responses of subjects with comparable body masses, irrespective of the back support. The effect was particularly negligible for body mass in the vicinity of 98 kg.

Figures 6(a) and (b) illustrate effect of hands position on the mean vertical AM, measured at the pan and the backrest, respectively, of subjects assuming automotive sitting posture with hands on laps and on the SW (Rakheja et al., 2006). The mean magnitude at the pan reveals single-DOF like behaviour for the hands in lap posture, while that at the backrest shows an additional secondary peak near 14.5 Hz. This may relate to the interactions associated with the upper body modes. The mean magnitude response with hands on SW exhibits the presence of both modes in the pan as well as the backrest responses, occurring around 6.5 Hz for the pan and 7.1 Hz for the back, and secondary mode in the 12–14 Hz range (Wang et al., 2004). The hands on the SW cause the peak AM magnitude to decrease substantially, while the primary frequency also decreases. The frequencies corresponding to the primary peak were considerably higher than those reported in vast majority of the studies (5–7 Hz). The difference is attributable to combinations of relatively low level of vibration (0.25 to 1.0 m/s^2 rms over a wide frequency range of 0.5–40 Hz), large pelvic rotation caused by the inclined pan (13° with respect to horizontal) and backrest (24° with respect to a vertical axis) and low seat height of 220 mm. Toward and Griffin (2010) observed that when subjects supported by a backrest held a SW, an additional resonance was evident around 4 Hz. Moving the SW away from the body reduced the AM at the primary resonance, and the AM magnitude at the 4 Hz resonance increased, suggesting that the 4 Hz resonance is associated with the arms and shoulders. Raising the SW had a similar, but smaller, effect to moving the SW forward.

Figure 6 Influence of hands position on the mean AMPS responses of seated occupants: (a) seat pan APMS; (b) Backrest APMS (—— , hands in lap; **——** , hands on steering wheel) (Rakheja et al., 2006)



Hands on the SW also resulted lower total VPA (Figure 7) as well peak power response in the vicinity of the primary resonance, irrespective of the back support orientation (Wang et al., 2006a). Higher VPA at higher frequencies, however, was obtained due to the pronounced secondary mode with hands on SW. Unlike the 'to-the-body' responses, the effect of hands position was more pronounced on the vertical and fore-aft STHT responses, especially for sitting without back support (Wang et al., 2006b). Sitting without a back support and hands on the SW resulted in greater magnitudes of vertical head vibration in the 3–10 Hz range, while the effect was notable at frequencies above the primary resonance for the back supported postures (Figure 8). The hands on the SW also caused slightly greater fore-aft vibration of the head in the vicinity of the primary resonance frequency. Mandapuram et al. (2010) also found that the hands on the SW sitting condition yields higher fore-aft and lateral AM magnitudes compared with those attained with hands on the laps in the 1–8 Hz range, while only minimal effect was observed on the fore-aft STHT, when seated without a back support. The responses with a back support, however, revealed considerably higher magnitudes with hands on the SW compared to those with hands on lap under both fore-aft and lateral vibration, particularly in the vicinity of the resonance. The study also observed strong effects of hands support at frequencies about 1.8 Hz in the backrest AM responses under fore-aft vibration.

Figure 7 Influence of hands position on total power absorbed by the seated 10 subjects with comparable body mass (70.5–78 kg) as a function of the back support under vertical vibration (Wang et al., 2006a)

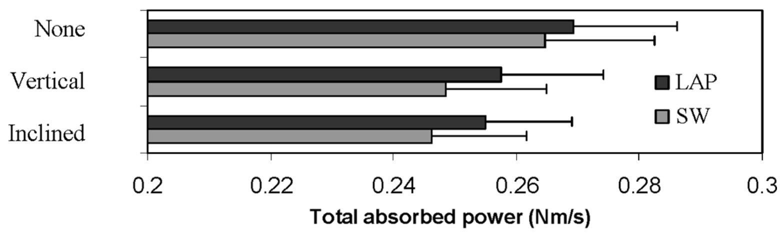
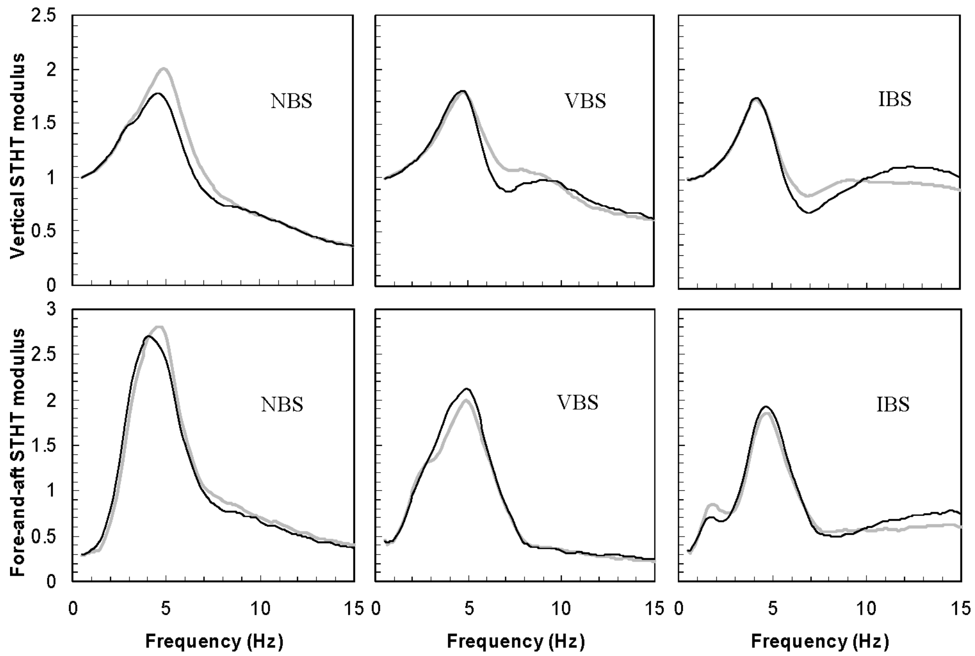


Figure 8 Influence of hands position on mean STHT responses of 12 subjects under different back supported conditions;—— hands in lap;━━ hands on steering wheel (excitation: 1.0 m/s^2 rms; NBS – no back support; VBS – vertical back support, and IBS – inclined back support) (Wang et al., 2006b)



The effect of hands position could also be noted from the few studies reporting biodynamic responses of body seated on cushioned seats. The effect of arm angle on vertical vibration transmitted to the head, chest and hip was observed to be very small, while the effect on thigh and shin vibration was most significant, which was most likely due to pelvic orientation and associated postural changes (Nishiyama et al., 2000). Unlike the vast majority of the reported studies, the low frequency vibration transmissibility magnitudes in this study were well below 1.0 (as low as 0.25), irrespective of the measurement location, which was attributed to multi-DOF dynamic behaviour of the seated body. The study by Stein et al. (2006) showed most significant effect of hands position during fore-aft vibration. The peak AM with hands on the SW was nearly twice that attained with hands in lap, when the SW was close to the seat. The peak magnitude decreased with increase in distance between the seat and the SW but remained considerably higher than that with hands on the lap. The frequency corresponding to the peak was also higher for hands on the SW.

The above studies suggest that the effect of hands position is strongly coupled with many other factors, such as backrest inclination, seat height and pelvic orientation, and further investigations are needed to identify the contributions of the hands position. It has also been suggested that placing the hands on the thighs may help dampen the higher modes of vibration (Holmlund and Lundström, 1998).

