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Chris L. Zimmermann · Thomas M. Cook

Effects of vibration frequency and postural changes on human responses to seated whole-body vibration exposure

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Abstract The present investigation evaluated the effects of changes in pelvic orientation and vibration frequency on the seated human's response to whole-body vibration (WBV). Seat-to-trunk and seat-to-head acceleration transmissibility, peak-to-peak pelvic motion and erector spinae EMG and mean erector spinae EMG was collected across three pelvic orientations (9° anterior pelvic tilt, neutral pelvis, and 9° posterior pelvic tilt) and frequencies ranging from 4.5 to 16 Hz. Subjects included 30 healthy males between the ages of 18 and 35. Ensemble averages, two vibration cycles in length, were produced for each subject within each frequency–pelvic orientation combination. Group ensemble averages within each frequency–pelvic orientation combination were then compared using ANOVA. Changes in pelvic orientation produced significant differences in acceleration transmissibility, pelvic motion, and erector spinae EMG. At frequencies below 6 Hz, acceleration transmissibility at the head and pelvic motion were significantly greater in the posterior pelvic orientation than in the other two. At frequencies above 6 Hz, acceleration transmissibility at the head and trunk were significantly greater in the anterior pelvic orientation than in the other two. Peak-to-peak EMG responses were similar across all pelvic orientations at frequencies below 6 Hz. However, above 6 Hz, the response was significantly greater in the anterior pelvic orientation than in the other two. Thus, vibration frequency and pelvic orientation were shown to have significant interactive effects on the seated human's response to WBV. These interactive effects need to be considered when determining appropriate vibration exposure guidelines.

Key words Vibration · Transmissibility · Low back pain · Sitting · Electromyography

C. L. Zimmermann (✉)
The University of Iowa, Physical Therapy Graduate Program, 2600 Steindler Building, Iowa City, IA 52242-5000, USA.
Tel. (office): (319)-335-4531; fax: (319)-335-9709
E-mail: chris-zimmermann@uiowa.edu

T. M. Cook
The University of Iowa, Department of Preventive Medicine, Injury Prevention Research Center, 158 AMRF, Iowa City, IA 52242, USA

Introduction

The prevalence of low back pain (LBP) is near 70% in some industrialized societies (Frymoyer et al. 1983) and is costing billions of dollars annually (Cats-Baril and Frymoyer 1991). Several investigators have implicated static sitting as a major reason for LBP's prevalence (Cyriax 1975; Keegan 1953; McKenzie 1981). The importance of posture during static sitting and its effects on intradiscal pressure (Nachemson 1966) and erector spinae electromyographic activity have been investigated (Andersson et al. 1974a, 1974b, 1975). The effects of postural changes have also been evaluated during static sitting in a driver's seat (Andersson et al. 1974b). The static sitting literature indicates a significant influence of seated posture on intradiscal pressure and muscular demands during static sitting.

The WBV epidemiological literature has quantified that sitting in a driving or vibratory environment produces even greater risk of LBP or injury than static sitting (Bongers and Boshuizen 1990; Damkot et al. 1984; Dupuis and Zerlett 1987; Frymoyer et al. 1983; Heliovaara 1987; Hulshof and Veldhuijzen van Zanten 1987; Kelsey 1975; Kelsey and Hardy 1975; Spear et al. 1976). Unfortunately, literature on the human response to seated whole body vibration (WBV) exposure is incomplete regarding the effects of changes in seated posture and the erector spinae's response to seated WBV exposure. These factors and their effects on the proposed mechanisms responsible for WBV exposure's associated increased risk of LBP (Sandover 1981) continue to need investigation.

A number of studies have quantified the human's mechanical response to WBV in terms of transmissibility (Cohen et al. 1977; Dieckmann 1957; Garg and Ross 1976; Griffin 1975; Messenger and Griffin 1989; Paddan and Griffin 1988; Park 1991; Rowlands 1977; Sandover 1982; Wilder et al. 1982) and impedance (Coermann et al. 1960; Dieckmann 1957; Miwa 1975; Sandover 1982; Weis et al. 1964; Wilder et al. 1982), yet these studies seldom take into account the trunk musculature's role of postural maintenance (Asmussen 1960; Floyd and Silver 1955; Portnoy and Morin 1956). However, sitting is a dynamic activity (Branton 1969), and this implies changing demands on the musculoskeletal system as seated postures are changed. Some investigations have addressed the effects of changes in seated posture on transmissibility (Griffin et al. 1979; Messenger and Griffin 1989; Part 1991; Sandover 1982; Wilder et al. 1982, 1993) and impedance (Griffin 1975; Park 1991; Sandover 1982; Wilder et al. 1982, 1993), while others have investigated the muscular response to vibration (Robertson and Griffin 1989; Seidel et al. 1986; Seroussi et al. 1989; Wilder et al. 1982; Zimmermann et al. 1993). However, these muscular response investigations have limited application to the adoption of different postures in the seated WBV environment.

The response of the musculature during WBV exposure has been classified into two separate categories, tonic and phasic activities [sometimes referred to as a vibration synchronous response (VSR)]. Regarding erector spinae activity during WBV exposure, there are conflicting reports of increases in tonic electromyographic (EMG) activity with (Zimmermann et al. 1993) and without (Robertson and Griffin 1989) accompanying increases in the phasic EMG component. However, if the VSR is reflexive in nature, VSR magnitude increases would be expected to be positively associated with mean EMG increases.

Several other discrepancies and un-investigated theories exist within the WBV literature. Acceleration transmissibility investigations have reported differences in results between subjects, which could not be accounted for and have thus been attributed to subtle changes in posture and "pelvic rocking," that is to say movement of the pelvis over the ischial tuberosities relative to the seat pan (Robertson and Griffin 1989; Sandover 1981, 1982). Pelvic rocking has also been proposed as a possible mechanism for the seated human's primary resonance at 4–6 Hz (Sandover 1982).

