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Influence of Support Conditions on Vertical Whole-body Vibration of the Seated Human Body

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Abstract: The vibration transmission to the lumbar and thoracic segments of seated human subjects exposed to whole body vibration of a vehicular nature have been mostly characterised without the back and hand supports, which is not representative of general driving conditions. This non-invasive experimental study investigated the transmission of vertical seat vibration to selected vertebrae and the head along the vertical and fore-aft axes of twelve male human subjects seated on a rigid seat and exposed to random vertical excitation in the 0.5–20 Hz range. The measurements were performed under four different sitting postures involving combinations of back support conditions and hands positions, and three difference magnitudes of vertical vibration (0.25, 0.5 and 1.0 m/s^2 rms acceleration). The results showed significant errors induced by sensor misalignment and skin effects, which required appropriate correction methodologies. The averaged corrected responses revealed that the back support attenuates vibration in the vertical axis to all the body locations while increasing the fore-aft transmissibility at the C7 and T5. The hands position generally has a relatively smaller effect, showing some influences on the C7 and L5 vibration. Sitting without a back support resulted in very low magnitude fore-aft vibration at T5, which was substantially higher with a back support, suggestive of a probable change in the body's vibration mode. The effect of back support was observed to be very small on the horizontal vibration of the lower thoracic and lumbar regions. The results suggest that distinctly different target body-segment biodynamic functions need to be defined for different support conditions in order to represent the unique contribution of the specific support condition. These datasets may then be useful for the development of biodynamic models.

Key words: Whole-body vibration, Lumbar and thoracic vibration, Seat-to-head vibration, Back support, Hands support

Introduction

Characterising the behaviour of the seated human body exposed to whole-body vibration (WBV) has been of interest since its first applications to defence sectors[1)] to the present issues of spinal disorders and low back pain (LBP) among the operators of vibrating machinery[2, 3)]. Considering the vibrating human body to be a mechanical system, biodynamic functions such as the apparent mass (APMS) and seat to head acceleration transmissibility (STHT) have been experimentally derived[4, 5)]. Experimental studies on human subjects have been performed to extract these transfer functions by controlling a plethora of independent conditions including the type, magnitude and direction of input excitation[6)], with subjects of varying anthropometry and gender[7)], assuming different postures[8)]. A few studies have also considered the influences of the seat pan and backrest geometries on the biodynamic functions[9, 10)]. Both the APMS and STHT responses have shown a peak gain between 4 and 6 Hz for the seated human body when exposed to vibration in the vertical axis, generally considered as the primary resonance[4)]. It has also been argued that STHT may be more representative of multiple vibration modes of the upper body than the APMS driving-point response[10)].

While the two biodynamic functions described above are derived from measurements at the seat or the head in the laboratory, the majority of the vibration–related health disorders at the workplace have been noted in the lower regions of the back[2, 3)]. The vertebral units of the musculoskeletal spine composed of bony vertebrae connected by softer elements (endplates, discs, ligaments and muscles) transmit and distribute induced vibration energy from the seat through the body to the head. It is widely believed that the high incidences of LBP and spinal disorders among the vibration-exposed working population could actually be attributed to local effects in the spine[11)]. However, the movements of the spinal sub-structures may not be sufficiently reflected by the 'global' force or acceleration measurements at the seat or head alone. Additionally, target datasets based only on the APMS and/or STHT functions have proven inadequate for the development and verification of analytical bio-models capable of depicting multiple vibration modes of the human body[4, 9)]. The measurements of responses at various segments of the

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human body in the seated condition are thus crucial for better understanding of the potential mechanisms that may induce LBP. Vibration transmission characteristics of the spine have been studied through different methodologies; the conventional approach involves collection of motion data such as displacement or acceleration at selected spine and/or body locations[12]. While 'invasive' studies with sensors inserted into vertebral bones may provide more realistic data on the motion of spine segments, only a few such studies have been undertaken due to ethical reasons[13, 14]. On the other hand, 'non-invasive' measurements methods with skin-mounted sensors are shown to be significantly influenced by the skin-tissue properties[15].

The reported experimental studies on vibration transmission to different locations of the upper body have provided considerable insight into the resonance and vibration transmission behaviour of the seated body under vertical vibration, which are summarised in Table 1, where each study is identified by its lead author. The measurements reporting only STHT functions are not included in the table, since these have been extensively reviewed by Paddan and Griffin (1998)[20]. Under vertical seat excitation Panjabi *et al*. (1986)[13] observed slightly higher peaking frequencies for vertical vibration response at the sacrum as compared to that between the lumbar vertebrae (L5 to L1). It was thus hypothesised that the critical area for spinal health was at the junction between the lumbar spine and the sacrum. However, Sandover and Dupuis (1987)[18] suggested that the phenomenon of whole-body resonance may be related to bending in the lumbar spine caused by rocking of the pelvis. While this mode has also been reported in other studies[21, 22], the measured data have also revealed the presence of pelvic pitch and lumbar spine extension-compression either coupled with or independent of the spine bending modes[23]. All of the afore-mentioned studies, however, were conducted with no consideration of the back support condition. Although a typical mobile machinery driving posture may involve the use of a back support and hand controls, only a few studies have commented on such postural effects on the vibration transmitted to the spine[14, 25].

Considerable disagreements are known to exist among the reported vibration transmission data, which may be attributed to a variety of factors including differences in (a) experimental variables, namely seating conditions, posture, type and magnitude of input excitation (Table 1); (b) subject parameters such as gender, anthropometry and the number of volunteers used (Table 1); and (c) data acquisition and analyses procedures. The reported data on vibration transmission to segments of the upper body are thus generally not directly comparable due to the interplay of these influences. Furthermore, it may be inappropriate to utilise such a wide range of responses for deriving target datasets for the formulation and validation of anthropometric biodynamic models for representing multidimensional body movements[26].

In this experimental study, the vibration responses in the vertical and fore-aft axes were measured at the head and at selected vertebrae by miniature skin-mounted accelerometers on twelve male human subjects exposed to random vertical excitation, while sitting on a rigid seat under specific postural conditions. The acquired data, corrected for measurement errors arising due to sensor and skin dynamics, were used to analyse the influences of back support and hands position on the vibration transmitted to the body segments.

Methods

Subjects and experimental setup

Twelve healthy male subjects (25–39 yr of age) with no prior history of back injuries or LBP participated in this study. The standing height of the subjects ranged from 1.63 to 1.85 m (mean: 1.74 m; SD: 0.06 m), while the standing mass varied from 63 to 91 kg (mean: 74.7 kg; SD: 9 kg).

Table 1. Experimental conditions and measurement locations used in selected studies reporting transmission of vertical seat vibration to upper body segments

	Study	Subjects	Supports: Back (Hands)	Vibration Type (Magnitude*)	Frequency range (Hz)	Measurement Location on Body (Axes)
Invasive	Panjabi (1986)[13]	5	NB (Lap)	Sine (1, 3)	2–15	L1, L3, Sacrum (*x*, *z*, Pitch)
	Sandover (1987)[18]	1	NB	Sine (10 mm)	2–7	T12, L1, L2, L3, L4 (*x*, *z*)
	Pope (1989)[29]	3	NB	Impact	2–30	L3, Sacrum (*z*)
	Pope (1991)[30]	2	NB (Supported)	Random (0.5, 1, 1.5)	5, 8	L3, L4, L5 (*z*, Pitch)
	Magnusson (1993)[14]	3	B, NB (SW)	Impact (6)	0–32	L3, L4 (*x*, *z*)
	El-Khatib (1998)[25]	1, 6	B, NB	Random (1.5)	0.8–25	L1 – L5, Sternum (*z*)
Non-invasive	Donati (1983)[32]	15	NB (SW)	Random, (1.6) Sine (1.6)	1–10	Sternum (*z*)
	Hinz (1987)[28]	4	NB	Sine (1.5, 3)	2–12	Head, Shoulder, T5 (*z*)
	Hinz (1988a)[31]	3	NB	Sine (1.5, 3)	4.5, 8	Head, Shoulder, L3, L4 (*x*, *z*)
	Zimmermann (1997)[33]	30	NB	Sine (1)	4.5–16	Head, T5, Pelvis (*z*)
	Kitazaki (1998)[23]	8	NB	Random (1.7)	0.5–35	Head, T1, T6, T11, L3, Sacrum (*x*, *z*)
	Matsumoto(1998)[21]	8	NB	Random (1)	0.5–20	Head, T1, T5, T10, L1, L3, L5, Pelvis (*x*, *z*, Pitch)
	Mansfield (2002)[7]	12	NB	Random (0.25 to 2.5)	0.2–20	Upper & lower abdomen, L3, illiac crest and spine (*x*, *z*)
	Yoshimura (2005)[22]	1	NB	Random (0.7)	Up to 20 Hz	Head, C7, T1, T4, L1, L2, L3, L4, L5 (*z*)

*Magnitude in m/s^2 unless stated; NB-No back support; B-Back support; Lap-Hands in Lap; SW-Hands on steering wheel.

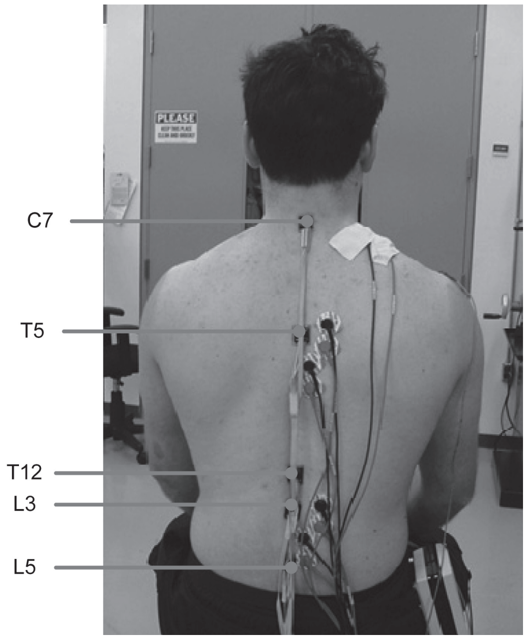


Fig. 1. A test subject prior to vibration exposure, instrumented with skin-accelerometers at the indicated vertebral levels (Note: The head strap was secured after the subject mounted the vibration platform for safety reasons).