4.4 Effects of feet support and position

The majority of the reported studies have employed either fixed or adjustable feet supports, while the heights are not specified. The height of the feet supports can substantially affect the body mass supported by the seat pan, contact with the thighs, pelvic orientation and the upper body posture. Some of the earlier studies reported biodynamic responses of the body seated with feet unsupported, which is not representative of the vehicle environment (Coermann, 1962; Vykukal, 1968; Miwa, 1975; Mertens, 1978; Matsumoto and Griffin, 1998a). In this situation, the entire body weight is supported by the seat, which causes significantly higher vertical AM of the body up to 10 Hz. Few studies have employed a stationary footrest, which can cause relative motions across the legs under vertical vibration and thus influence the measured responses. Fairley and Griffin (1989) investigated the effect of height of a static footrest on the vertical AM response, and reported substantially lower AM magnitude at low frequencies (1–2 Hz) for a lower foot rest. The effect of the height was very small in the 2–10 Hz range. The effect of height of a moving footrest, however, was observed to be very small due to negligible relative motion across the legs. The AM magnitude, however, increased as the height of the moving footrest was lowered, as opposed to that observed for the stationary footrest. Nawayseh and Griffin (2003, 2004, 2005a) in a similar manner investigated the footrest position under vertical and fore-aft vibration, where the height was varied to achieve maximum thigh contact, average thigh contact, minimum thigh contact, and legs hanging freely. The vertical AM at the seat showed highest magnitudes for legs hanging; and comparable magnitudes for the other three footrest positions, suggesting only small effects. Rakheja et al. (2002) also reported negligible effect of leg angle on the vertical AM measured under automotive sitting posture. The similar trend was found in the total VPA by Nawayseh and Griffin (2010) in a study with same feet positions. The effect of thigh contact on the resonance frequencies was also insignificant under vertical vibration but significant under horizontal vibration. The resonance frequency increased as the legs were moved away from seat and then decreased as the legs approached extreme horizontal (Toward and Griffin, 2010).

Griffin et al. (1978) investigated the effects of leg orientation and the height of the feet supports on the STHT response, and concluded that varying the leg angle or feet support height had only little effect, even when the legs were hanging. Sitting with a higher footrest so as to the raise the thighs well above the seat level resulted in slightly lower STHT in the 6–9 Hz range, which was attributed to reduced thigh contact.

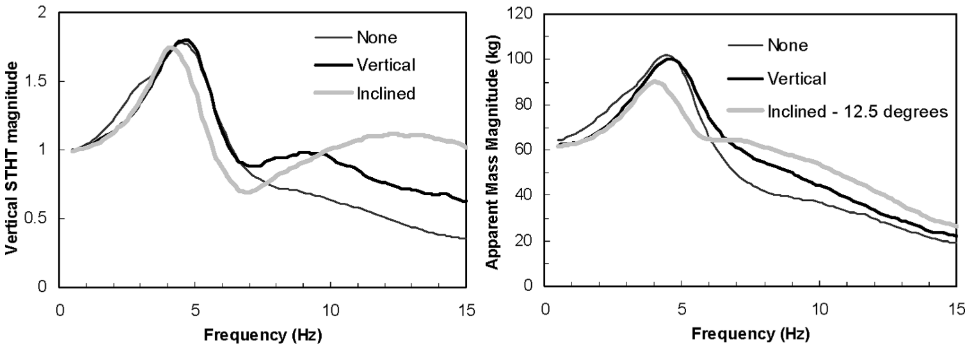
4.5 Effects of seat geometry and back support

The vast majority of the reported studies on experimental biodynamics have considered subjects seated without a back support. While this is not representative of the sitting condition in the majority of vehicles, but it justifies the consideration of a single driving-point at the buttock–seat interface. A coupled occupant-seat system in a WBV environment constitutes multiple driving-points, where the vibration enters the body at the buttocks, hands, feet and the back.

The effect of backrest support on 'to-the-body' and 'through-the-body' responses of the seated body to vertical and horizontal vibration have shown strong influences despite the differences in the seat and backrest geometry. Figure 9 illustrates the effect of back support on the mean measured AM and STHT magnitudes under vertical vibration.

Paddan and Griffin (1988a, 1988b, 1994, 2000) investigated the influence of back support on STHT along all the six axes under vertical, horizontal, roll, pitch and yaw vibration, applied independently. Under vertical vibration, the STHT magnitude increased with the back support (inclined at 6°) in the 0.25–20 Hz range. The vertical STHT magnitude with back support was significantly higher than that without a back support at frequencies above 4.5 Hz, while the use of backrest caused the primary frequency to increase substantially from 4.2 Hz (without back support) to 6.2 Hz. Wang et al. (2006b) also reported an increase in vertical and fore-aft motions of the head at frequencies above 8 Hz due to an inclined backrest, while the shift in primary frequency, reported by Paddan and Griffin (1988a), was not observed (Figures 8 and 9). Pranesh et al. (2010) reported that the back support has significant influence on vertical vibration transmission at C7, T5, T12, L3 and L5, and head, while the effect on the horizontal responses measured at the lower regions of the torso, namely T12, L3 and L5, was notable only in the lower frequency range.

Figure 9 Influence of back support on the STHT, and AM under vertical vibration (Wang et al., 2006b)



The magnitudes of STHT strongly rely on inclination of the backrest. A distinct peak in the *x*-axis motion near 2 Hz, attributed to pitch rotation of the upper body (Kitazaki and Griffin, 1995), was also noted with an inclined backrest. Toward and Griffin (2009) and Shibata and Maeda (2010) studied the effect of backrest inclination up to 30° on the AM responses and observed the shift in principal resonance frequency. Increasing the back incline to 30° resulted in lower peak normalised AM magnitudes near 5 Hz with a notable secondary peak near 7.5 Hz (Shibata and Maeda, 2010). Increasing the backrest inclination also resulted in slight reduction in the total VPA computed in the 1–20 Hz frequency range. Toward and Griffin (2009) also reported an increase in primary resonance frequency in the vertical AM with increasing inclination of a rigid backrest up to 30^{o}, while an opposite trend was observed with foam backrests. Only minimal effect of foam thickness on the AM magnitude was observed for backrest inclinations below 30°, but at 30°, an increase in foam thickness resulted in lower primary resonance frequency. It was further reported that a back support, whether vertical or inclined, helps to limit fore-aft motion of the head substantially at frequencies below 8 Hz.

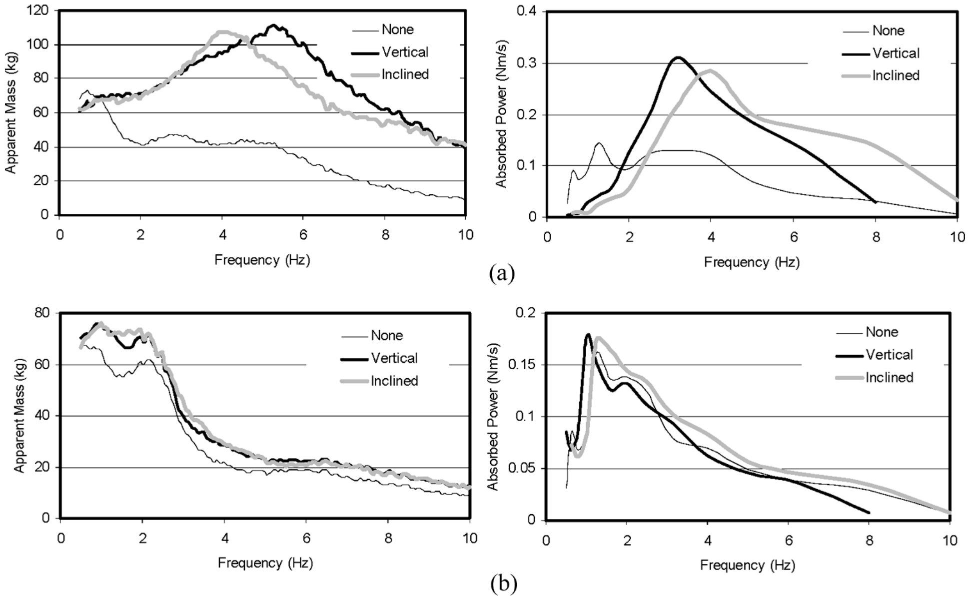
Under *x*-axis vibration, the presence of a backrest also caused considerably higher magnitudes of fore-aft, vertical and pitch vibration of the head in the entire frequency range, while the fore-aft STHT phase was substantially smaller than that obtained without the back support (Paddan and Griffin, 1988b). The effect of backrest on the STHT of the *y*-axis vibration, however, was small, but it caused greater roll motion of the head. The stiffening effect of the backrest support was also observed in the STHT under the yaw vibration, which resulted in a higher primary frequency of 3 Hz compared to 2 Hz with no back support. The effect of back support on the fore-aft responses was observed to be most significant in the entire frequency range (Mandapuram et al., 2010). The effect was also significant on lateral AM response, although relatively small, which is again attributable to the motion resistance offered by the back support. Magnusson et al. (1993) concluded that the backrest inclination ranging from 10° forward to 30° backward had only minor effect on the transmitted vibration to the L3 under a vertical shock motion.