There is also a dearth of information attempting to associate the muscular and mechanical responses generated during seated WBV exposure (Park 1991; Seidel et al. 1986). If changes in seated posture are accompanied by changes in mean and VSR EMG responses, how do these relate to changes in the seated human's mechanical response? In order to formulate the most appropriate WBV exposure guidelines and update ISO 2631 (International Organization for Standardization 1985), further investigation is required. Therefore, the present study was undertaken to increase our understanding of the effects of changes in seated posture and vibration frequency on the muscular and mechanical aspects of the human response to seated WBV.

Methods

The subject population was made up of 30 healthy males between 20 and 35 years of age. Descriptive characteristics (mean and SD) include age 26.5 years (2.9 years), height 179.5 cm (7.5 cm), and weight 77.6 kg (10.7 kg). All subjects were verbally screened for any history of neck or back problems. Persons reporting a history of structural injuries or deformities, or current neck or low back pain, were excluded from the study. All subjects read and signed informed consent forms in compliance with The University of Iowa Human Subjects Committee guidelines.

Whole-body vibration was produced using a Ling Dynamics Systems PA2000 amplifier and Model V-716 exciter controlled by a Model DSC 8MK II digital sine controller (Ling Dynamic Systems, Yalesville, Conn.). Subjects sat on a rigid wooden seat pan (without padding) mounted directly to the vibrator surface with their feet supported on a free-standing adjustable foot rest. A strain gauge accelerometer mounted on the seat pan monitored vertical seat pan acceleration.

Consistency and stability of the subject's head orientation during testing were facilitated by means of visual feedback from two adjustable markers, aligned in the horizontal plane, attached to an adjustable headband's right side and viewed in a mirror. An electrolytic tilt sensor (Spectron CG-57s, Spectron Glass and Electronics, Hauppauge, N.Y.) attached to a hinge riveted to an aluminum strip and positioned on a figure-of-eight harness just above the fifth thoracic vertebra (T5) monitored the subject's trunk position. An analog meter positioned in front of the subject displayed the sensor's output (Fig. 1).

An accelerometer (Endevco 7290A-10, Endevco, San Juan Capistrano, Calif.) mounted on the adjustable headband and positioned in a vertical line with the left ear (Sandover 1982) monitored vertical or *Z*-axis (ISO 2631) head acceleration (Fig. 1). Another accelerometer (Endevco 7290A-10) attached to a hinge riveted to an aluminum strip and positioned on the figure-of-eight harness just below the T5 level (Fig. 1) monitored vertical trunk acceleration. The mass of the tilt sensor, trunk accelerometer, hinges and attachment plate was 47 g. A customized dual potentiometer (Spectrol 140-0-0-203, Spectrol Electronics, PO Box 1220, City of Industry, Calif.) system, allowing tracking of pelvic inclination relative to the seat pan, monitored the angular position and movements of the pelvis (Fig. 1). The potentiometer and attachment plate had a combined mass of 16.5 g.

Surface EMG signals were collected using electrodes containing circuitry for on-site amplification with a gain of 35 (Therapeutics Unlimited, Iowa City, Iowa). The electrodes, when coupled with the differential main amplifier (GCS 67, Therapeutics Unlimited), allowed a gain of 500–10,000 with a band width of 40 Hz to 4 kHz. On-line data sampling, analysis, and collection were performed using a microcomputer (Gateway 4DX2-66V, Gateway 2000, N. Sioux City, S.D.) and customized software.

All subjects were exposed to 1 m/s^2 r.m.s. sinusoidal vibration at frequencies of 4.5, 5, 6, 8, 10, 12, and 16 Hz. Anterior, neutral, and posterior pelvic orientations were evaluated in the study (Fig. 2). Each of the pelvic orientations was obtained and maintained (verbal feedback from investigator) while the subject kept his head level and trunk upright (visual feedback system described earlier). The neutral pelvic orientation (NPO) was self-selected by the subject following the command "Sit up straight". The anterior pelvic orientation (APO) was an anterior inclination of the pelvis 10° from neutral (accentuating the subject's lumbar extension or lordosis). The posterior pelvic orientation (PPO) was a 10° posterior inclination of the pelvis relative to neutral (accentuating the subject's lumbar flexion or kyphosis).

Each subject's session began with adjustments of the shake table's air bladder (to endure vibration over-travel did not occur) and the adjustable footrest (to ensure lower extremity alignment in a position of 90° of hip, knee, and ankle flexion with feet resting flat on the footrest). Bipolar surface electrodes were centered over the erector spinae mass at the L1 and L3 spinal levels (L1L, L1R, and L3L or L3R) with the electrode centers parallel to the underlying muscle fibers and attached using double-sided adhesive washers and adhesive tape. A reference electrode was placed over the distal anterior right tibia.