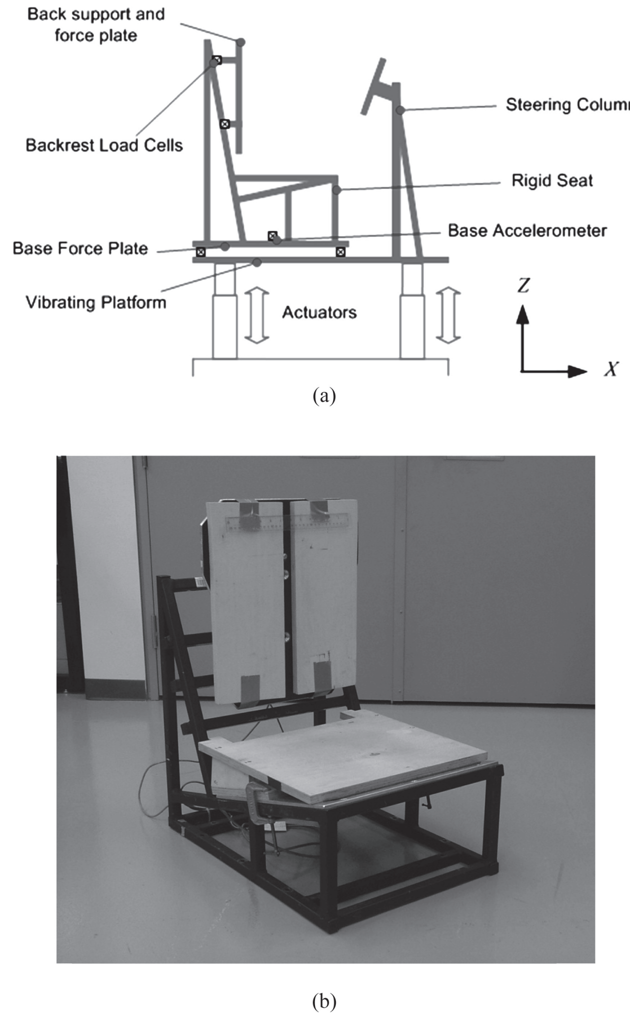


Fig. 2. (a) Schematic of the Whole Body Vertical Vibration Simulator (WBVVS) showing its components and sensors; (b) Backrest modified with a central slot to avoid trunk accelerometer adhesion in the back-supported postures.

Prior to the experiments, each subject was advised on the experiment design and safety procedures, and asked to sign a consent form that was approved by the Human Research Ethics Committee at Concordia University. Subsequently, each subject was instrumented with accelerometers that were located mid-sagittally at selected locations over the trunk and the head. The location of the accelerometers along the trunk is pictorially shown in Fig. 1. The head accelerometer (Analog Devices ADXL05 EM-3) was mounted on a plastic head-strap with a ratchet mechanism for tension adjustments around the head[10]. The total mass of the head acceleration measurement unit was 300 g. The strap was adjusted so as to orient the head accelerometer parallel to the biodynamic axis. A total of five three-axis micro-accelerometers (mass=5 g) were mounted on the skin near the seventh cervical (C7), fifth and twelfth thoracic (T5, T12), and third and fifth lumbar (L3, L5) vertebrae. The acceleration signals along the fore-aft (x) and vertical (z) axes alone were acquired, since insignificant levels of lateral acceleration would be expected for sitting subjects exposed to vertical WBV[10]. The skin location corresponding to the spinous processes of the selected vertebrae were identified by palpation. Body hair and dead tissue over the skin were removed around the chosen locations (20×20 mm) by shaving and filing, respectively. The skin was then cleaned with medical wipes (alcohol) so as to provide a relatively smooth surface for adhesion. The accelerometers (Analog Devices ADXL-330) were affixed to the cleaned locations on the skin using double-sided adhesive tape.

Figure 2(a) shows the schematic of the Whole Body Vertical Vibration Simulator (WBVVS) used in this study, which comprises of a vibration platform supported on two electro-hydraulic actuators with a maximum stoke of 250 mm (peak to peak). The closed loop vibration controller for the WBVVS was equipped with a number of safety features with programmable limits for peak displacement and acceleration. Additionally, the operator and the subject were each provided with manual emergency stop switches. A rigid seat was mounted on the platform through a force plate with four capacitive Kistler load cells to measure the dynamic force at the seat base. A single-axis accelerometer (B&K 4370) was attached to the force plate to measure vertical acceleration of the platform. The WBVVS was also equipped with a steering column fixed to the base plate to provide the hands support. The vertical seat backrest was equipped with two 445 N strain-gauge load cells (Omegadyne, LCHD-100) to measure the force developed at the body-backrest interface in the back supported postures along an axis normal to the backrest.

Initial static tests showed direct backrest contact with the miniature accelerometers fixed at the T5, T12 and L3 vertebral levels. This phenomenon was identifiable from the flat unity vertical transmissibility in addition to almost insignificant

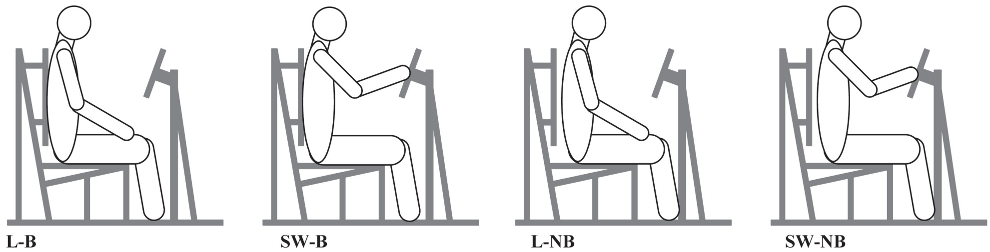


Fig. 3. Schematic of the postures assumed by test subjects in this study. (L–Hands in Lap; SW–Hands on Steering Wheel; B, NB–with and without Back contact, respectively).

horizontal responses at these trunk locations. It was imperative to have adequate back contact with the plate while avoiding accelerometer adhesion with the backrest. Subsequently, the backrest was modified by fixing two rigid wooden panels so as to form a slot for the back accelerometers to be accommodated without contacting the vibrating surface, as shown in Fig. 2(b). Subject trials showed significant differences in the vibration transmissibility responses with different slot sizes, probably due to local skin-tissue stretching. A width of 30 mm provided the required leeway for independent skin-sensor movements while ensuring sufficient back contact area.

Experiment design

The experiments were conducted under three different levels of Gaussian random vibration with nearly flat acceleration power spectral density (PSD) in the 0.5 to 20 Hz frequency range. and overall root mean square (rms) accelerations of 0.25, 0.5 and 1 m/s^2. Such vibration spectra were synthesised using a programmable vibration controller (Vibration Research: VR 8500). The subjects assumed four different postures involving combinations of two back support conditions and two hands positions: (i) back in contact with hands in lap, L-B; (ii) sitting erect with no back support and hands in lap, L-NB; (iii) back in contact with hands resting on steering wheel, SW-B; and (iv) sitting with no back support and hands on steering wheel, SW-NB. Three trials were conducted for each combination of the experimental conditions. Figure 3 depicts a schematic of the postural conditions used in this study.

Data acquisition and analyses

The measured signals were acquired in a multi-channel spectral analysis system (B&K PULSE 11.0). The data corresponding to each experimental condition were acquired for the duration of 96 s, and analysed to determine auto-spectra, cross-spectra, vibration transmissibility and their corresponding coherence functions using a 50 Hz bandwidth with a resolution of 0.0625 Hz. The data analysis corresponding to each trial involved 21 Hanning-windowed averages with an overlap of 75%. Additionally, the head and trunk acceleration signals along the fore-aft (x) and vertical (z) axes were used to derive fore-aft and vertical transmissibility from the seat base to the corresponding body location, respectively. Both the apparent mass and the vibration transmission from the seat to a particular body segment were calculated using the H_1 function involving the complex ratio of the cross-spectrum between the excitation and response, and the auto spectrum of the vertical seat acceleration, such that:

$$G_{Sx} = \frac{G_{SxA}(j\omega)}{G_{AA}(j\omega)};\ G_{Sz} = \frac{G_{SzA}(j\omega)}{G_{AA}(j\omega)} \tag{1}$$

where, $G_{Sx}(j\omega)$ and $G_{Sz}(j\omega)$ are the complex vibration transmissibility functions computed in the accelerometer's mid-sagittal coordinates; x being the axis normal to the local plane of the skin or fore-aft and z along the along plane, respectively. $G_{SxA}(j\omega)$ and $G_{SzA}(j\omega)$ are the cross-spectra of measured x- and z-axis responses of a specific segment (S) and the seat acceleration, respectively, and $G_{AA}(j\omega)$ is the auto-spectrum of the vertical seat acceleration.