The data acquired with back supported postures also revealed significant reduction in the inter-subject variability in all of the reported studies (Paddan and Griffin, 1988a, 1988b; Hinz et al., 2002; Wang et al., 2006a), which was attributed to increased upper body stability. This may also be related to peak body-seat interface pressure, which was observed to be considerably lower for sitting with a back support compared to the unsupported sitting (Wu et al., 1999). Lower inter-subject variability in AM has also been reported with increasing seat pan angle, which also contributes to more stable sitting (Nawayseh and Griffin, 2005b; Wang et al., 2006b). The influence of seat pan angle on the vertical AM magnitude and the corresponding frequency was reported negligible for pan inclination up to 10° in two studies. The primary vertical resonance frequency, however, increased considerably with a pan angle of 15° coupled with a vertical back support. This may be attributed to pelvic orientation, increased stability and stiffening of the buttock tissue by shifting of the interface contact pressure towards the tuberosities.

A number of studies have systematically analysed the effects of vertical and inclined back supports on ‘to-the-body’ responses, particularly under vertical vibration. The support against a backrest suppresses the peak vertical AM magnitude considerably compared to the no back support, while the effect on the primary frequency is very small (Wang et al., 2004; Nawayseh and Griffin, 2005b; Patra et al., 2008). The back support, however, yields relatively higher AM magnitude at frequencies above 5.5 Hz, and the larger bandwidth suggests an increase in the energy dissipation property of the body. The use of a back support also reduces the VPA at frequencies below the principal resonance but yields higher VPA at higher frequencies (Nawayseh and Griffin, 2010). The study, however, reported negligible effect of back support on the total VPA, except when sitting with minimal thigh contact. Wang et al. (2004) further showed that the bandwidth of the response increased with an inclined back support compared to the vertical back support. Similar significant effects of back supports on the vertical MI magnitude have also been reported by Boileau and Rakheja (1998). The peak MI magnitude and the corresponding frequency for the no-back support were observed to be higher than those obtained for vertical and inclined (14°) back supports. The inclined back support, however, resulted in higher frequency corresponding to the peak compared to the vertical support. The difference in the frequency, however, would most likely diminish, if the data were to be presented in terms of AM. Mansfield and Maeda (2007) reported that the peak vertical AM with a vertical support was 6% lower than that without a support, which was attributed to the portion of body mass supported by a vertical backrest, while the effect on the median primary frequency was small. The effect of vertical back support on the peak AM magnitude under twisted postures was observed to be very small, although the back support resulted in slightly higher primary frequency (Mansfield and Maeda, 2005a).

Fleury and Mistrot (2006) reported AM responses of body seated on a seat with a locked horizontal suspension and exposed to fore-aft vibration, considering 3 different back support conditions: none, lumbar region support only and full back support. The AM data revealed peaks at 2.5, 3 and 2.8 Hz, respectively, for the three support conditions, suggesting stiffening effect of the back support under fore-aft vibration. The primary frequencies of the fore-aft AM responses of the unsupported body seated on a rigid seat, however, have been reported to be considerably lower: 1.3 Hz (Lee and Pradko, 1968); and near 0.7 Hz (Fairley and Griffin, 1990; Mandapuram et al., 2005, 2010). The latter three studies have shown that the back support causes significantly higher AM magnitude and the corresponding frequency under horizontal vibration, while the effect is small under lateral vibration (Figure 10). The addition of a backrest showed only one peak in the 3.5–5.5 Hz range in the *x*-axis and 1.5 Hz in the *y*-axis. It was concluded that the back rest limits the rocking and swaying motions by stiffening the upper body under fore-aft vibration. Mansfield and Maeda (2006) also illustrated the same effects of back support on the AM responses under *x*-, *y*- and combined *x*-, *y*- and *z*-axes vibration.

Figure 10 Influence of back support on the apparent mass and absorbed power properties under fore-aft (a) and lateral (b) vibration (Rakheja et al., 2006)



Wang et al. (2006a) investigated the effect of seat height, pan inclination and back support on the power absorption under vertical vibration. The study reported negligible effect of pan angle (0–5°). Highest total VPA was observed for unsupported back followed by the vertical and inclined back supports (Figure 7). The peak power for the unsupported back occurred in the 6.3 Hz, while that for the inclined support in the 8 Hz band. The seat height effect was notable for inclined back support only, where a higher seat resulted in higher total power and power spectrum above 8 Hz; which was attributed to greater interactions of the upper body with the back, particularly with the hands on the SW. Rakheja et al. (2008) observed a significant effect of back support on VPA under small x-axis vibration (Figure 10). The use of a back support caused stiffening effect with significantly higher peak in power with corresponding frequency shifting towards 4 Hz, compared to 1.25 Hz for the no back support. However, supporting the back with the vertical backrest decreased the power absorbed at the seat at low frequencies but caused higher VPA at higher frequencies.

From the reported studies, it is apparent that the geometry of the seat affects the biodynamic responses to WBV in a significant manner. The pan and backrest inclinations, seat height, and seat-to-footrest and seat-to-steering distances, can greatly affect the sitting posture by changing the spine curvature and pelvic orientation, and thus the mechanical properties (stiffness and energy dissipation) of the body. Furthermore, greater inertial forces would develop at the backrest interface due to relatively large proportion of the body mass supported by the backrest. A backrest helps limit the horizontal and rotational body motions, particularly in the sagittal plane. The friction between the backrest and the upper body could also help limit the lateral and roll motion of the body. Exposure to vertical vibration is known to yield considerable fore-aft and pitch motions of the upper body (Griffin, 1990), which tend to instigate considerable dynamic interactions with the back support. Such interactions have been investigated in a few studies through measurement of forces at the upper body-backrest interface and cross-axis AM. Nawayseh and Griffin (2004, 2005b) and Rakheja et al. (2006) showed considerable magnitudes of fore-aft force developed at the upper body-backrest interface under vertical vibration, which tended to be higher for an inclined backrest, while the forces along the lateral axis were small, suggesting greater coupling between the vertical and fore-aft motions of the body. This behaviour was also reported by Mansfield and Maeda (2006), which presented AM responses under vibration along single and multiple axes. The fore-aft cross-axis AM was most significantly affected by the backrest, which resulted in relatively higher frequency, as observed in studies reporting horizontal AM. It was further shown that greater seat pan angle yields greater magnitudes of cross-axis AM, which was attributed to increased interactions of the upper body (Nawayseh and Griffin, 2005b).

4.6 Effects of postural variations

Studies have reported the biodynamic responses under different controlled sitting postures in order to enhance understanding of the nonlinearity in biodynamic responses. These include sitting erect or tense, relaxed, slouched and twisted upper body. It has been shown that an erect posture with an unsupported back causes higher contact pressure and force at the buttock-seat interface than an erect posture with a back support (Wu et al., 1998). Some of the postural effects are clearly evident from the responses acquired without a back support, while others show inconsistent trends, due to strong coupled effects of muscle tension with the intensity of vibration. The body stiffness and the vertical mode resonance frequency would be higher for postures involving tense muscles, while increasing the vibration intensity is known to soften the body, which may be attributed to thixotropic behaviour of the human muscles (Hagbarth et al., 1995).

The WBV-induced head motion is particularly more sensitive to postural changes. Griffin (1975) observed that two extreme postures produced STHT by as much as 600% in the 7–75 Hz range, even though the postural changes were not quite obvious to an observer. Griffin et al. (1978) further reported that a stiff posture causes greater mean STHT at frequencies above 6 Hz but lower below 6 Hz, compared to the normal sitting.

The STHT response with a relaxed posture was comparable with the normal posture, with magnitude being only slightly lower at frequencies above 10 Hz. The study also measured the head motion of a single subject under seven different postures, ranging from fully erect to completely slouched. The erect posture resulted in considerably higher STHT magnitude at frequencies above 5 Hz, particularly the peak magnitudes in the 10–15 Hz range, while the phase response was considerably smaller than that for the slouched posture. The STHT magnitude corresponding to the slouched posture was well in excess of 1.0 (near 1.4) at the low frequency of 1 Hz, while the peak magnitude near the fundamental frequency (around 6 Hz) was below 0.6, suggesting possible measurement errors. Coermann (1962) reported somewhat contradictory finding; the relaxed posture resulted in higher peak STHT magnitude and lower fundamental frequency compared to the erect posture. The study also reported only little effects of muscle tension on the STHT magnitude, which was measured with only one subject with tensed arms, shoulder, leg, neck and abdomen muscles.