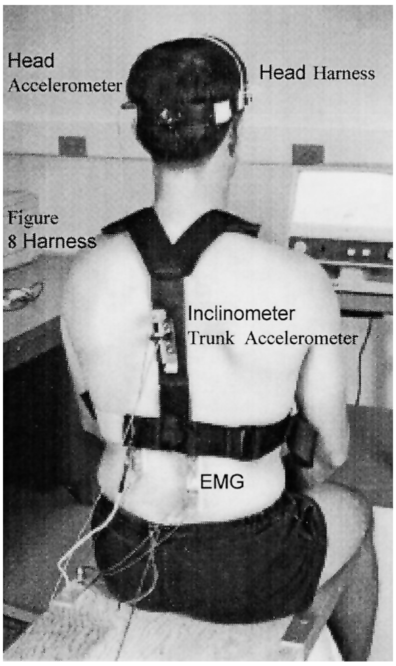


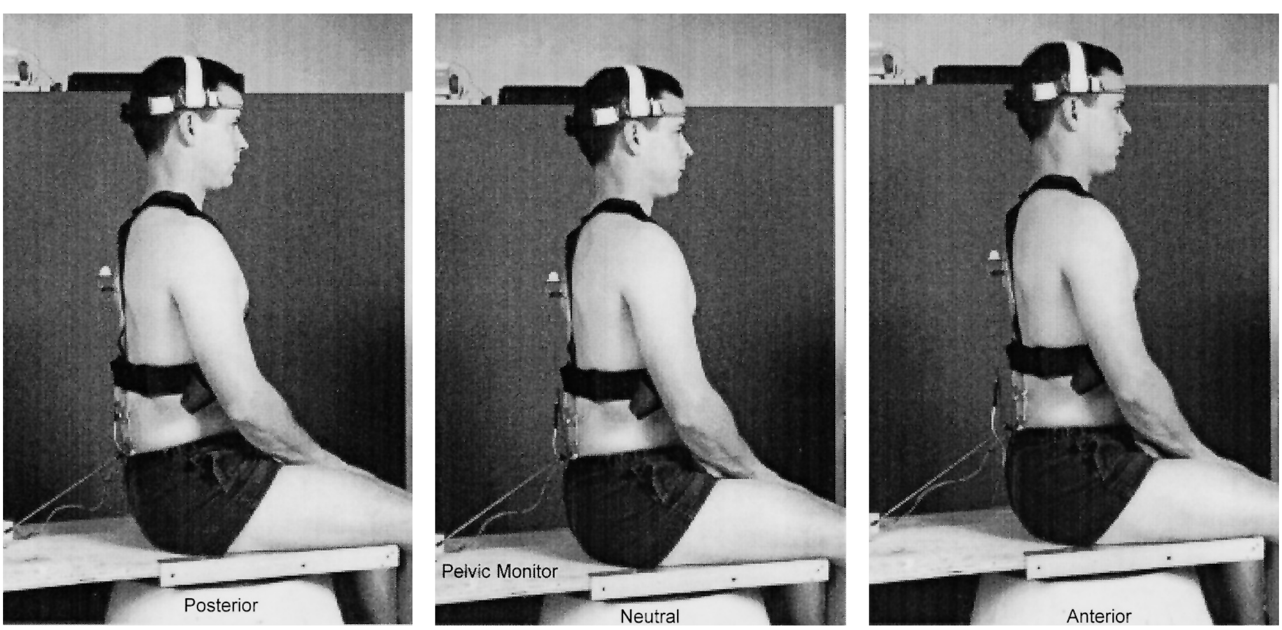
Fig. 1 Head accelerometer harness and figure-of-eight trunk harness and attachments. Head accelerometer harness includes adjustable headband, accelerometer and visual feedback devices. Figure-of-eight harness attachments include trunk inclinometer and trunk accelerometer. Pelvic Motion Monitor (includes two potentiometers, sliding rods, seat attachment and pelvic attachment plate)

Each subject was fitted with the figure-of-eight harness with attached accelerometer and tilt sensor. The harness was adjusted to align its vertical portions over the subject's anterior and posterior mid-line and the accelerometer and tilt sensor attachment plate at the T5 level, with the trunk accelerometer and inclinometer adjusted perpendicular to vertical. The headband was placed on the subject's head and the adjustable markers and accelerometer were aligned in the horizontal plane. The sacral motion monitor was attached to the subject's sacrum (along the sacrum's superior edge and centered over the sacral mid-line) with double-sided adhesive washers and adhesive tape (Fig. 1).

Following initial set-up, pretest resting baseline EMG was collected for 30 s in each of the three pelvic orientations. Vibration exposure was then initiated, the subject assumed one of the predetermined pelvic orientations, and data were collected. The subject then assumed the second and third pelvic orientations and the data collection procedure was repeated. This process was repeated at each vibration frequency. During testing, the sampling period for each specific frequency was as follows: 30 s at 4.5 Hz, 25 s at 5 Hz and 20 s at each remaining frequency. After collection of the data with the last frequency–pelvic orientation combination, vibration was stopped and a 30 s post-test baseline EMG record was collected for each of the three pelvic orientations. Total setup and testing time for each subject was approximately 1.5 h. Vibration exposure duration during testing approximated 0.5 h.

The experimental design for the study was a 3×7 repeated measures design. The order of exposures to frequencies was increasing for one half of the subjects and decreasing for the other half. The order of pelvic orientations was balanced across all subjects. Independent variables included pelvic orientation and vibration frequency. Dependent variables included peak-to-peak pelvic motion (pelvic motion), seat-to-trunk acceleration transmissibility (TAT), seat-to-head acceleration transmissibility (HAT), mean rectified erector spinae EMG, and peak-to-peak rectified erector spinae EMG.

Fig. 2 Subject in all three pelvic orientations (anterior, neutral and posterior pelvic orientations obtained while maintaining an upright trunk and level head)



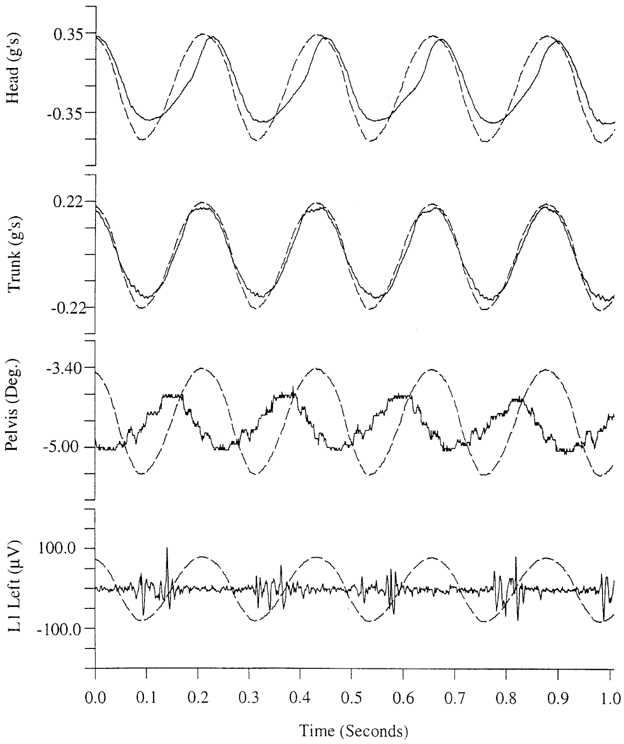


Fig. 3 Example of raw data collected at 4.5 Hz in subject 5 (*sinusoidal dashed tracings* are seat pan accelerations). Values for pelvis are relative to vertical not subject neutral