Inclinations of the skin mounted accelerometers due to the contour of the spinous processes of the vertebrae or postural adjustments by the seated subjects could induce errors in measurement due to relative change in the sensor's orientation from the biodynamic axis. Sandover and Dupuis[18] suggested that the knowledge of accelerometer attitude could lead to improved accuracy. It was also shown by Dong *et al*.[19] that misalignment of an embedded accelerometer for assessing the anti-vibration properties of gloves could cause measurement errors in excess of 20%. The horizontal and vertical transmissibility responses in the basicentric biodynamic axes, $G_{SX}(j\omega)$ and $G_{SZ}(j\omega)$, respectively, may be obtained by transformation of the response-axes using the complex components of the measured x- and z-axis transmissibility functions, such that:

$$\begin{Bmatrix} G_{SX}(j\omega) \\ G_{SZ}(j\omega) \end{Bmatrix} = \begin{bmatrix} \cos\alpha & \sin\alpha \\ -\sin\alpha & \cos\alpha \end{bmatrix} \begin{Bmatrix} G_{Sx}(j\omega) \\ G_{Sz}(j\omega) \end{Bmatrix} \tag{2}$$

Where G_{SX} and G_{SZ} are the corrected acceleration transmissibility functions, while the above transformation provides an estimate of the accelerometer orientation (α) at each location, as:

$$\alpha = \tan^{-1}\begin{Bmatrix} G_{Sx}(j\omega) \\ G_{Sz}(j\omega) \end{Bmatrix} \tag{3}$$

The orientation is estimated at a low frequency of 0.5 Hz in order to ensure minimal contributions due to dynamic responses of the seated body to vertical WBV. Table 2 summarises the means of the estimated accelerometer misalignment errors for the 12 subjects at different locations on the body. Since the head position was visually monitored and rectified during the experiments, no correction was applied for this location. The results suggest considerable misalignments of the accelerometers mounted at different locations of the trunk. It should also be noted that while the accelerometers at C7 and T5 showed posterior orientation, the response data for T12, L3 and L5 required correction in the opposite sense (-α).

Table 2. Mean values of the accelerometer orientations at the measured trunk locations

Location	C7	T5	T12	L3	L5
Posture	Accelerometer Orientation (degrees)†				
L-B	35	17	–6	–6	–13
L-NB	35	18	–7	–7	–9
SW-B	34	14	–6	–7	–12
SW-NB	34	14	–6	–7	–12

†Negative values indicate posterior (backward) orientation of the sensor.

Posterior deviation from the biodynamic axis is considered positive in this study. The lower thoracic and lumbar regions show relatively smaller angles with lesser variation amongst the segments, probably because of the erect posture assumed by the subjects. It may be observed that the maximum mean deviations in sensor orientation occur in the upper torso region (C7 and T5), in excess of 25°. Except at the T5 and L5, the static misalignment for other locations is relatively small for the postures considered in this study and may produce only negligible effect on the measured responses. The SW-B posture is observed to lower static anterior rotation at the T5 level probably due to the two motion constraints provided by the steering wheel and the backrest. Moreover, there seems to be greater posterior misalignment at the L5 with no back support. The application of the correction procedure mentioned above showed significant changes in the transmissibility responses almost in the entire frequency range and yielded greater attenuation of *x*- and *z*- axes vibration around the peaking frequencies.

Correction of dynamic skin-effect

The dynamic effect due to the mechanical characteristics of the skin and certain endodermic tissue on the responses of the skin-mounted sensors has been acknowledged in many studies. Mathematical techniques have been developed for compensating the tissue response effects in both the time and the frequency domains. Most studies derive a tissue transfer function from the free response tests of the skin-accelerometer system and employ its inverse as a correction function to the measured vibration responses[16, 17]. The skin-mounted accelerometer at each location was initially pulled to a displacement of 10 mm and released to simulate a damped free-response test. The same procedure was employed for each location of the subjects in the vertical and horizontal directions. The time histories of the acceleration signals were analysed to develop the correction functions using the Mathworks-MATLAB software.

Assuming a single degree-of-freedom (DOF) system response, and that the peak magnitude occurs at the natural frequency (f_0) depicted by the measured response, the tissue damping ratio (ζ) were estimated at each trunk location for every subject using the difference in frequencies of the half-power points ($\Delta f_\pm$) on either side of the response peak extracted from the Fourier transformed outputs of the time signal, such that:

$$\zeta = \mp \frac{1-(1 \pm \Delta\beta)^2}{2(1 \pm \Delta\beta)} \qquad (4)$$

Where, $\Delta\beta$ is the frequency ratio $\Delta f_\pm / f_0$. The results suggested relatively small influence of the fore-aft skin correction in the frequency range of 0.5–20 Hz for the subjects selected for this study. Consequently, no correction was applied for the skin effect to the measured fore-aft acceleration transmissibility. Figure 4 illustrates the range and inter-quartiles of the estimated natural frequencies and damping ratios of the measured functions at different locations for the twelve subjects. The range of natural frequency seems to increase towards the lower segments of the body, being highest at L5. The median values of segment natural frequencies seem to lie between 15 and 20 Hz. The central value of the damping ratios at different locations seems to be in the range of 0.51 to 0.62. It should be noted that the vertical free response tissue natural frequency around 40 Hz reported for the L3 vertebra by Kitazaki and Griffin[17] seems to be much higher than that observed in this study. Although the reason for this difference is unclear, the weight of the accelerometer unit, the type and characteristics of the mounting adhesive, the area of skin contact, etc., may play major roles in determining the subsystem's characteristics.

Multi-factorial analyses of variance (ANOVA) were performed on the corrected vibration transmissibility data using SPSS to identify the statistical significance levels of the main factors on the horizontal and vertical vibration response magnitudes at each measurement location. The main factors included the hands support, back support and the excitation magnitude.

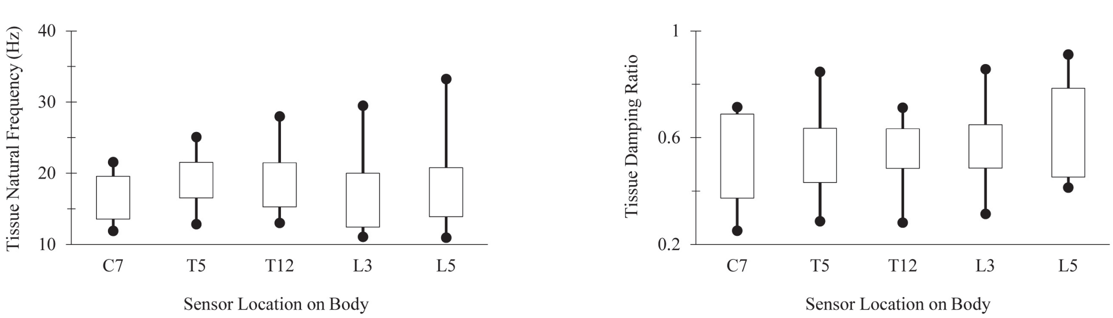


Fig. 4. Ranges for the values of skin-tissue natural frequency and damping ratio calculated for 12 test subjects through free-vibration (pull) tests on the skin-mounted accelerometers at the measurement locations.

Results

The vibration transmissibility data acquired for the subjects revealed considerable scatter in the entire frequency range, while the magnitude peaks generally occurred in a relatively narrow frequency band. As an example, Fig. 5 illustrates the magnitudes of corrected vertical, G_{SZ}, and fore-aft, G_{SX}, vibration transmissibility responses at different locations of 12 subjects for the L-NB posture and exposed to 1 m/s^2 vertical base excitation. However, the discussions in the following sections are not limited to these conditions alone. The results show consistent trends in the magnitude responses at all the locations. With the exception of the horizontal responses at the head and C7, and vertical transmissibility to T5 and L5, the results generally show relatively smaller inter-subject variability. Such dispersions are attributable to a number of contributory factors such as subject anthropometry, variations in the sitting posture and the individual's physical state. The data obtained with a few individuals showed markedly different trends from the other test subjects at some of the body segments. For example, the fore-aft responses at the C7 vertebrae of subjects 3 and 4, and the vertical transmissibility to L3 and L5 of subjects 8 and 11 differed considerably from the remaining population. Such anomalies were addressed by considering these subjects as outliers and removing the corresponding responses from the particular dataset. The mean and standard deviations of the datasets were subsequently computed and are presented in Fig. 6 for the same excitation and postural conditions depicted in Fig. 5. The figure also illustrates the mean phase curves for vertical body-segment transmissibility together with the standard deviation of the mean. The vertical phase responses showed a relatively small degree of scatter, while the coefficient of variation (CoV) was in the order of 20% below 7 Hz for all locations with the exception of the L5, where the CoV approached 40%.

The highest data scatter with CoV in the order of 40% was observed in the fore-aft response of the head for all postures in the 3 to 8 Hz frequency range. The scatter in the vertical axis data was generally less than that in the horizontal responses of the trunk segments, except at L5, which in-part may be caused by variations in the muscle tension of the subjects, apart from differences in their body build. While the change in posture strongly influenced the variability among the vertical responses of subjects only at the L3 and L5 levels, the dispersion in horizontal vibration transmissibility to all the segments was observably affected by the back support condition and in certain cases additionally by the hands position. Significantly greater inter-subject variability (maximum CoV: 50%) was obtained in the data acquired with backrest contact postures in the vertical responses at L3 and L5 between 4 and 12 Hz. In the same frequency range, the dispersion in the horizontal vertebral vibration transmission was greater with the back support. The greater variability in vertical transmissibility data is attributable to variations in the contact area of the upper body with the back support. Backrest interaction also yields an additional source of vibration to the upper body, which probably contributed to higher magnitudes of the fore-aft vibration transmissibility.

In the L-NB posture, the peak magnitudes in vertical vibration transmissibility at all body locations tend to occur in a narrow frequency band between 4 and 5 Hz (Figs. 5 and 6). A second peak is also slightly visible in the range of 7–12 Hz in vertical transmissibility of the head, T12, L3 and L5 in most of the subjects, although this peak is far more pronounced at L5, both in individual subject data as well as in the mean curves. Three of the subjects' responses revealed significantly lower magnitude of this secondary peak at the L5 level, which contributed to high dispersion of the data in this frequency range, as shown in Fig. 6. The fore-aft vibration responses of the body segments show varying trends across the measured locations. While the data presented in Figs. 5 and 6 clearly show peaks in the fore-aft vibration transmission to the head and C7 for most subjects between 5 and 6 Hz, no such characteristic is observed in the *x*-axis responses at other locations. The results further show insignificant fore-aft motion of the T5 in the entire frequency range for subjects seated without the back support. The mean fore-aft transmissibility of L5 in the NB postures shows three slight peaks around 3, 7.5 and 13 Hz. Interestingly, the mean horizontal curves for both head and L5 seem to show a clear characteristic peak at 3 Hz with no backrest.