Pope et al. (1990), on the other hand, reported lower vibration transmitted to L3 in the erect posture than the relaxed posture under vertical shock motion. Hinz et al. (2002) observed lower STHT for the 'forward bending' posture in the 2.5–5.5 Hz range, while sitting relaxed with unsupported back caused STHT peak near 4.1 Hz, which reduced to 2.89 Hz for the 'forward bending' posture. Relatively lower resonance frequencies could be due to coupling with the elastic seat used in the study. The difference in pelvic orientations corresponding to the forward and backward bending sitting was nearly 17°. The pelvic rotation during vibration, however, was found to be greatest for the backward bending posture in the 5–7 Hz range, which was negligible above 7 Hz, most likely due to small input displacement (Zimmermann and Cook, 1997). The study also claimed comparable trunk acceleration for all three postures, although the reported data showed considerable differences around 5 Hz. Moreover, both the trunk and head acceleration transmissibility magnitudes were above unity value in most of the frequency range.

The effect of a lumbar support on vertical vibration transmitted to lumbar spine (L1–L5), head, chest and sternum, investigated with seven fresh cadavers, showed extreme inter-subject variability (El-Khatib et al., 1998). The study claimed nearly constant vertical vibration transmissibility of all the lumbar spine locations, suggesting that measurement at a single point would be equally representative. Moreover, the mean seat-to-lumbar vertebrae transmissibility magnitude remained close to unity up to 25 Hz, with a small magnitude resonance peak in the 5–7 Hz range. The addition of lumbar support resulted in relatively higher resonance frequencies.

Coermann (1962) reported considerably higher MI magnitude and the corresponding frequency under an erect posture compared to the relaxed sitting posture under vertical vibration, suggesting lower muscle stiffness and higher damping with a relaxed posture. Miwa (1975), on the other hand, showed only small differences in MI with an erect and relaxed posture, with the relaxed posture causing only slightly lower MI magnitude with negligible change in the corresponding frequency. Fairley and Griffin (1989) reported higher fundamental frequency and higher magnitude of AM under erect posture compared to the slouched posture. Boileau and Rakheja (1998) and Kitazaki and Griffin (1998) also reported similar differences in vertical MI and AM responses, respectively, due to erect and slouched postures, while the postural differences in the two studies were very small compared to that reported by Coermann (1962). Kitazaki and Griffin (1998) reported mean resonance frequencies of 4.4 and 5.2 Hz, respectively for the slouched and erect sitting, and greater shear deformation of the buttock tissue in the slouched posture.

These studies further showed larger differences between the resonance frequencies associated with slouched and normal postures. This was in-part believed to be caused by shifts in the upper body mass centre and the buttock-seat centre of pressure.

Holmlund et al. (2000) reported similar differences in peak vertical MI and corresponding frequencies between the erect and relaxed sitting postures. Lundström and Holmlund (1998) investigated VPA responses under *x*-, *y*- and *z*-axes vibration for subjects seated with erect as well as relaxed postures. Peak VPA magnitude under vertical vibration was higher and occurred at a lower frequency for the relaxed posture than the erect posture, as it is observed in the MI and AM data, which was attributed to relaxed back and abdominal muscles for the relaxed posture. Under *x*- and *y*-axes vibration, however, the data did not show differences, although the posture was found to be a significant factor. The influences of relaxed and erect sitting on the MI under horizontal vibration have been reported in a single study (Holmlund and Lundström, 1998). The MI magnitude response under fore-aft vibration revealed almost one-DOF like behaviour with peaks occurring in 2–4 Hz range for the erect posture, while the relaxed posture resulted in a distinct primary peak near 2 Hz and a secondary peak in the 5–6 Hz range. Under lateral vibration, the effect of posture was strongly coupled with the gender effect and vibration magnitude.

Mansfield and Griffin (2000) investigated the effect of posture on AM and pelvis pitch responses to vertical vibration considering 9 different postures, including normal upright, anterior and posterior lean, kyphotic or slouched, increased pressure on the tuberosities, sitting on a bead cushion and sitting with an elastic belt. The peak pelvis pitch was observed in the 10–15 Hz range for all postural conditions, as opposed to 5–7 Hz reported by Zimmermann and Cook (1997), while sitting on a bead cushion resulted in highest pelvis pitch transmissibility. Sitting with normal upright back support and high pressure at the ischial tuberosities showed peak pelvis pitch near 10 Hz. The effect on frequency corresponding to the peak AM magnitude, however, was not evident under higher magnitudes of excitations (1.0 and 2.0 m/s^2 rms) for most of the postures. Significant differences, however, were observed under lower excitation of 0.2 m/s^2.

The effect of periodic muscle activity on the AM responses to vertical vibration were investigated by considering seven different upper body postures, including upright sitting as the reference, tense upper body, and periodic back-abdomen, and folding-stretching arms back-front and rest to front (Huang and Griffin, 2006). Sitting with tense upper body resulted in higher resonance frequency and wider bandwidth than the normal sitting posture, suggesting high stiffness and damping properties under sustained voluntary muscle tension. The periodic bending postures resulted in lower peak AM and slightly lower frequencies compared to the upright posture.

The drivers of forklifts, farming and construction vehicles often assume twisted trunk postures, which have been associated with greater energy absorption (Magnusson et al., 1987). Mansfield and Maeda (2005a) reported that the AM responses of most subjects with twisted postures (shoulders oriented around 45° to the mid-sagittal plane without backrest contact) were similar to those obtained for sitting upright with back-on or back-off. The study also observed that moving postured (repetitive change in the posture every 2 s) resulted in considerably lower AM magnitude at frequencies below 6 Hz, which was attributed to low frequency movements. Furthermore, the moving posture suppressed the secondary peak in AM near 12 Hz that was clearly evident for twisted and upright sitting postures. Mansfield and Griffin (2002) further showed that most small postural changes yield only small effects on the AM or vibration transmissibility.

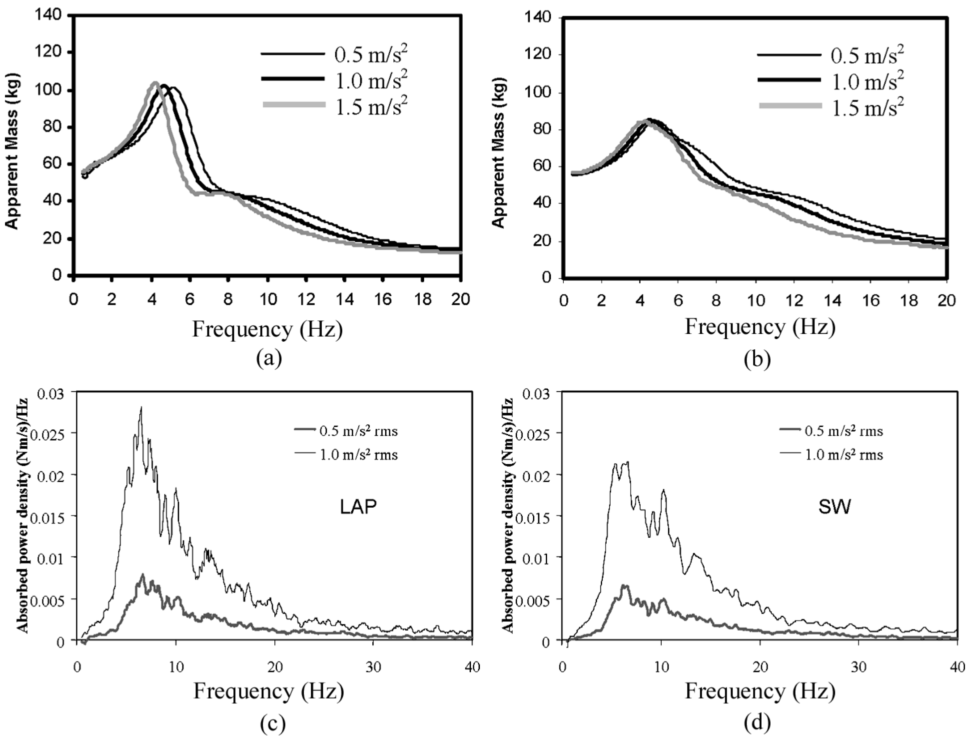
Moreover, the biodynamic responses have generally been evaluated for subjects sitting in rigid seats and thereby the effect of possible relative motion across the legs is not known. In a vehicular environment, the lower limbs would encounter relative movements due to deflections of suspension at the seat and/or seat cushion, which may cause increased pitch rotation of the pelvis. It was shown that lower limb motion reduces the peak AM and increases the corresponding frequency (Lemerle and Boulanger, 2006). The effect may be related to increased pelvis rotation, while seated on a free vertical suspension. Body coupling with an elastic cushion, on the other hand, yields lower peak AM magnitude and the corresponding frequency (Dewangan et al., 2013a, 2015). The primary frequency of the body coupled with the seat is further dependent on viscoelastic properties of the seat cushion.