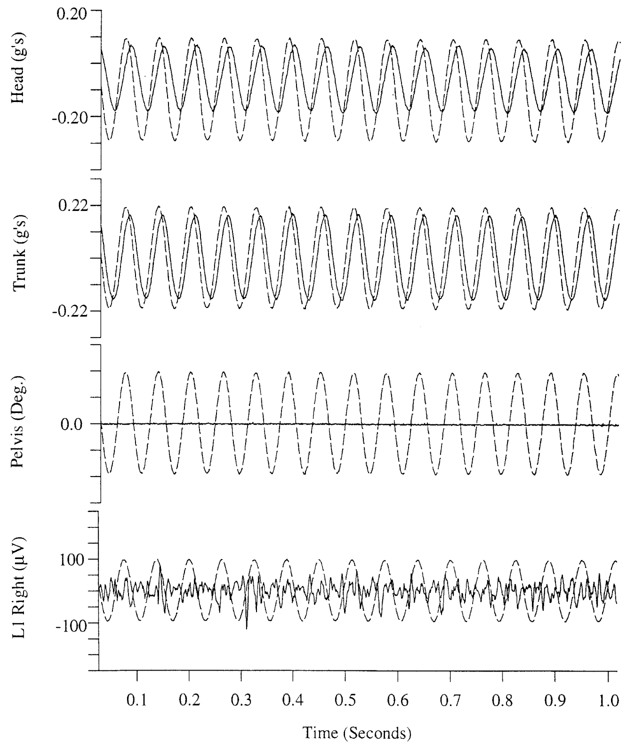


Fig. 4 Example of raw data collected at 16 Hz in subject 15 (*sinusoidal dashed tracings* are seat pan accelerations). Values for pelvis are relative to vertical not subject neutral. Note stability of pelvis

Using the downward zero crossing of the seat pan accelerometer as a trigger, raw acceleration, EMG, and pelvic motion ensemble averages, two cycles in length, were created during on-line data collection. All data were sampled at 1024 Hz. Sampling duration varied by frequency allowing collection of 60 two-cycle epochs at 4.5 Hz, 50 two-cycle epochs at 5 Hz and a number of two-cycle epochs equivalent to ten times the vibration frequency value at frequencies of 6 Hz or higher. Figures 3 and 4 provide examples of the raw data for acceleration, pelvic motion and EMG at 4.5 and 16 Hz, respectively. Ensemble average data examples for acceleration, EMG and pelvic motion are presented in Fig. 5 through 7 for subject 5.

Mean and peak seat, trunk, and head acceleration, mean pelvic orientation, pelvic motion and mean and peak-to-peak erector spinae EMG values for each subject were obtained from the ensemble averages. Transmissibility was determined by division of the peak head or trunk acceleration by one half the peak-to-peak seat pan acceleration for each trial. Group means for all dependent variables were used for statistical comparisons. Between-frequency and between-pelvic orientation comparisons were made for pelvic motion, TAT, HAT, mean erector spinae EMG, and peak-to-peak erector spinae EMG (P-P EMG).

Statistical analysis was performed using SAS software (SAS Institute, SAS Circle, Box 8000, Cary, N.C.). All main effects and interactions were evaluated using a two-way repeated-measures analysis of variance (ANOVA) procedure. For comparisons demonstrating an interaction effect, a simple effects analysis was done, using a one-way repeated-measures analysis of variance within each pelvic orientation and within each frequency. Tukey's and Scheffe's post hoc analyses were used to identify significant differences within the pelvic orientation and frequency analyses, respectively (main and simple effects). The level of significance was established at the $P < 0.05$ level for main effects analysis and adjusted for simple effects and posthoc analyses.

Results

As a measure of independent (pelvic orientation, vibration frequency) and experimental control (seat pan acceleration, trunk and head orientation) variable consistency across all frequency–pelvic orientation combination (FPOC), the group mean was determined for each FPOC. Differences in mean pelvic position between adjacent pelvic orientations were 8.75° anterior relative to neutral and 8.8° neutral relative to posterior. Each pelvic orientation was significantly different from every other ($P < 0.05$). The peak seat pan acceleration across all FPOCs exhibited differences in peak acceleration of 0.0134 g. This difference was the result of a minimum mean value (0.13 g) at 16 Hz vibration in the NPO and a maximum mean value (0.143 g) at 6 Hz in the APO $-$ 7.1% and $+$ 2.4% of the intended input (0.14 g r.m.s.), respectively.

Trunk and head orientation (relative to upright and horizontal, respectively) were also evaluated. The difference between the maximum and minimum trunk acceleration means across FPOCs (0.003 g) was equivalent to a 0.17° change in the accelerometer's orientation. The difference in mean head acceleration across FPOCs (0.0027 g) was equivalent to 0.15° change in accelerometer orientation. A DC offset range (equivalent to 1.49–1.64° angular orientation) of the