Effects of support conditions

Figure 7 illustrates comparisons of the mean body-segment vertical and fore-aft acceleration transmissibility magnitudes corresponding to the four sitting postures, *viz*., L-B, L-NB, SW-B and SW-NB, assumed by the subjects exposed to 1 m/s^2 vertical vibration. The results clearly show that the back support has significant influence on the vibration transmission properties through the upper body. This effect is obvious in the vertical vibration transmitted to all the body-segments, while the effect on the horizontal responses measured at the lower regions of the torso, namely T12, L3 and L5, are notable only in the lower frequency range. The influence of the hands position is generally relatively small, although the effect is quite important in the fore-aft C7 and vertical L5 movements. The results show that the use of a back support tends to slightly reduce the fore-aft transmissibility to the head, while the peak horizontal responses at the C7 and T5 vertebrae increase considerably. A secondary mode around 3 Hz is also evident in the horizontal transmissibility to the head while seated assuming the L-NB posture, which seems to be slightly attenuated when the hands are supported by the steering wheel. This mode, however, is not observed in the vertical responses of all the segments and the head with back supported postures.

Interestingly, all the four postures show different fore-aft vibration tendencies at the C7 level. The back support increases the peak horizontal transmissibility magnitude at the C7 around 6 Hz, which tends to be lower with hands in lap compared to the hands on the steering wheel. An opposite effect of hands support on the fore-aft vibration at C7 is observed when the back is not supported. The hands support tends to considerably lower the peak magnitude in this case. The fore-aft response at C7 also reveals a broad secondary peak around 15 Hz, irrespective of the support condition. The results further show that the back support causes significantly higher motion at the T5 near the primary resonance frequency. Moreover, the fore-aft response at T5 exhibits a distinct resonance peak near 5.4 Hz in the back supported postures, which

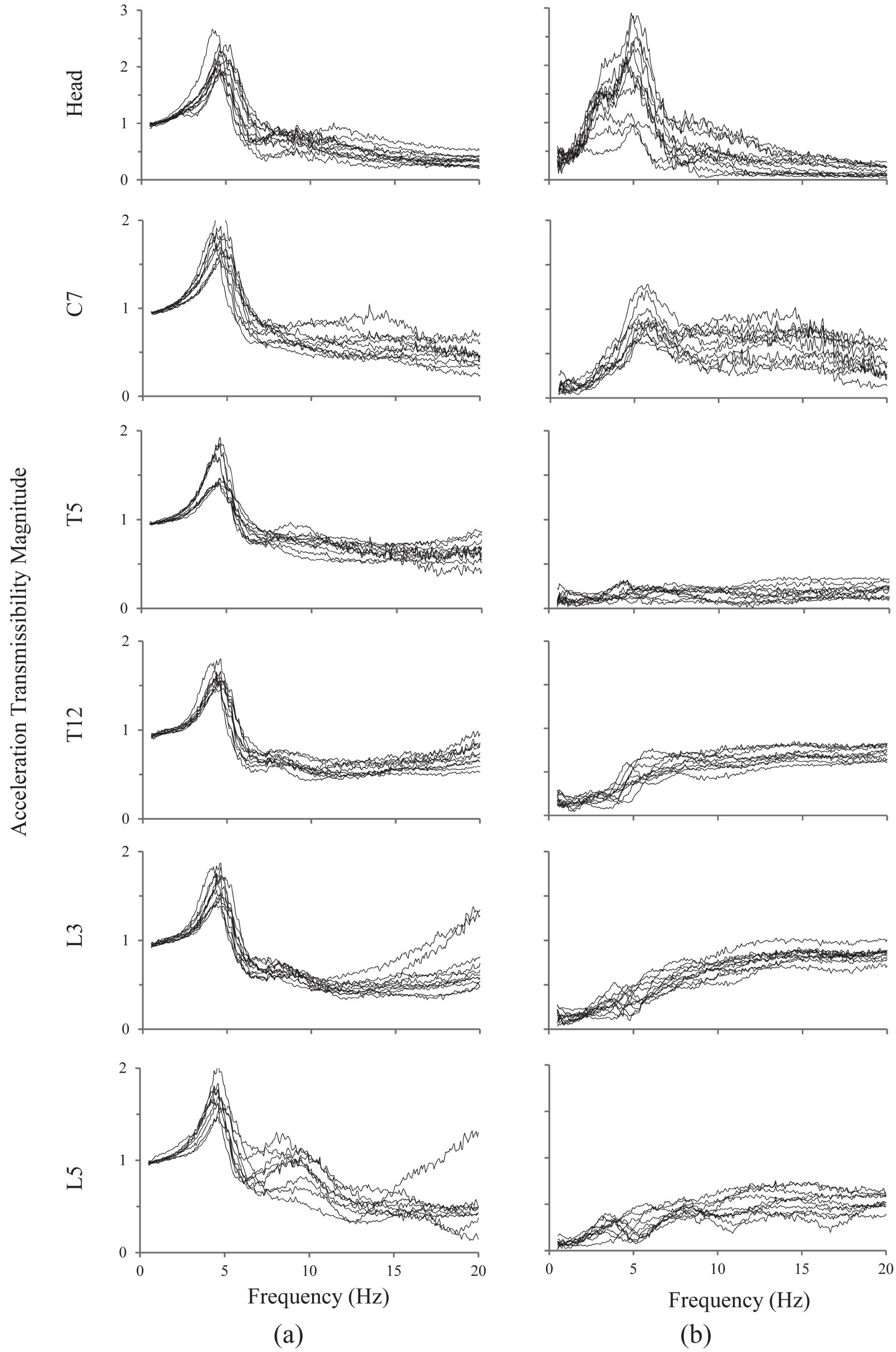


Fig. 5. (a) Vertical and (b) horizontal transmissibility magnitudes measured at different locations of 12 subjects seated in the L-NB posture and exposed to 1 m/s² (RMS) random vertical vibration.

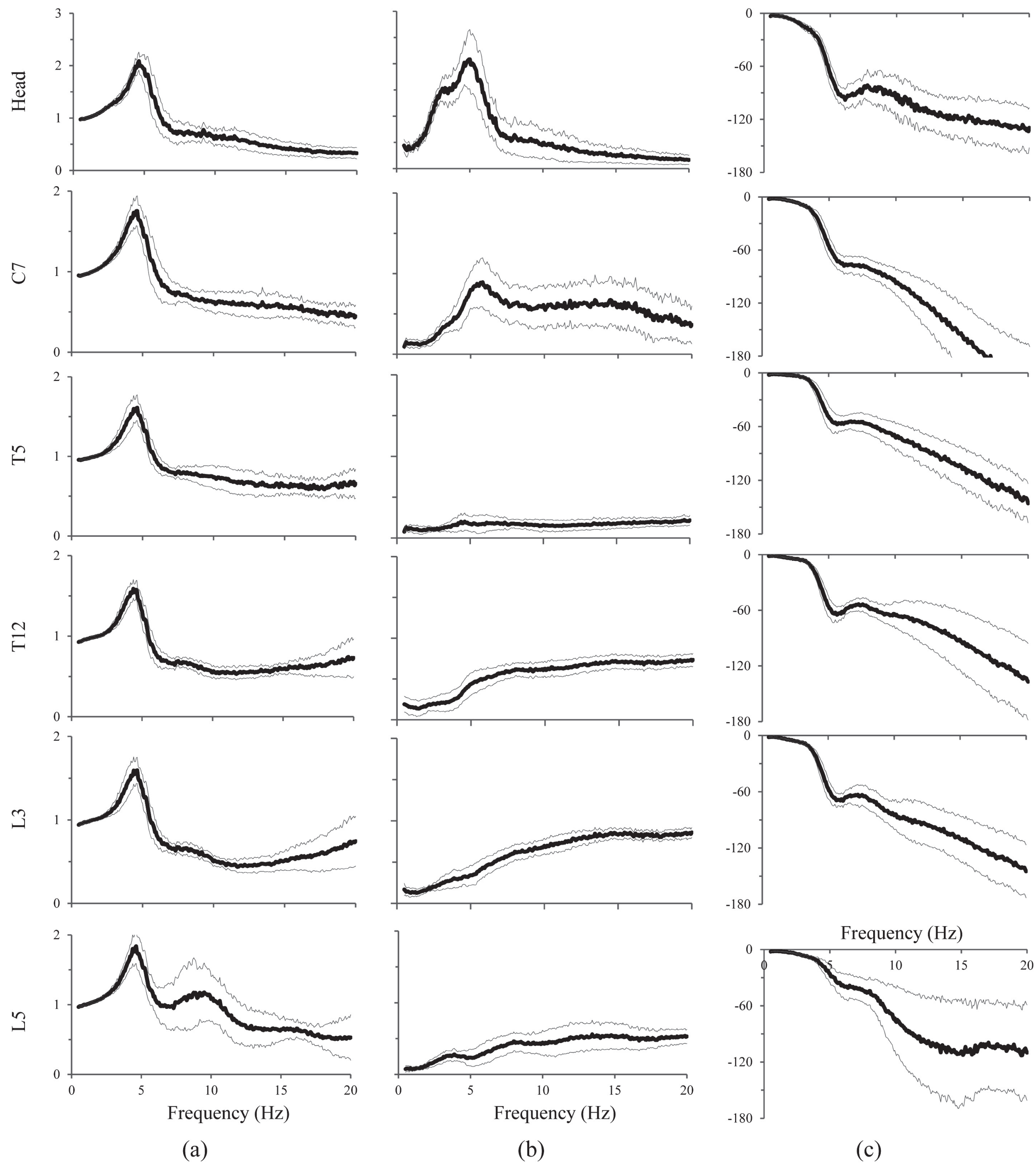


Fig. 6. Mean and standard deviation of the mean responses at the measured locations for 12 subjects seated in the L-NB posture and exposed to 1 m/s² vertical excitation: (a) vertical; and (b) horizontal transmissibility magnitude; (c) vertical transmissibility phase (degrees).

is not evident in the absence of the backrest. Such clear effects of the back support in the fore-aft responses are not observed at the T12 and L3 vertebrae, although slight changes in fore-aft motion at L5 are identifiable in the 3–8 Hz range. The horizontal responses at L5 exhibit peaks near 3 and 8 Hz when seated without a back support, while the 3 Hz peak is mostly suppressed when the vertical backrest is used. The bandwidth of the secondary peak, however, is increased with the back support in the range of 5 to 9 Hz.