4.7 Effects of vibration type, magnitude, and frequency

The biodynamic responses of the seated body have been mostly evaluated under idealised sinusoidal or broad band random vibration with flat acceleration spectral density. A few studies have also investigated the responses to field measured vibration synthesised in the laboratory (Boileau and Rakheja, 1998; Hinz et al., 2002; Rakheja et al., 2002). Donati and Bonthoux (1983) measured MI responses to vertical sinusoidal and broad-band vibration with identical overall rms acceleration of 1.6 m/s^2 in the 1–10 Hz range. The study concluded insignificant effect of type of stimuli, except in the vicinity of the primary resonance, where MI response to sinusoidal vibration was slightly larger. This may be caused by voluntary postural changes attributed to enhanced subjective sensation to sinusoidal motions compared to the random motions (Mandapuram et al., 2005). Mansfield and Maeda (2005b) measured AM responses to different types of vertical vibration, namely: broad-band random in 1–40 Hz range (1 m/s^2), and discrete sinusoidal vibration at centre frequencies of octave bands with weighted rms acceleration of 0.2 and 0.4 m/s^2 at 1 and 2 Hz, respectively, and 0.5 m/s^2 above 2 Hz. Both types of stimuli resulted in similar AM magnitude despite their different magnitudes; the significant differences were obtained only in the 1 and 16 Hz bands. The peak response to sinusoidal vibration occurred at 4 Hz compared to the median frequency of 5.2 Hz under random vibration, since the sinusoidal vibration was applied at selected discrete frequencies only. Boileau and Rakheja (1998) reported MI responses to vertical broad-band random, sinusoidal and vibration spectra of agricultural tractors and construction vehicles, defined in ISO-5008 (1990) and ISO-7096 (2000), and showed similar MI magnitudes under all stimuli. Rakheja et al. (2002) also showed only small differences in AM of subjects assuming automotive posture exposed to broad-band random and track-measured vibration of comparable magnitude. Mandapuram et al. (2005), on the other hand, noted considerable differences in biodynamic responses under fore-aft and lateral sinusoidal and random vibration, particularly at very low frequencies, which were attributed to enhanced perception of sinusoidal motion than the random vibration. The subjects revealed higher voluntary muscle tension and greater shifting of body weight to the feet under high magnitude sinusoidal vibration. The VPA due to multiple axes vibration arising from city buses, forestry skidders and mining trucks have been evaluated on the basis of measured transfer functions (Mandapuram et al., 2015). The results were quite different from those observed under broad band vibration. Mansfield et al. (2001)

measured AM and absorbed power properties under continuous vertical vibration coupled with equally and unequally spaced shocks. The AM resonance frequencies were slightly higher in the presence of shocks, although the effect of spacing of shock pulses was not notable. With increasing magnitude of the stimuli, the body revealed a stiffening effect under exposure to shocks, as opposed to the softening effect under contours vibration, which may be caused by higher voluntary tension under shock motions.

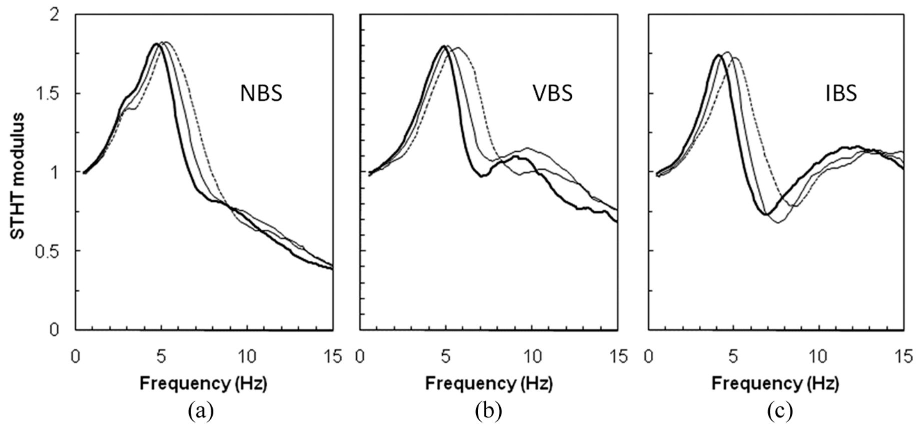
Figure 11 Comparisons of apparent mass (top – 9 subjects; body mass = 75 ± 7 kg) and absorbed power responses (bottom – 27 subjects) under different magnitudes of vertical vibration: (a) without a back support; (b) with a back support; (c) with back support and hands in lap and (d) with a back support and hands on steering wheel (SW)



The reported AM responses have consistently shown nonlinear dependency on the magnitude of vibration (Hinz and Seidel, 1987; Fairley and Griffin, 1990; Mansfield and Griffin, 2002; Rakheja et al., 2002; Wang et al., 2004; Mansfield et al., 2006; Mandpuram et al., 2010; Pranesh et al., 2010; Qiu and Griffin, 2010). The to-the-body responses have consistently shown a softening tendency of the body under increasing vibration magnitudes, irrespective of the direction and type of vibration, although considerable differences exist in the quantitative sense. Moreover, the findings appear to be inconsistent with regard to effect of excitation magnitude on the response magnitudes. While the fundamental mechanism leading to such nonlinearity has not yet been clearly identified, it is believed to be attributed to strong contributions of other factors, such as body support, gender, and support and postural conditions, and thixotropic and time-varying properties of the muscles. Greater softening tendency has been reported for male subjects compared to the female subjects, which was coupled with the sitting condition (Dewangan et al., 2013c). Relatively small effect of vibration magnitude on the peak vertical AM or MI responses have been reported under normal upright sitting without a back support (Fairley and Griffin, 1989; Boileau and Rakheja, 1998; Mansfield et al., 2001, 2006; Mansfield and Griffin, 2002; Rakheja et al., 2002; Wang et al., 2004; Huang and Griffin, 2006; Patra et al., 2008), as seen in Figures 11(a) and (b). Others have shown either increase (Matsumoto and Griffin, 2002; Nawayseh and Griffin, 2005a, 2005b) or decrease (Holmlund et al., 2000; Hinz and Seidel, 1987) in peak AM or MI magnitude with increase in vertical vibration magnitude.

Only a few studies report the effects of vibration magnitude on the 'through-the-body' responses. The STHT response of a single subject to different magnitudes of vertical vibration, ranging from 0.4 to 2.8 m/s^2 rms, showed negligible change in STHT magnitude and phase (Griffin et al., 1978). The study also investigated the STHT responses under three different vertical vibration spectra (equal energy, and dominant low and high frequency contents) of identical overall magnitude, and concluded very little effect above 9 Hz, while the STHT magnitude was inversely proportional to the low frequency energy of the input spectrum. Other studies have shown that the primary as well as secondary resonance frequencies decrease with increasing vibration magnitude (Hinz and Seidel, 1987; Wang et al., 2006b). The STHT responses have consistently shown shifts in the resonance towards lower frequencies under increasing vertical vibration (Figure 12). These have also shown relatively small changes in peak STHT magnitudes due to change in vibration magnitude. The magnitude of the fore-aft head motion, however, increased considerably in the 3-8 Hz frequency range, with increasing vertical vibration magnitude, irrespective of the back support condition (Wang et al., 2006b).