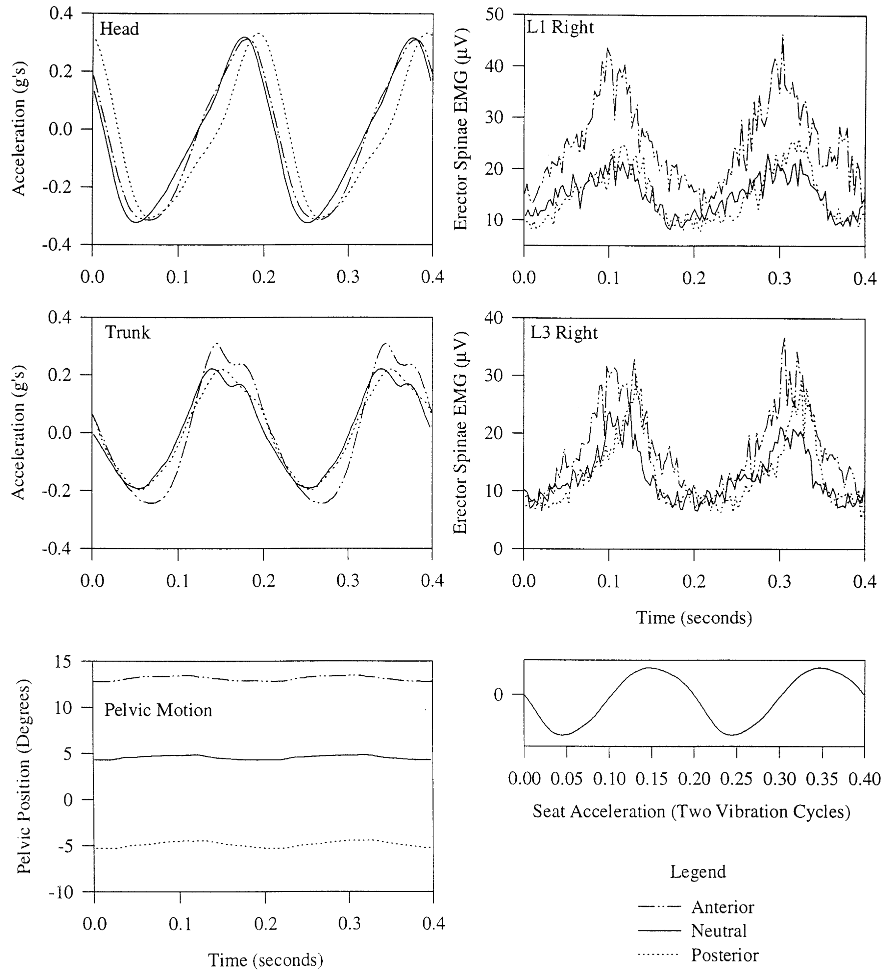


Fig. 5 Ensemble averages of raw data collected in subject 5 at 5 Hz

accelerometer relative to horizontal was also apparent. In summary, considering the dynamics of the environment investigated and their effects on position sense, the data indicate that the independent variables in the study were well controlled.

Pelvic motion

Inspection of the group mean pelvic motion ensemble averages across FPOCs revealed considerably greater pelvic motion at frequencies of 6 Hz and lower, with little or no pelvic motion occurring at frequencies higher than 6 Hz. Across pelvic orientations there was greater separation in the 4.5–6 Hz region, with a strong convergence of all postures at frequencies greater than 6 Hz (Fig. 8). (It should be noted that the resolution of the potentiometer was 0.33°.) Statistical analysis of the pelvic motion data identified a significant interaction between posture and frequency and significantly greater pelvic motion at vibration frequencies below 8 Hz than at frequencies above 8 Hz in all pelvic orientations. Also, significantly greater pelvic motion occurred in the PPO than in the APO and NPO at frequencies of 4.5 and 5 Hz (Fig. 8).

Trunk acceleration transmissibility

Trunk acceleration transmissibility values remained relatively unchanged within pelvic orientations. The least variability across frequencies occurred in the PPO and the greatest, in the NPO. A significant interaction between pelvic orientation and frequency occurred, because of the lack of difference across pelvic orientations at 4.5 Hz. At frequencies of 6 Hz and more there were significant differences between all pelvic orientations. The greatest differences across frequency occurred between 4.5 Hz and frequencies 6 Hz or higher in the APO and NPO. Generally, the PPO exhibited no differences across frequencies (Fig. 9).

Head acceleration transmissibility

Head acceleration transmissibility (HAT) displayed an obvious and significant interaction between pelvic

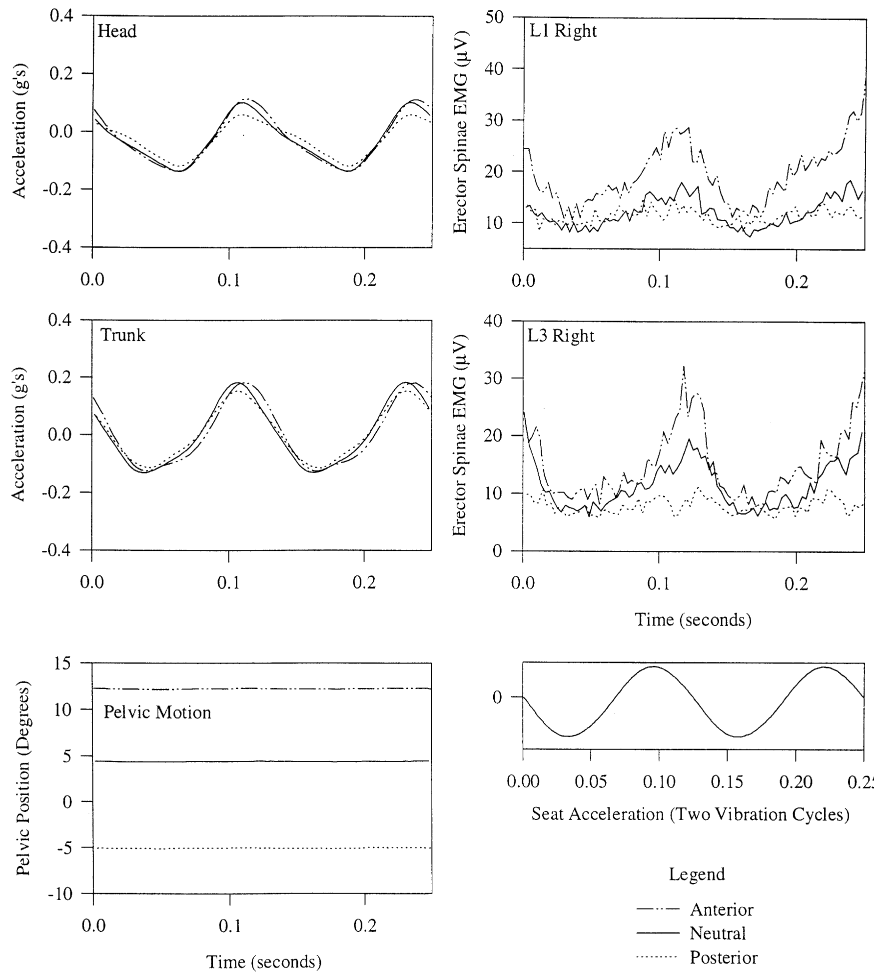


Fig. 6 Ensemble averages of raw data collected in subject 5 at 8 Hz

orientation and frequency, with frequencies lower than 6 Hz exhibiting significantly greater HAT than frequencies of 6 Hz and higher. Near the resonant frequencies (4.5–6 Hz), the differentiation between pelvic orientations was less clear than that at higher frequencies. In the region of 6 Hz a reversal in magnitude for the PPO relative to APO and NPO occurred. All HAT minima occurred at 8 Hz and all maxima occurred between 4.5 and 5 Hz. HAT in the APO was significantly less than that in the PPO at 4.5 Hz and significantly greater at frequencies of 6 Hz and higher. HAT in the APO was also significantly greater than in the NPO at frequencies from 6 to 12 Hz. HAT in the NPO was significantly less than in the PPO at 4.5 Hz and significantly greater at frequencies higher than 6 Hz (Fig. 10).