The back support tends to reduce peak transmission magnitude along the vertical axis to all the segments at the primary resonance frequency around 5 Hz. However, it also introduces a dominant secondary vertical peak around 9 Hz at the head and L5, and in the 10–15 Hz range at C7. It should be noted that in the back supported postures, the C7 is not in direct contact with the backrest as shown in Fig. 3. The back

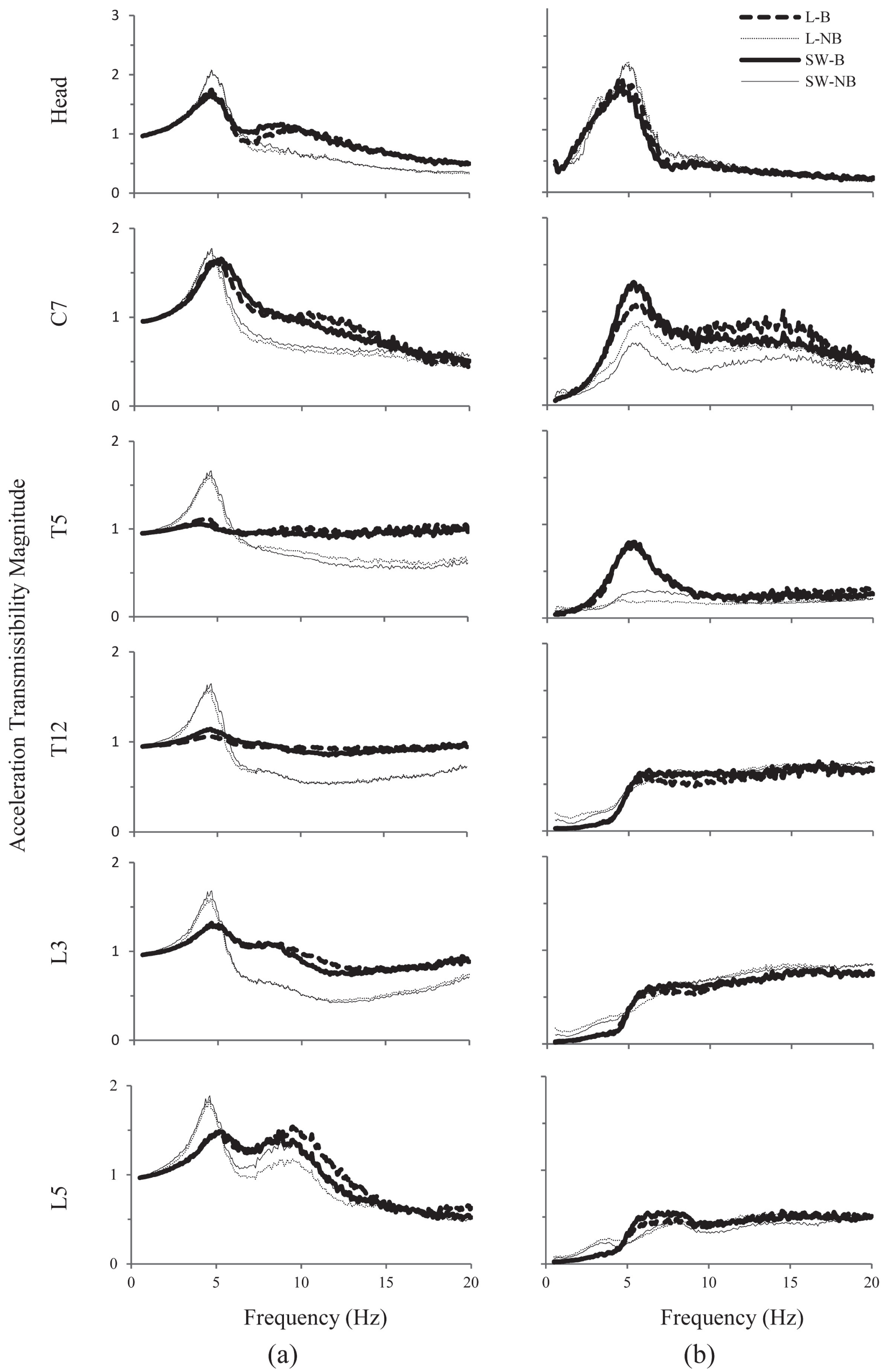


Fig. 7. Comparisons of mean responses of the body segments with different support conditions under 1 m/s² vertical excitation. (L-B: Back supported with hands in lap; SW-B: Back supported with hands on steering wheel (SW); L-NB: Hands in lap and no back support; SW-NB: Hands on SW): (a) vertical and (b) horizontal transmissibility magnitude.

support also yields higher frequency corresponding to the peak vertical responses at C7, T12, L3 and L5. The vertical vibration response at C7 exhibits considerable increase in the primary resonant frequency when the back is supported, while the peak magnitude changes only slightly. Furthermore, the vertical transmissibility peak at C7 is considerably broader with the backrest postures. Furthermore, in the L-B posture a broad peak is observed in the vertical transmissibility at C7 around 12 Hz, somewhat similar to the corresponding horizontal response in the range of 10–15 Hz. The vertical transmissibility of thoracic segments (T5 and T12) clearly depict almost complete attenuation of the peak at resonance just below 5 Hz in the back supported postures. Additionally, in the lumbar region a secondary peak, similar to that at the head, is introduced between 8 and 10 Hz with a back support. The vertical transmissibility at L5 clearly shows this peak at 10 Hz assuming equal significance to the primary resonance magnitude with body-backrest contact. The effect of the hands support, only slightly identifiable at the L3 level, is obvious in both the vertical and horizontal responses for L5 in Fig. 7. The secondary peak magnitude of vertical transmissibility to L5 is greater with the hands in lap with backrest contact. In the same response, a steering wheel hands position seems to show greater peak in the no-back posture, suggestive of all four postural conditions assuming significance at this vertebral level.

Effect of input vibration magnitude

The mean segmental transmissibility responses of 12 subjects measured with the L-NB posture under vertical vibration magnitudes of 0.25, 0.5 and 1 m/s^2 are shown in Fig. 8. The influence of input excitation is clearly identifiable in the vertical responses at all body locations and in the fore-aft axis for the head and neck (C7). The primary resonant frequency for these responses decreases with increasing vibration magnitude. Additionally, it may be noticed that for vertical responses the difference in the resonant frequencies is larger between 0.25 and 0.5 m/s^2 excitation than that observed from 0.5 to 1 m/s^2 excitation levels. This non-linear "softening" effect of the human body due to input vibration magnitude has been reported in a number of studies in both the APMS and STHT responses[10, 27]. While an increase in peak vertical transmissibility magnitude due to higher excitation levels is observable in most segments, the head and neck horizontal responses at the primary resonant frequency depict the opposite trend. Additionally, a secondary fore-aft peak in the STHT around 3 Hz is prominently identifiable at 0.25 and 0.5 m/s^2. Similarly, a higher frequency peak, between 8–13 Hz, slightly identifiable in the vertical transmissibility to T5, T12, L3 and L5, is progressively suppressed and the corresponding peak frequency decreased with increasing vibration magnitude.

Discussion

The results clearly show that the misalignments of the accelerometers, either by the mounting error or the curvature of the seated body, strongly alter the magnitude and frequency characteristics of the measured trunk transmissibility responses (C7, T5, T12, L3 and L5). Magnusson *et al.*[14] reported a maximum of 4° deviation in the pin orientation at the L3 vertebra with a vertical back support. In the present study, the postures with a backrest, *i.e.*, L-B, SW-B, show a mean shift of 7° at the L3 for the twelve subjects (Table 2). Additionally, mean transducer inclination at the C7 was observed in the order of 35°, which is comparable to the 20–35° range reported by Matsumoto and Griffin[21] at T1 for 8 male subjects seated with no backrest. It may therefore be concluded that body segment transmissibility responses need to be derived in the basicentric axes prior to any further analyses on the data. On the other hand, changes in transducer orientation may also occur due to involuntary postural adjustments made by the subjects for reasons of enhanced stability or comfort during the data acquisition, especially while sitting without a back support. Unfortunately, apart from the experimenters ensuring consistency in the subject's posture, the contributions due to such additional orientation error could not be considered. Another source of error in the biodynamic responses would be the relative movement of the skin tissue over the measured vertebral location. A single-DOF system approach, developed by Kitazaki and Griffin[17], was utilised in this study to estimate skin tissue properties. The natural frequencies of the skin tissue extracted from the measured free vibration response at different locations were in the range of 10 to 20 Hz, which are slightly higher than those reported by Hinz *et al.*[16] at the corresponding vertebrae. The median value of tissue damping ratios varied from 0.51 to 0.62, which are also slightly higher than the reported values[16, 17]. However, the skin resonant frequency values obtained at L3 were significantly lower than those reported by Kitazaki and Griffin[17]. This discrepancy may be attributed to the significant differences in the accelerometer mass and mounting techniques used in different studies. The mass of the accelerometer and its mounting employed in this study was in the order of 5 g, which is considerably smaller than those used in most of the reported studies.