Figure 12 Influence of excitation magnitude on mean vertical APMS and STHT responses of six subjects with hands in lap posture: ——1.0 m/s^2 rms; ····· 0.5 m/s^2 rms; —— 0.25 m/s^2 rms; (a) No back support; (b) Vertical back support and (c) inclined back support



The softening property under increasing vertical vibration has also been demonstrated in the vertical and pitch transmissibility responses measured at vertebrae, pelvis, and abdominal walls (Mansfield and Griffin, 2002; Matsumoto and Griffin, 2002; Pranesh et al., 2010). Matsumoto and Griffin (2002), however, showed relatively small change in peak pitch magnitude at various thoracic and lumbar locations, while the head pitch motion decreased considerably with increasing vertical vibration magnitude. Similar discrepancies also exist in the AM or MI magnitude responses to horizontal vibration. Mansfield and Lundström (1999) and Mandapuram et al. (2005) showed increase in normalised fore-aft as well as lateral AM in the vicinity of the respective fundamental frequencies under increasing *x*- and *y*-axes vibration, respectively, for normal upright sitting without a back support. The study showed higher AM magnitudes at frequencies above 2 Hz and 1.5 Hz, respectively, under higher *x*- and *y*-axes vibration, while opposite trends was reported by Mansfield and Lundström (1999). Holmlund and Lundström (1998, 2001), on the other hand, showed lower MI magnitude under the *x*- and *y*-axes vibration, not only in the vicinity of the fundamental frequency, but in the entire frequency range. Fairley and Griffin (1990) showed very similar peak AM response to fore-aft vibration near the fundamental frequency of 0.7 Hz for 0.5 and 1.0 m/s^2 excitations, but Mandapuram et al. (2005) reported higher peak magnitude under 2 m/s^2 excitation. Furthermore, lower fore-aft vibration resulted in larger AM magnitude at frequencies above 1.5 Hz, as reported by Mansfield and Lundström (1999). Under the *y*-axis vibration, the effect of magnitude was more evident only at frequencies above 1.5 Hz, where the AM magnitude increased with decreasing excitation. Mandapuram et al. (2005) further showed greater increases in the fore-aft and lateral AM measured at the seat pan and the backrest with decreasing *x*- and *y*-axes excitations, respectively, for subjects seated with a back support.

The discrepancies with regards to the effect of excitation magnitude may be attributed to nonlinearity in the biodynamic responses and the coupled effects of many factors, such as sitting posture and support conditions, apart from the individual factors. A few studies, reporting vertical 'to-the-body' biodynamic responses with back and hand supports, suggest that the softening tendency of the body under increasing excitation diminishes with back support (Wang et al., 2004; Rakheja et al., 2006; Patra et al., 2008), a trend that is not evident in the responses to horizontal vibration. Figure 11(c) and (d) also show the mean absorbed power responses of subjects seated with back support under two different magnitudes of vertical vibration (Wang et al., 2006a). The results show only small differences in frequencies corresponding to peak values, while the magnitude of power has been approximately related to the square of input acceleration magnitude (Lundström et al., 1998; Mansfield and Griffin, 1998; Mansfield et al., 2001; Rakheja et al., 2008; Nawayseh and Griffin, 2010).

4.8 Responses to single- vs. multi-axis vibration

The WBV environment generally encompasses appreciable vibration along the multiple axes (Table 1). Owing to the complexities in the biodynamic responses, measurement methods and multi-axis vibration synthesisers, the biodynamic responses have been mostly studied under single-axis vibration. These have undoubtedly facilitated interpretations of the responses and contributed to greater understanding. Furthermore, the single-axis vibration along the vertical and fore-aft axes have shown considerable sagittal plane motions (vertical, fore-aft and pitch) of the upper body suggesting strong coupling effects. These are also evident from the cross-axis biodynamic forces and AM responses (Nawayseh and Griffin, 2004; Rakheja et al., 2006) and multi-axis motions of the upper body (Paddan and Griffin, 1988a, 1988b, 1993). With the developments in multi-axis vibration controllers, a few recent studies have explored the seated body responses to vibration applied simultaneously along the two- and three-translational axes in terms of direct and cross-axis AM, while the coupling effects of multiple-axes vibration on the biodynamic forces have not been examined.

Mansfield and Lundström (1999) investigated the effect of simultaneously applied *x*- and *y*-axes motion on AM and showed that the fore-aft AM response quickly adapts to the *y*-axis response when a *y*-axis motion component is introduced, suggesting relative dominance of the *y*-axis motion in the response. Qiu and Griffin (2010) found that resonance frequency in the vertical AM is reduced as the magnitude of fore-aft excitation increases, and the resonance frequency in the fore-aft AM is reduced as the magnitude of vertical vibration increases. Mansfield and Maeda (2006) concluded that fore-aft and lateral AM responses to combined *x–y* vibration were quite similar to those attained under respective single-axis vibration. The vertical AM responses to coupled *x–z* and *y–z* axes vibration were also comparable with those measured under purely *z*-axis vibration. The AM magnitude under *z*-axis vibration, however, was slightly larger than those under coupled axes vibration at frequencies above 6 Hz. The data also showed considerable magnitudes of fore-aft and vertical cross-axis AM under individual, *z*- and *x*-axes motions, similar to those presented by Nawayseh and Griffin (2004) and Rakheja et al. (2006). The results generally suggest that the direct and cross-axis AM responses to dual axis vibration occur at a slightly lower frequency compared to those to single-axis vibration. This may be in-part attributed to relatively higher resultant magnitudes of two-axis vibration compared to the single-axis vibration.

Hinz et al. (2006) in a similar manner measured the direct AM responses under individual (*x*-, *y*-, *z*-), dual (*x–y*) and three (*x–y–z*) axes vibration of three different magnitudes. In most cases, the effects of dual or three-axis vibration on the direct AM properties were small in relation to those established under single axis vibration. Moreover, definite patterns could not be established over the selected excitation magnitudes. The mean peak fore-aft and lateral AM magnitudes under individual *x*- and *y*-axis vibration were slightly higher than those measured under dual-axis (*x–y*) vibration, while a definite effect on the frequency variations was not observed. The effect of multiple axis vibration, however, was observed to be statistically significant on the vertical AM response. Peak vertical AM magnitude and the corresponding frequency were observed to be lower under three-axis excitations compared to the vertical axis alone. Mansfield and Maeda (2007) presented the direct components of the AM responses to single and three-axis vibration of equal effective magnitudes and showed that frequencies corresponding to peak magnitudes decrease under multiple axis vibration. The relatively poor frequency resolution of 0.25 Hz used in the study, however, would make it difficult to establish definite trends. The extent of coupling was found to be small and comparable to the nonlinear softening effect of the vibration magnitude.

The absence of clear coupling among the responses to multiple axis vibration is likely due to lack of correlation between vibration applied along different axes. An alternate frequency response estimator, denoted as H*v*, was proposed to study the coupling of responses to multiple axis uncorrelated vibration (Mandapuram et al., 2011), which showed notable effects of dual axis vibration compared to the single-axis vibration. Hinz et al. (2010) measured STHT under individual (*x*-, *y*-, *z*-) and three (*x–y–z*) axes vibration and found that motion pattern of the head remains nearly unchanged with the number of vibration axes. The reduction in magnitude of STHT with increasing excitation magnitude was generally observed. Mandapuram et al. (2010) suggesting weakly nonlinear effect of multi-axis vibration, and that biodynamic response to dual-axis vibration could be estimated from the direct- and cross-axis responses to single-axis vibration.

5 Biodynamic responses of the standing body to WBV

The biodynamic responses of the standing body have been generally evaluated under vertical vibration, while the responses to horizontal vibration are addressed in only a few studies (Starck et al., 1991; Paddan and Griffin, 1993). The 'through-the-body' responses have been mostly presented in terms of floor-to-head vibration transmissibility (FTHT), although a few have reported vibration transmitted to different locations of the body (Starck et al., 1991; Harazin and Grzesik, 1998; Liu et al., 1998).