Mean rectified erector spinae EMG

The mean EMG data were consistent across frequencies within pelvic orientations. The PPO exhibited the maximum variation across frequencies at all four electrode placement sites: L1L, 18%; L1R, 16.2%; L3L, 13.5%; and L3R, 24.4%. The mean EMG values across frequencies were much more consistent in the APO and NPO at all four sites. These varied from a minimum of 4.6% (L1L, APO) to a maximum of 13.1% (L3L, APO). Statistical analysis of the mean EMG data across all FPOCs revealed a significant effect for pelvic orientation in all four electrode placement sites. However, in the case of the mean L1L EMG, a significant interaction was noted between pelvic orientation and frequency. Simple effects analysis of this case revealed that, at frequencies of 4.5 and 16 Hz, there was not a significant difference in the mean EMG activity level between the APO and NPO. However, the APO mean EMG was greater than that in the NPO (Fig. 11). Mean erector spinae EMG in the APO was significantly greater than in the NPO or PPO in all cases of electrode placement, with the L1L site findings reported for 4.5 and 16 Hz being the only exception. Mean erector spinae EMG in the NPO was significantly greater than in the PPO for the L1R, L3L and L3R electrode placement sites. The L1L site showed no significant difference between the NPO and PPO at any vibration frequency. The NPO mean EMG value,

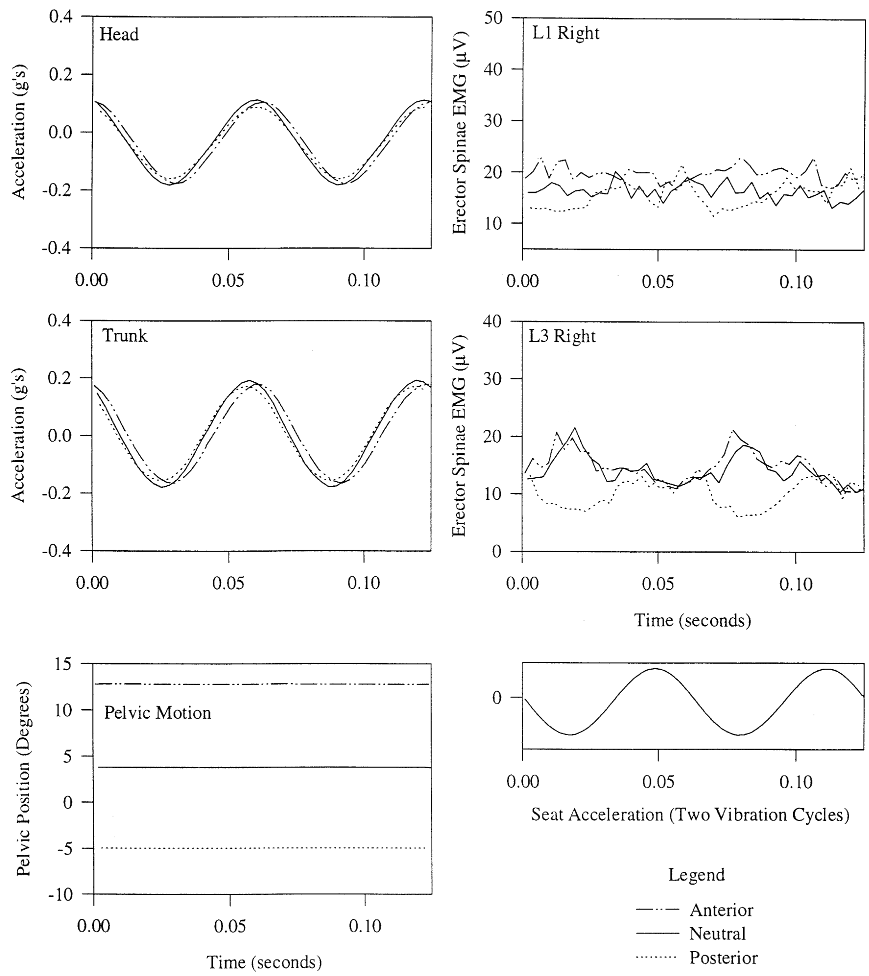


Fig. 7 Ensemble averages of raw data collected in subject 5 at 16 Hz

however, was greater than that for the PPO at all frequencies.

Peak-to-peak rectified erector spinae EMG

The peak-to-peak rectified erector spinae EMG (P-P EMG) response was similar for all electrode placement sites. All sites displayed a convergence of P-P EMG values across pelvic orientations as vibration frequency decreased from 6 Hz. As the frequency increased from 6 Hz, the separation of responses across pelvic orientations became more obvious. At 16 Hz the right and left sides behaved differently, with a convergence between NPO and APO on the right side at the L1 and L3 levels and no apparent change in separation on the left side. Statistically these changes were noted as an interaction between pelvic orientation and frequency for the L1 level. The L3 level displayed no statistical interaction, which may be due to the reduced sample size for the L3 electrode placement sites. Therefore, any discussion of the L3 level results in the following paragraph is based upon post hoc analyses.

There were no significant differences in P-P EMG between pelvic orientations at frequencies below 8 Hz. The P-P EMG response in the PPO was greatest in the frequency range of trunk resonance (4.5–6 Hz). The P-P EMG response in the APO was significantly greater than in the NPO for frequencies above 6 Hz in the cases of L3L and L1R. It was also true for L1L at frequencies of 12 and 16 Hz, but not at 8 and 10 Hz. This was not found to be the case at any of the frequencies of interest for L3R (Fig. 12).