In this study, greater variability between the responses of the subjects was observed around the primary resonant frequency (4–7 Hz) in both the vertical and horizontal vibration transmissibility magnitudes at most locations. Additionally, the dispersions in this frequency range were generally higher for responses in the horizontal axis (Figs. 5 and 6). The fore-aft motion at the head and C7 showed relatively higher degrees of scatter in the L-NB posture. Matsumoto and Griffin[21] under a similar posture, also reported large inter-subject variability in the horizontal axis at the head. The data scatter is also influenced by the sitting support condition; Fig. 9 shows the mean and standard deviation oft the mean horizontal response at C7 for the four sitting postures, *i.e.* L-B, L-NB, SW-B and SW-NB and under 1 m/s^2 seat excitation. The back supported sitting postures (L-B, SW-B) reveal greater deviations when compared to their corresponding counterparts with no backrest contact (L-NB, SW-NB). Similar effects were observed (but not presented here) with the backrest condition on the fore-aft responses at all other locations. While a vertical back support tends to reduce or have insignificant effect on scatter in the vertical responses, it shows the opposite trend in the horizontal axis at the thoracic and lumbar locations. Additionally, at the C7, the SW hands position is seen to reduce variability in the fore-aft axis, irrespective of the back support condition (B, NB). While the steering wheel serves as an additional source

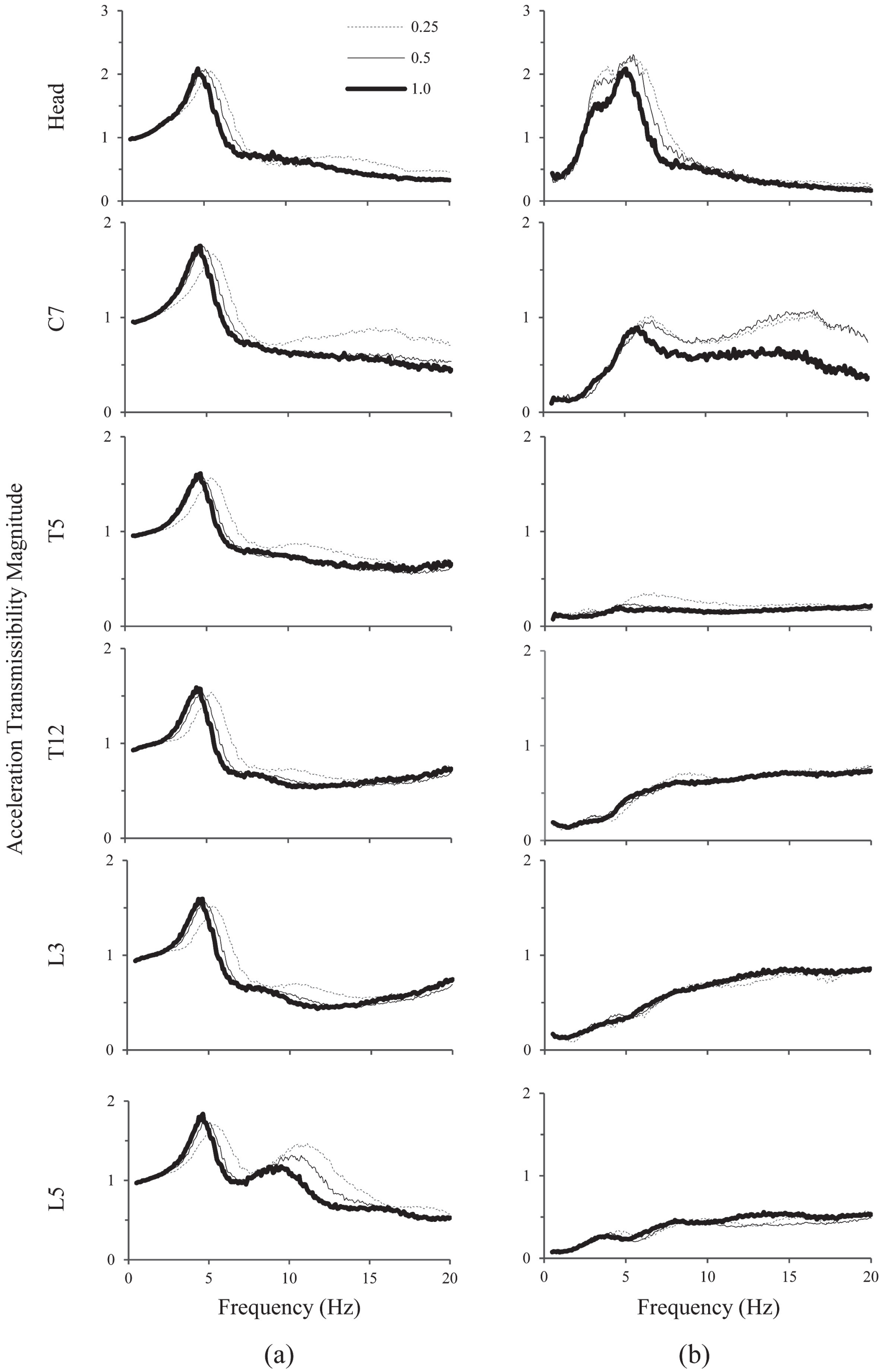


Fig. 8. Influence of vertical vibration on the (a) vertical and (b) horizontal responses of body segments corresponding to the L-NB posture. (Note: Excitation magnitude in the legend is m/s^2).

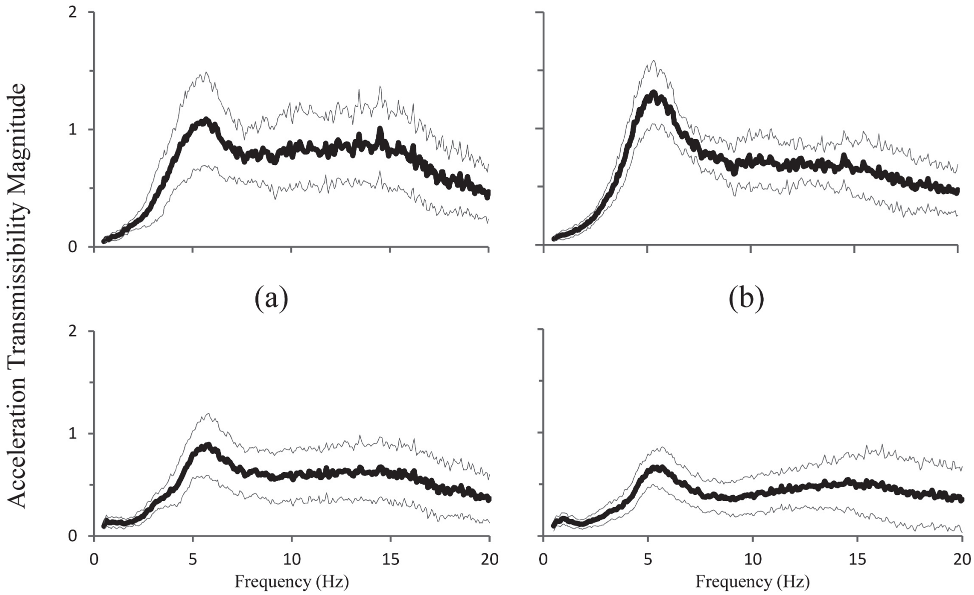


Fig. 9. Mean and standard deviation errors of horizontal responses at C7 in the four sitting postures exposed to 1 m/s^2 seat excitation: (a) L-B; (b) SW-B; (c) L-NB; and (d) SW-NB.

of vibration into the seated body, it can also be conceptualised as an additional musculoskeletal constraint at the upper thoracic region. This may elicit activity in the muscles of the hands and upper torso so as to stabilise the fore-aft motion of the body and thus reduce scatter in the horizontal response at C7.

The mean responses shown in Figs. 7 and 8 depict a clear dependence of vibration transmission properties through the body on the support condition and input vibration magnitude. Individual subject data at selected discrete frequencies were further utilised to analyse the statistical significance of these parameters. Tables 3 and 4 summarise the levels of significance in terms of '*p*' values obtained through multi–factorial ANOVA with factors including the back support condition and input excitation level, respectively, on the horizontal and vertical body segment transmissibility magnitudes. The statistical analyses suggest very strong influences of the back support on horizontal response characteristics at C7 and T5 in nearly entire frequency range ($p<0.001$), irrespective of the hands position (L and SW). Similar effects are revealed below 5 Hz for the fore-aft transmissibility of T12, L3 and L5 ($p<0.001$). This trend may also be observed as slight differences below 4 Hz for the corresponding mean magnitudes illustrated in Fig. 7. Furthermore, the back support tends to affect horizontal motion of the head between 4 and 9 Hz, and lumbar vertebrae in the range of 5 to 10 Hz ($p<0.05$). Vertical vibration transmission to all the segments seem to be significantly affected ($p<0.05$) above 2.5 Hz by the back support condition (Table 3). However, the vertical responses of the segments with hands holding the steering-wheel show greater influence with the back support, almost in the entire frequency range. In the thoracic and lumbar vertebrae, these effects may be observed in the form of vibration attenuation by the backrest (Fig. 7).

When compared to the back support condition, the hands positions seem to have relatively less influence on the vibration transmitted to body segments in both the *x*- and *z*-axes. Further, most of the existing effects of the hands support are observable with some consistency only in the postures without a back support, and primarily on the fore-aft responses. Hence, only the ANOVA results pertaining to horizontal responses in the NB posture are discussed here. The results suggest a very strong influence of hands position on the fore-aft response at C7 in the absence of a backrest ($p<0.001$). In corroboration, the mean horizontal responses of C7 (Fig. 7) reveal the differences due to the Lap and SW hands positions at almost all frequencies above 3 Hz. The T5 fore-aft transmissibility, although of small magnitude in the NB postures, seems to be slightly greater above 5 Hz with hands holding the steering wheel. ANOVA results also suggest this significant influence in the frequency range of 6 to 10 Hz ($p<0.05$). The multi-factorial design of the statistical testing techniques employed in this study analysed the combined effect of the excitation magnitude and support conditions. These demonstrated negligible levels of significance on the vibration transmission properties of the body. However, the excitation magnitude taken individually had significant influence on the response necessitating the following discussions.