Coermann (1962), Hornick (1962), and Harazin and Grzesik (1998) measured FTHT response under vertical vibration by strapping accelerometer to the head or the forehead and considering some variations in the standing posture. The majority of these have shown fundamental frequency in the vicinity of 5–6 Hz. Starck et al. (1991) observed vertical mode resonance near 4 Hz. The fundamental frequency of vibration in standing posture is comparable to the seated body. Rao (1982) also noted this similarity, while Kobayashi et al. (1981) showed lower vertical but higher fore-aft head vibration of the standing body compared to those of sitting body. Paddan and Griffin (1993) observed excessive variability with the peaks occurring at 5, 7, 11 and 16 Hz. The legs locked and unlocked postures showed similar FTHT responses, while the legs unlocked resulted in slightly lower FTHT magnitude beyond the primary resonance suggesting greater isolation by the lower limbs, and stiffening of the body, as indicated by Hornick (1962). Through measurement of vertical vibration transmission to different locations of the seated and standing body, Hagena et al. (1985) also showed greater vibration of the spine near 4 Hz for the standing posture. Matsumoto and Griffin (2000) found that the vibration transmitted to the lower spine was considerably different for the sitting posture as compared to standing posture, but similar for the head and thoracic region vibration. The standing posture resulted in larger vibration of the lumbar spine and the pelvis than the sitting posture. The variations in the fundamental frequencies of various studies are most likely attributable to the differences in the standing posture and muscle tension.

Harazin and Grzesik (1998) measured the transmission of vertical floor vibration to the metatarsus (mid-foot), ankle, knee, hip, shoulder and head through surface mounted accelerometers under 10 different standing postures, including standing on steps, bent knees, on toes, etc. The study concluded that posture did not affect the metatarsus vibration except for standing in steps, and ankle, and knee vibration except for standing on toes, while posture effect was evident from the hip, shoulder and head vibration transmissibility. The data showed greatest vibration of the metatarsus up to 8 Hz and above 20 Hz. Starck et al. (1991) measured vertical and horizontal vibration transmitted to the knee joint, hip and forehead of 10 standing subjects under horizontal vibration and observed peak horizontal vibration transmissibility below 1 Hz at these locations. It was further shown that each joint tends to amplify the magnitudes of resonant oscillations, and translate the horizontal vibration to vertical motion of the body segments. However, this pattern of vibration transmission was not observed by Liu et al. (1998) at tarsal and tibia bones, pelvis, upper torso and head under shock excitations realised through drop tests. The result of the study by Starck et al. (1991) may be due to resonance of the motion platform that revealed considerable vertical vibration. Paddan and Griffin (1993) observed strong coupling between the fore-aft, vertical and pitch motions. Under fore-aft motion, the FTHT data revealed primary resonance near 1.5, which decreased to near 1 Hz when the rail was gripped only lightly. This was attributed to stiffening of the body. The peak lateral transmissibility occurred below 3 Hz when feet separated, which was coupled with low levels of roll motion. The greatest reduction in FTHT magnitude beyond resonance was observed for bent knees posture, although the peak magnitude was the highest.

The natural frequencies of the standing body have also been identified using free vibration responses. Randall et al. (1997) measured fundamental mode frequency of the body standing on a flexible beam. A two-DOF model of the coupled beam-person system was used to estimate the uncoupled frequency of the standing body from the measured free vibration response. The experiments revealed mean natural frequencies of 12.2 Hz for male and 12.8 Hz for the female subjects, which are substantially higher than those identified from the biodynamic responses. This is most likely attributed to the shortcoming of the measurement method used.

The 'to-the-body' responses of standing subjects also exhibit similar primary frequencies and strong dependence of the body posture under vertical vibration. Coermann (1962) showed resonance frequency close to 5.9 Hz for standing erect with stiff knees. The frequency, however, decreased substantially to 2 Hz with the bent legs posture. Miwa (1975) measured MI responses in different postures and concluded the primary frequency near 6.5 Hz. The study found insignificant differences under erect or relaxed upper-body postures, when legs were held erect. In a bent-knees posture, the data showed three peaks of comparable magnitudes near 3, 20 and 60 Hz. Edwards and Lange (1964) identified resonance in the 4–5 Hz range for relaxed standing on the basis of measured MI. Matsumoto and Griffin (1998b) and Subashi et al. (2006) measured the AM responses and noted considerable reduction in the primary frequency under vertical vibration, when standing with bent-knees or on one leg, compared to the upright erect standing.

Edwards and Lange (1964) reported a decrease in resonance frequency in the MI response corresponding to relaxed standing from 5 to 4 Hz, when vibration magnitude increased from 0.2 to 0.5 g. Matsumoto and Griffin (1998b) noted this decrease from 6.75 to 5.25 Hz with vibration magnitude increasing from 0.125 to 2.0 m/s^2 rms, on the basis of measured AM corresponding to normal upright posture. Subashi et al. (2006) in a similar manner noted the softening tendency in AM response for different postures, including upright, lordotic, anterior lean and bent knees. The softening tendency was most significant with vibration magnitude increasing from 0.125 to 0.25 m/s^2, although the significance of vibration magnitude diminished for most postures when the responses to 0.25 and 0.5 m/s^2 were compared. Greater discrepancies, however, were evident on the MI or AM magnitude due to effects of vibration magnitude. Edwards and Lange (1964) noted a decrease in MI magnitude near resonance with increasing vibration magnitude, while Matsumoto and Griffin (1998b) concluded insignificant effect on the AM magnitude near resonance frequency. Subashi et al. (2006) showed decrease in peak AM magnitude with increasing excitation magnitude for the upright standing posture, and increase in the peak magnitude for the knees bent posture. The lordotic and anterior lean postures resulted in lower peak AM, which was attributed to increasing upper-body muscles activity and thus larger damping.

Matsumoto and Griffin (2000) compared the 'to-the-body' and 'through-the-body' responses of standing subjects with those of the seated body to vertical vibration.

The primary resonance frequency was greater for the standing posture than the sitting posture, while the peak AM was larger of the sitting body. The AM magnitude for the standing body, however, was greater than that of the seated body at frequencies above 7 Hz. Matsumoto and Griffin (2011) observed a principal peak in the lateral AM around 0.5 Hz. Increasing the vibration magnitude and separation of the feet resulted in decrease in both the AM magnitude and the corresponding frequency. The fore-and-aft AM showed a peak at a frequency less than 0.125 Hz.

The vertical vibration generally causes pitch motions of the upper body about the pelvis, bending of the spine and rotational motions about the ankle, as in the case of the seated body. These are evident from the appreciable fore-aft cross-axis AM reported by Subashi et al. (2006) and notable fore-aft vibration of the head, lumbar and thoracic, reported by Matsumoto and Griffin (2000). The cross-axis AM magnitude revealed peaks near the same primary frequency and near 12 Hz for the upright, lordotic and bent knees postures, which was suspected to be caused by pitch mode of the pelvis. The latter study also derived relative deflections from the measured accelerations and showed large deflections between T10–L1, L1–L3 and L3–L5 near the resonance frequencies. Stiffening of different muscles could greatly affect the rotational motions and thus fore-aft cross-axis biodynamic forces. Furthermore, the phase between the rotations about the angle, knee and the pelvis could alter the fore-aft vibration of the upper body.

6 Discussion

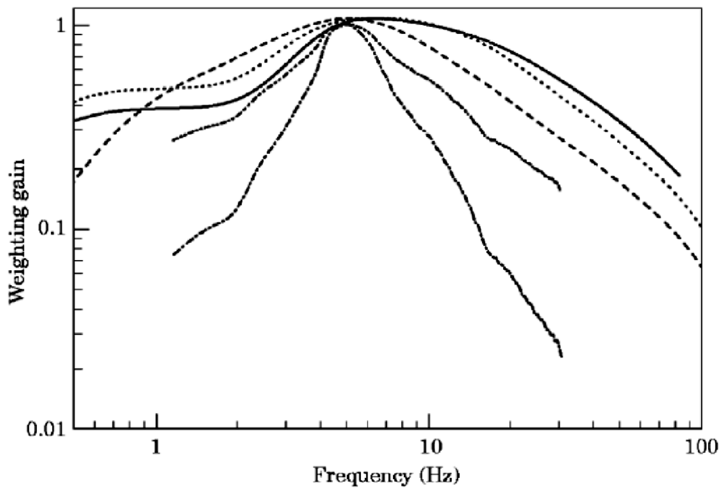
Epidemiologic studies have undoubtedly established high incidences of LBP among occupational vehicle drivers. It is, however, extremely difficult to identify the particular aspects of the driving occupation responsible for the LBP due to presence of several confounding factors. Developments in effective models of the seated and standing human body have been widely suggested in order to determine the potential effects of WBV and other posture-related stressors on the stresses induced in the spine. These would further allow for identification of dose-effect relations and to integrate the human operator dynamics in the design process. Methodical understanding and interpretations of the biodynamic responses, however, form the essential basis for building reliable models and parameter identifications. This, however, appears to be a formidable task considering the nonlinear effects of various factors. Moreover the effects of anthropometry, body supports, posture and nature of WBV are decisively coupled in a highly complex manner. The reliable target values of biodynamic responses could be established through further efforts in experimental biodynamics under constrained and well-defined conditions that would be applicable to work situations, e.g., class of vehicles. These would subsequently help the design of effective interventions, standardisation efforts and for defining frequency weightings for particular applications, which are discussed below.