Pelvic motion and head acceleration transmissibility

A positive relationship between pelvic motion and HAT was identified within pelvic orientation at frequencies of vibration less than 6 Hz. Across pelvic orientations there is considerable variability in pelvic motion magnitude with very little variation in HAT. Within pelvic orientation at frequencies greater than 8 Hz there is a negative association between pelvic motion and HAT. Across pelvic orientations, in this frequency range, there is very little

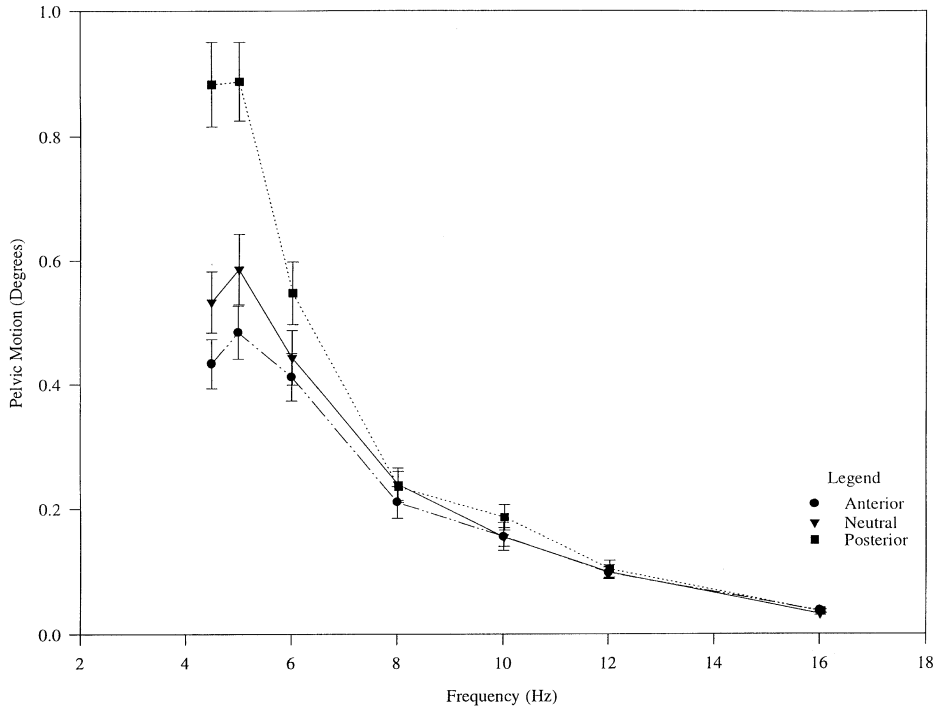


Fig. 8 Group mean pelvic motion (*vertical bars* standard error)

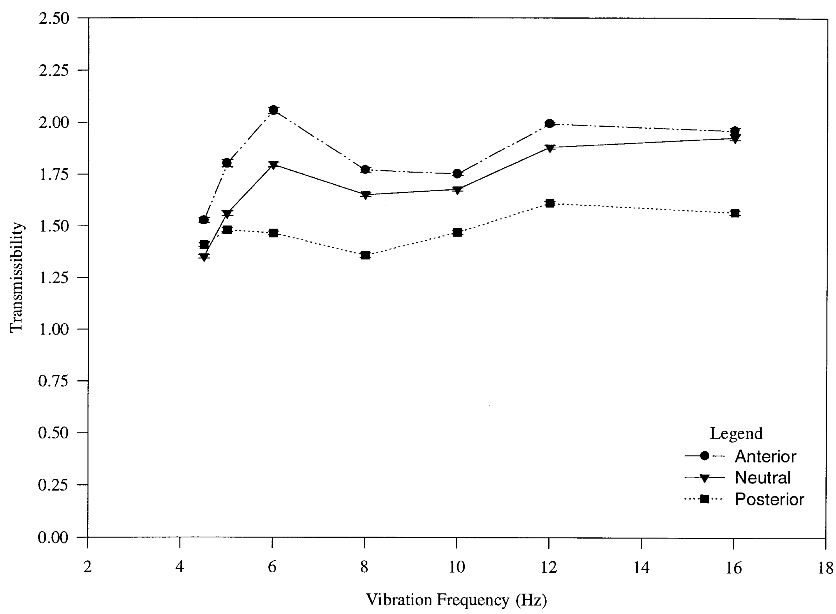


Fig. 9 Group mean trunk acceleration transmissibility (*vertical bars* standard error)

difference in pelvic motion with significant differences in HAT. The Pearson product moment correlation coefficients for these associations (group mean ensemble averages) were determined within the frequency ranges up to and including 6 Hz and over 6 Hz by pelvic orientation. Association magnitudes ranged from $R = 0.77$ to 0.99, and r^2 values ranged from 0.59 to 0.99 (Fig. 13). (Note: associations for frequencies up to 6 Hz include only 3 points.)

Peak-to-peak erector spinae EMG and pelvic motion

It has been proposed that the erector spinae P-P response would be sensitive to the magnitude of pelvic motion (Zimmermann 1995). The P-P EMG pelvic motion association across all conditions was found to have an r^2 of 0.19, with the relationship having a positive slope. Evaluations of the association within pelvic orientation for frequencies up to and including 6 Hz