The decrease in resonant frequencies ("softening") of the responses at most of the segments with increasing input vibration (Fig. 8) is further confirmed by the statistical results summarised in Table 4. Fore-aft transmissibility is strongly affected by the excitation magnitude at the head ($p<0.001$) and C7 ($p<0.05$) primarily beyond 5 Hz. However, this influence seems to be slightly greater at higher frequencies at C7 for the hands-in-lap postures ($p<0.05$). Additionally, the results also show '*p*' values less than 0.05 at 2.5 and 4 Hz for horizontal response of the head with the hands-in-lap posture, suggestive of the influence of excitation magnitude on the sec-

Table 3. ANOVA results in terms of '*p*' values* showing the influence of the backrest condition on fore-aft and vertical responses at various body locations (in both the hands positions)

		Frequency (Hz)	Head	C7	T5	T12	L3	L5
Fore-Aft Acceleration Transmissibility	Hands in Lap	1	0.387	0	0	0	0	0
		2.5	0.873	0	0.930	0	0	0
		4	**0.033**	**0.001**	0	0	0	0
		4.5	**0.044**	0	0	0	0	0
		5	**0.022**	0	0	**0.010**	**0.019**	0.188
		5.5	**0.028**	0	0	0.136	0.791	0.196
		6	**0.039**	0	0	0.779	0.050	**0.003**
		7.5	**0.022**	0	0	0.512	0.133	0
		9	**0.034**	0	0	**0.010**	**0.018**	0.146
		10	0.650	0	0	**0.048**	**0.033**	0.301
		12.5	0.375	0	0	0.265	0.604	0.067
		15	**0.004**	**0.040**	0	0.924	0.660	**0.014**
	Hands on Steering Wheel	1	0.166	0	0	0	0	0
		2.5	0	**0.002**	0.521	0	0	0
		4	0.129	0	0	0	0	0
		4.5	**0.020**	0	0	0	0	0
		5	**0.003**	0	0	**0.009**	**0.001**	0.924
		5.5	**0.004**	0	0	0.066	0.228	**0.025**
		6	**0.012**	0	0	0.281	0.612	**0.001**
		7.5	**0.015**	0	0	0.142	0.294	**0.001**
		9	**0.014**	0	0	**0.016**	0.133	0.130
		10	0.182	0	0	0.092	0.215	0.115
		12.5	0.765	0	0	0.467	0.878	0.061
		15	0.087	**0.041**	0	0.367	0.417	0.043
Vertical Acceleration Transmissibility	Hands in Lap	1	0.871	0.220	0.054	0.514	0.989	**0.007**
		2.5	0	**0.002**	0	0	**0.001**	**0.001**
		4	0.411	**0.001**	0	0	0	0
		4.5	0	0	0	0	0	0
		5	0	0.116	0	0	0	**0.002**
		5.5	**0.001**	0.095	0	0	0.382	0.377
		6	**0.031**	0	**0.001**	0.985	**0.001**	0
		7.5	0.072	0	0	0	0	0
		9	0	0	0	0	0	0
		10	0	0	0	0	0	0
		12.5	0	0	0	0	0	0
		15	0	0	0	0	0	**0.010**
	Hands on Steering Wheel	1	0.468	0.273	0.150	0	**0.028**	**0.001**
		2.5	0	**0.001**	0	0	0	0
		4	0.115	0	0	0	0	0
		4.5	0	0	0	0	0	0
		5	0	**0.005**	0	0	0	0.008
		5.5	0	0.325	0	0	0.046	0.217
		6	0	0	0	0	0.053	0
		7.5	**0.007**	0	0	0	0	0
		9	0	0	0	0	0	0
		10	0	0	0	0	0	0
		12.5	0	0	0	0	0	0
		15	0	0.404	0	0	0	**0.002**

*$p<0.05$ indicated in bold fonts.

ondary peak observed below 4 Hz in Fig. 8. Horizontal transmissibility responses at the T5, T12 and L5 also show some influence ($p<0.05$) of input vibration levels with the hands holding the steering wheel, mostly around 5 Hz. On the other hand, strong influence of vibration magnitude ($p<0.001$) is evidenced, in agreement with observed resonant frequency and magnitude shifts, in the range of 5–6 Hz for the vertical transmissibility at all the measured body locations in all the

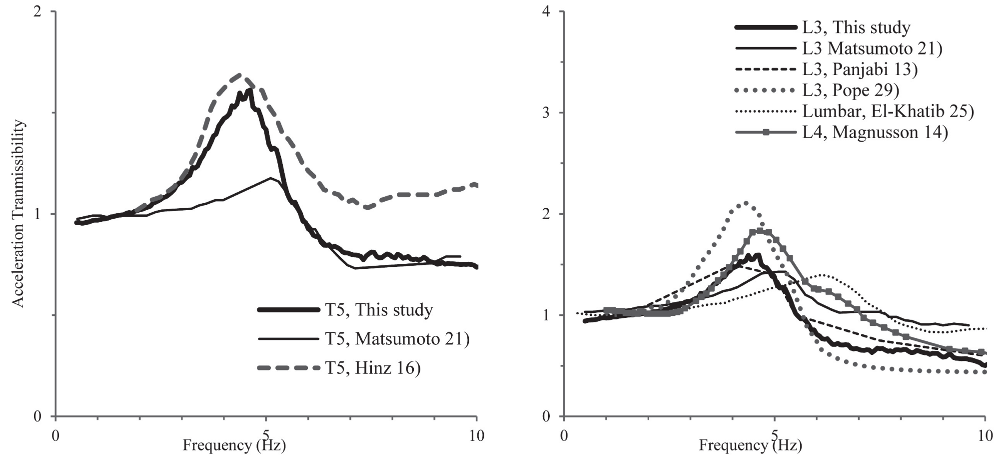
Table 4. ANOVA results in terms of '*p*' values* showing the influence of the excitation magnitude on fore-aft and vertical responses at various body locations

		Frequency (Hz)	Head	C7	T5	T12	L3	L5
Fore-Aft Acceleration Transmissibility	Hands in Lap	1	0.922	0.581	0.411	0.270	0.343	0.709
		2.5	**0.041**	0.197	0.570	0.520	0.091	0.871
		4	**0.016**	0.176	**0.028**	0.480	0.081	0.541
		4.5	0.361	0.348	0.321	0.331	0.639	0.102
		5	0.569	0.760	0.662	**0.036**	0.063	0.222
		5.5	**0.006**	0.175	0.119	0.057	**0.018**	0.108
		6	0	**0.005**	0	0.718	0.249	0.092
		7.5	0	0	0	0.582	0.978	0.893
		9	0.290	**0.005**	0	0.263	0.908	0.157
		10	0.653	**0.015**	0	0.638	0.774	0.204
		12.5	0.220	**0.010**	**0.002**	0.393	0.216	0.355
		15	**0.010**	**0.002**	0	0.177	0.201	0.253
	Hands on Steering Wheel	1	0.740	0.774	0.984	0.417	0.858	**0.017**
		2.5	0.561	0.304	0.122	0.198	0.566	0.145
		4	0.604	**0.015**	0	0.617	0.603	0.464
		4.5	0.672	**0.018**	0	0.094	0.287	**0.004**
		5	0.877	0.119	**0.039**	**0.002**	**0.001**	0.070
		5.5	**0.003**	0.103	0.295	**0.003**	0	**0.036**
		6	0	**0.005**	**0.006**	0.407	**0.046**	0.437
		7.5	0	0	**0.001**	0.951	0.730	0.439
		9	0.350	0.002	0	0.820	0.993	**0.003**
		10	0.659	0.009	0	0.857	0.943	**0.010**
		12.5	0.071	0.139	0	0.597	0.496	0.988
		15	**0.005**	0.030	**0.001**	0.312	0.222	0.651
Vertical Acceleration Transmissibility	Hands in Lap	1	**0.001**	0.076	0.541	0.872	0.550	0.816
		2.5	**0.034**	**0.001**	**0.008**	0.221	0.122	0.087
		4	**0.006**	0	**0.001**	0	0	0
		4.5	**0.001**	0	0.064	**0.004**	**0.010**	**0.002**
		5	0.528	0.167	0.418	0.069	0.735	0.453
		5.5	0	0	0	0	0	**0.008**
		6	0	0	0	0	0	0
		7.5	0.749	0.203	**0.001**	**0.048**	0.065	0.813
		9	0.211	0.992	0.098	**0.002**	**0.015**	0.699
		10	0.649	0.830	0.072	0	0	0.330
		12.5	0	0.022	**0.002**	0.165	**0.001**	0
		15	0	0	0.111	0.785	0.599	**0.032**
	Hands on Steering Wheel	1	0.139	0.101	0.366	0.714	0.298	**0.019**
		2.5	0.116	**0.005**	0.092	0.174	0.141	**0.004**
		4	**0.020**	0	**0.008**	0	**0.001**	0
		4.5	0.067	0	0.151	**0.007**	**0.010**	0
		5	0.955	**0.043**	0.226	0.716	0.602	**0.006**
		5.5	0	**0.001**	0	0	**0.001**	**0.025**
		6	0	0	0	0	0	**0.004**
		7.5	0.810	0.260	**0.003**	**0.015**	0.070	0.889
		9	0.173	0.483	**0.029**	**0.001**	**0.003**	0.356
		10	**0.010**	0.739	**0.025**	0	0	0.149
		12.5	**0.001**	0.994	**0.017**	0.440	**0.002**	0
		15	0	0.910	0.128	0.924	0.705	**0.003**

*$p<0.05$ indicated in bold fonts.

postures. Furthermore, the observed effects on the vertical transmissibility peak of the lower thoracic and lumbar vertebrae (Fig. 8) may be found to correlate well with results of ANOVA ($p<0.001$) at 10 Hz for T5 and L3, and 12.5 Hz for T12 (Table 4).