6.1 Frequency weighting

The discomfort and potential injury due to WBV exposure are assessed on the basis of frequency-weighted acceleration measured at the point of entry using the standardised weightings, such as those defined in ISO-2631-1 (1997) and BS-6841 (1987). The frequency-weightings suggest higher sensitivity to horizontal and vertical WBV in the vicinity of 1 Hz and 5 Hz, respectively, which directly relate to primary resonances observed from the respective biodynamic responses. The reported measured data suggest that the resonance frequencies of the seated body tend to be considerably different when a back support is used, particularly under horizontal vibration (Figure 10). It is thus speculated that the proposed weightings may be applicable for sitting without a back support. Moreover, both the standards consider acceleration magnitude, frequency content and duration as the key variables that account for the potential injurious effects of WBV exposure. The standardised methods do not account for sitting posture and muscle tension that strongly affect the coupling of the body with the vibrating platform. Both the target biodynamic responses and the results derived from proven models could help identify effective frequency-weightings applicable for different axes of vibration and sitting conditions.

The measured 'to-the-body' biodynamic responses in terms of AM or MI and absorbed power have been applied in a few studies to determine frequency weightings. The AM responses, however, provide guidance only with regard to the critical frequencies, while the effects of WBV intensity and exposure duration cannot be evaluated. The use of absorbed power has thus been proposed to account for both the duration and intensity of the exposure, which unlike the frequency-weighted acceleration due to source vibration relates to energy absorbed or transferred to the body (Lundström et al., 1998). A convenient formula for deriving the power absorption-based frequency weighting from a biodynamic response function has been proposed (Dong et al., 2006).

Figure 13 Comparisons of frequency weighting derived on the basis of absorbed power (– · – ·) and root of power (━ · ━ ·) with W_k (·······), W_b (——) and W_g (- - - -), as reported by Mansfield and Griffin (1998)



Mansfield and Griffin (1998) applied the normalised absorbed power response of body seated upright without a back support to vertical vibration to derive a frequency weighting. Considering that the absorbed power is approximately related to square of the acceleration magnitude, a weighting based upon the square-root of the power was also deduced. The identified frequency weightings were compared with W_b defined in BS-6841 (1987), W_k in the current ISO-2631-1 (1997) and W_b in the previous version of ISO-2631-1 (1985), as seen in Figure 13. Considerable differences could be observed between the deduced and standardised weightings in the low as well as high frequency ranges. It was suggested that the weighting based upon absorbed power would permit considerably higher magnitudes of vertical vibration at higher frequencies, and it will not provide a good assessment of discomfort due to vertical vibration. Lundström et al. (1998) computed the acceleration levels corresponding to constant absorbed power with reference to W_k-weighting at 6 Hz. The computed acceleration levels were found to agree with those presented by Lee and Pradko (1968). Owing to the observed strong gender effect, the study proposed different acceleration-frequency curves on the basis of data acquired with the male and female subjects. It was also concluded that the W_k-weighting under- and overestimates the risk of WBV at frequencies below and above 6 Hz, respectively. This assertion, however, would differ if an alternate reference frequency was chosen.

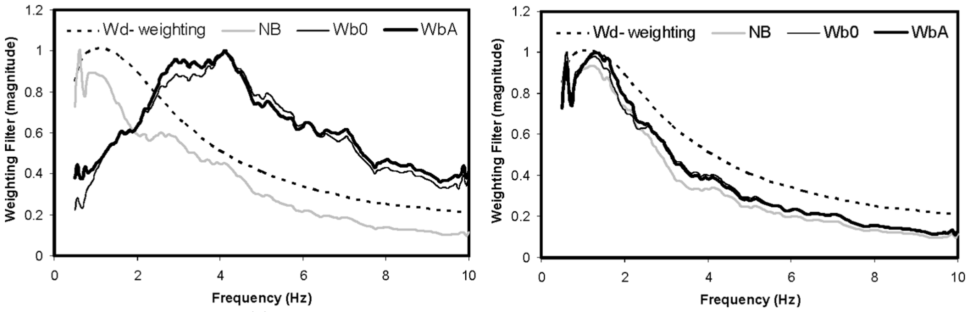
The absorbed power responses measured under *x*- and *y*-axes vibration were also applied in a similar manner to determine the acceleration levels as a function of vibration frequency corresponding to a constant power level (Lundström and Holmlund, 1998). On the basis of comparison with the W_d-weighting, defined in ISO-2631-1 (1997), it was suggested that the current weighting would underestimate the risk in the 1.5–3.0 Hz range and overestimate at frequencies above 5 Hz. While the absorbed power responses to fore-aft vibration differ considerably from those to lateral vibration, particularly, when a back support is considered, the ISO-2631-1 recommends the use of identical weighting W_d for assessing WBV along both axes. Considering the most notable effect of the back support on responses to fore-aft vibration, the application of same weighting for back unsupported and supported conditions may be questionable. Rakheja et al. (2008) proposed frequency-weightings on the basis of measured power responses to fore-aft and lateral vibration by considering three different back support conditions (None, vertical and inclined backrests). It was shown that the absorbed power under horizontal vibration is statistically related to acceleration magnitude in the following manner:

$$P_{avg} = \alpha a^{\beta}$$

where the constant α and exponent β, respectively, ranged from 1.15 to 1.39 and 1.84 to 2.07 for the fore-aft axis, and from 1.09 to 1.21 and 1.81 to 1.86 for the lateral axis, depending upon the back support condition. The frequency weighting with respect to acceleration was thus derived by taking β-root of the P_{avg} spectra, and normalised to the peak value. The resulting weightings for the *x*- and *y*-axes vibration are compared with the W_d-weighting in Figure 14. It was concluded that the weighting derived from absorbed power responses to side-to-side vibration, compares reasonably well with the W_d-weighting, although the W_d would slightly overestimate the exposure risk at frequencies above 2 Hz. The derived weighting for the fore-aft axis was also quite close to W_d-weighting corresponding to unsupported back posture, while the W_d-weighting was judged to overestimate the exposure at frequencies above 1 Hz. Considerable differences, however, were shown between the computed and W_d-weighting for the back-supported postures (Figure 14), which were attributed to greater interactions of the upper body with the back support under fore-aft motion. It was concluded that for the back supported conditions, the W_d-weighting would greatly overestimate the risk below 2.5 Hz and greatly underestimate the risk above 2.5 Hz.

The above studies suggest the need for further efforts in defining reliable biodynamic responses for defining effective frequency weightings based on both the ‘to-the-body’ and ‘through-the-body’ responses.

Figure 14 Comparisons of weighting function magnitudes derived from mean absorbed power responses corresponding to three back support conditions (NB – no back; Wb0 – vertical; and WbA – inclined) with the W_d-weighting: (a) fore-aft vibration and (b) lateral vibration



7 Conclusions

The biodynamic responses of the human body to whole body vibration, studied in terms of 'to-the-body' and 'through-the-body', have shown strong and highly complex and nonlinear effects of majority of the contributory factors, such as those related to posture, body support, anthropometry and nature of vibration. Moreover, various factors have shown coupled effects. Owing to the wide ranges of experimental conditions used in different studies, it is quite complex to clearly identify the effects of individual factors. The reported studies thus often conclude on conflicting effects of many factors. Consequently, further systematic efforts in response characterisation under representative postural and vibration conditions are vital for established a better understanding of the effects of contributory factors and effective predictions of potential injurious effects of WBV.

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