The results suggest clear influence of both the back support and input excitation magnitude on the bi-dimensional motion of the upper body. The hands position, however, seems to exhibit discernible effects mostly in the fore-aft axis at the C7 and vertical response at L5, and especially in the absence of



(a) (b)

Fig. 10. Comparison of mean measured vertical responses at T5 and L3 with the reported data on vibration transmitted to the spine in the (a) thoracis, T5; and (b) lumbar region, L3 and L4.

a back support. Further, while the vibration magnitude may primarily affect the resonance frequencies with some influence on the peak magnitude, the back support remarkably alters the vibration transmission properties through the body. It may be hypothesised with a comfortable level of confidence that in order of importance for the understanding of the body movements under vertical vibration, the back support condition assumes prime significance followed by the excitation magnitude and hands position.

In comparison with the APMS and vertical STHT, only a few studies have reported vibration transmission to the spine. In addition, most of these have been performed with subjects sitting in an erect posture without a backrest. However, the few experiments that included some form of a back support seem to show conflicting results. For example, while Magnusson *et al*.[14] report almost no change in vertical responses of the L3 due to a backrest, El-Khatib *et al*.[25] showed significant contributions in the lumbar region around the resonance, partly due to a lumbar support. Thus, owing to the lack of sufficient published data in similar postures, the reported segmental vibration response to vertical seat excitation with subjects sitting without a back support are compared in Figs. 10(a) and (b). These also compare the mean vertical response measured at the T5 and L3, respectively, for the 12 subjects in this study under 1 m/s^2 vibration in the L-NB posture. It should be noted that while all the response data presented in Fig. 10 were acquired with no backrest interaction, some reported experiments may have been performed with different excitation parameters, hands positions and/or subject mass (Table 1). Irrespective of these differences, the majority of the results indicate vertical resonance of the thoracic and lumbar regions of the spine in the narrow range of 4–5.5 Hz. There are observable differences both in resonant frequency and the peak magnitude between the present study and the data reported by Matsumoto and Griffin[21]. However, there is acceptable agreement below 4.5 Hz, in the T5 response measured in this study with that of Hinz *et al*[28]. In Fig. 10(b), illustrating the lumbar transmissibility, four of the six datasets presented, including this study show peak magnitudes around 1.5. Furthermore, the measured L3 response shows good agreement with the invasive measurements of Panjabi *et al*.[13] and a good match in resonant frequency with Pope *et al*.[29] even though the latter study was performed with female subjects sitting on cushion seats and exposed to impacts.

It is evident from this study that further experimental efforts are needed to obtain sufficient numbers of comparable datasets so as to confidently characterise the multi-dimensional motion of the seated human body exposed to vibration. Furthermore, owing to significant influences of the hands and back support conditions on the vibration transmission properties of the upper body, it may also be concluded that separate sets of segmental biodynamic functions need to be extracted for different postural conditions so as to represent the unique contribution of the specific independent parameters. The datasets thus obtained may then be utilised as target functions for the development and validation of anthropometric bio-models for simulation and virtual testing.

Conclusions

The transmission of vertical seat vibration to the head and selected vertebral locations on the back of 12 seated male humans were measured non-invasively. The experiments involved four different sitting postures realised through combinations of two back support conditions, two hands positions, and three different magnitudes of random vertical vibration. The results clearly showed significant contributions due to visco-elastic properties of the skin and misalignments of the skin-mounted accelerometers with respect to the basicentric coordinate system. The application of specific mathematical correction procedures is thus vital for extracting the segmental vibration transmission responses. The body-segment transmissibility responses of the subjects depicted a clear dependence on the support conditions, particularly the back support. Backrest contact resulted in greater attenuation of vertical vibration to all the measurement locations, while increasing the fore-aft transmissibility at C7 and T5. The hands position generally showed a relatively smaller effect, while the hand 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. The effect of input vibration magnitude was also significant but relatively weaker than that of the back support. The results suggest most important influence of a back support on the vibration transmission followed by the excitation magnitude and the hands support. It is also evident that separate sets of segmental biodynamic functions need to be extracted for different postural conditions so as to represent the unique contribution of the specific independent parameters, and to identify target functions for the development and validation of anthropometric bio-models for simulation and virtual testing.

References

1) Coermann R (1962) The mechanical impedance of the human body in sitting and standing position at low frequencies. Hum Factors **4**, 227–53.
2) Hoy J, Mubarak N, Nelson S, Sweerts de Landas M, Magnusson M, Okunribido O, Pope M (2005) Whole-body vibration and posture as risk factors for low back pain among forklift truck drivers. J Sound Vib **284**, 933–46.
3) Bovenzi M, Pinto I, Stacchini N (2002) Low back pain in port machinery operators. J Sound Vib **253**, 3–20.
4) Fairley T, Griffin M (1989) The apparent mass of the seated human body: vertical vibration. J Biomech **22**, 81–94.
5) Paddan G, Griffin M (1988) The transmission of translational seat vibration to the head–I. Vertical seat vibration. J Biomech **21**, 191–7.
6) Mansfield N, Maeda S (2005) Comparison of the apparent mass of the seated human measured using random and sinusoidal vibration. Ind Health **43**, 233–40.
7) Mansfield N, Griffin M (2002) Effects of posture and vibration magnitude on apparent mass and pelvis rotation during exposure to whole-body vertical vibration. J Sound Vib **253**, 93–107.
8) Rakheja S, Stiharu I, Boileau P-É (2002) Seated occupant apparent mass characteristics under automotive postures and vertical vibration. J Sound Vib **253**, 57–75.
9) Rakheja S, Stiharu I, Zhang H, Boileau P-É (2006) Seated occupant interactions with seat backrest and pan, and biodynamic responses under vertical vibration. J Sound Vib **298**, 651–71.
10) Wang W, Rakheja S, Boileau P-É (2006) Effects of back support condition on seat to head transmissibilities of seated occupants under vertical vibration. J Low Freq Noise Vib **25**, 239–59.
11) Wilder D, Pope M (1996) Epidemiological and aetiological aspects of low back pain in vibration environments – an update. Clin Biomech **11**, 61–73.
12) Sandover J (1998) The fatigue approach to vibration and health: Is it a practical and viable way of predicting the effects on people? J Sound Vib **215**, 699–721.
13) Panjabi M, Anderson G, Jorneus L, Hult E, Mattson L (1986) *In vivo* measurement of spinal column vibrations. J Bone Joint Surgery **68-A**, 695–702.
14) Magnusson M, Pope M, Rostedt M, Hansson T (1993) Effect of backrest inclination on the transmission of vertical vibrations through the lumbar spine. Clin Biomech **8**, 5–12.
15) Lafortune M, Henning E, Valiant G (1995) Tibial shock measured with bone and skin mounted transducers. J Biomech **28**, 989–93.
16) Hinz B, Seidel H, Dieter B, Menzel G, Blüthner R, Erdmann U (1988b) Examination of spinal column vibrations: a non-invasive approach. J App Physiology **57**, 707–13.
17) Kitazaki S, Griffin M (1995) A data correction method for surface measurement of vibration on the human body. J Biomech **28**, 885–90.
18) Sandover J, Dupuis H (1987) A reanalysis of spinal motion during vibration. Ergonomics **30**, 975–85.
19) Dong R, Rakheja S, Smutz W, Schopper A, Welcome D, Wu J (2002) Effectiveness of a new methods (TEAT) to assess vibration transmissibility of gloves. Int J Ind Ergon **30**, 33–48.
20) Paddan G, Griffin M (1998) A review of the transmission of translation seat vibration to the head. J Sound Vib **215**, 863–82.
21) Matsumoto Y, Griffin M (1998) Movement of the upper body of seated subjects to vertical whole body vibration at the principal resonance frequency. J Sound Vib **215**, 743–62.
22) Yoshimura T, Nakai K, Tamaoki G (2005) Multi-body dynamics modelling of seated human body under exposure to whole-body vibration. Ind Health **43**, 441–7.
23) Kitazaki S, Griffin M (1998) Resonance behaviour of the seated human body and effects of posture. J Biomech **31**, 143–9.
24) Mandapuram S, Rakheja S, Ma S, DeMont R, Boileau P-É (2005) Influence of back support condition on the apparent mass of seated occupants under horizontal vibration. Ind Health **43**, 421–35.
25) El-Khatib A, Guillon F, Domont A (1998) Vertical vibration transmission through the lumbar spine of the seated subject–first results. J Sound Vib **215**, 763–73.
26) ISO 5982 (2001) Mechanical vibration and shock – Range of idealized values to characterize seated-body biodynamic response under vertical vibration. International Organization for Standardization, Geneva.
27) Mansfield N (2005) Impedance methods (apparent mass, driving point mechanical impedance and absorbed power) for assessment of the biomechanical response of the seated person to whole-body vibration. Ind Health **43**, 378–89.
28) Hinz B, Seidel H (1987) The nonlinearity of the human body's dynamic response during sinusoidal whole body vibration. Ind Health **25**, 169–81.
29) Pope M, Broman H, Hansson T (1989) The dynamic response of subject seated on various cushions. Ergonomics **32**, 1155–66.
30) Pope M, Kaigle A, Magnusson M, Broman H, Hansson T (1991) Intervertebral motion during vibration. Proc Inst Mech Eng [H] **205**, 39–44.
31) Hinz B, Seidel H, Bräuer R, Menzel G, Blüthner R, Erdmann U (1988a) Bidimensional accelerations of lumbar vertebrae and estimation of internal spinal load during sinusoidal vertical whole-body vibration: a pilot study. Clin Biomech **3**, 241–8.
32) Donati P, Bonthoux C (1983) Biodynamic response of the human body in the sitting position when subjected to vertical vibration. J Sound Vib **90**, 423–42.
33) Zimmermann C, Cook T (1997) Effects of vibration frequency and postural changes on human responses to seated whole-body vibration exposure. Int Arch Occup Environ Health **69**, 165–79.