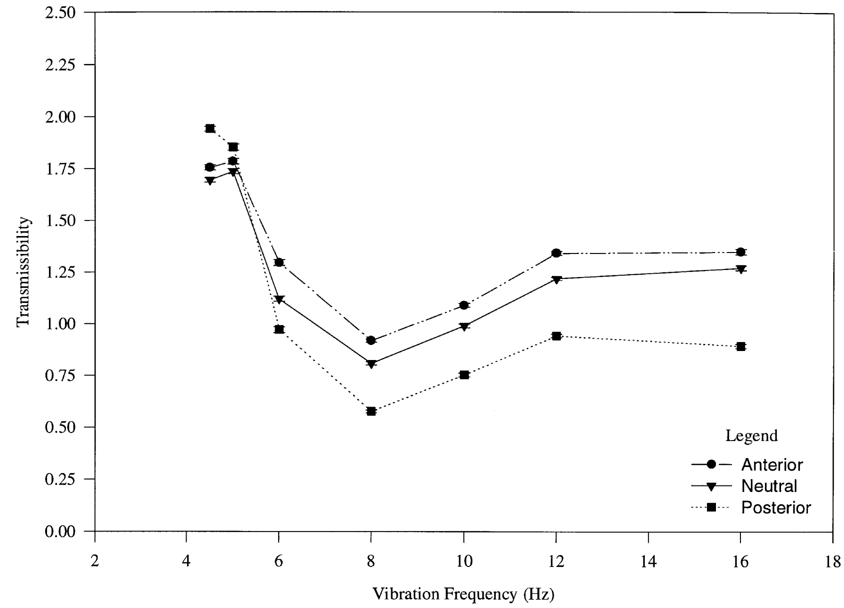


Fig. 10 Group mean head acceleration transmissibility (*vertical bars* standard error)

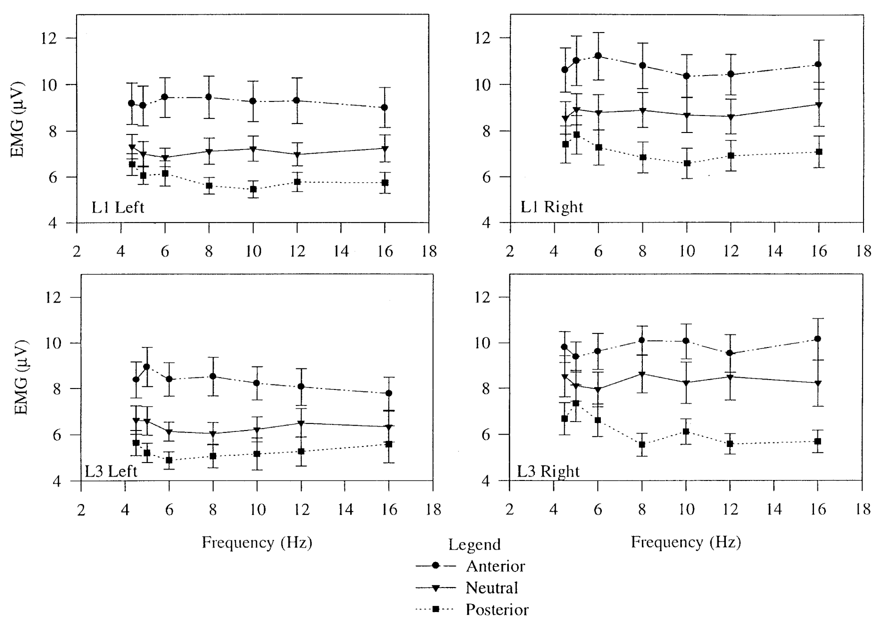


Fig. 11 Group mean erector spinae EMG (*vertical bars* standard error)

revealed that the strongest association between the two dependent variables [$r^2 = 0.57$ ($R = 0.75$)] occurred in NPO. The association's strength was considerably less in the APO and PPO. At frequencies higher than 6 Hz, associations within pelvic orientation were all very strong. The r^2 values were 0.80, 0.87 and 0.95 for APO, NPO and PPO, respectively (Fig. 13).

Mean and peak-to-peak EMG

The relationship between mean and P-P group ensemble mean EMG values in the present study was determined across all four EMG electrode placement sites and all 21 FPOCs. The Pearson correlation coefficient, R, for the association across all FPOCs was 0.79. The r^2, which describes the amount of P-P EMG variability than can be accounted for by the mean EMG level was 0.63. Evaluation of the relationship within pelvic orientations across frequencies revealed r^2 values of 0.51, 0.52, and 0.59 for anterior, neutral and posterior, respectively. The slopes for the associations were 1.24, 1.41 and 2.26, respectively (Fig. 13).

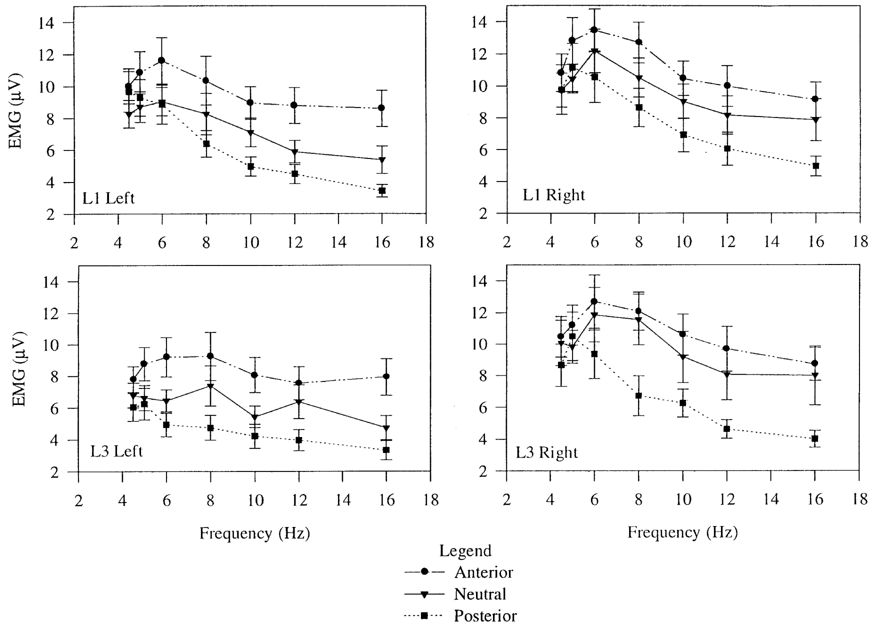


Fig. 12 Group mean peak-to-peak erector spinae EMG (*vertical bars* standard error)

Discussion

The results of this study indicate that the human response to WBV in an unsupported seated posture (as defined and applied in this study) is complex and interactively dependent on both vibration frequency and pelvic orientation. The frequencies used are similar to those used previously (Park 1991; Robertson and Griffin 1989; Seidel et al. 1986; Seroussi et al. 1989; Wilder et al. 1982) and reported to commonly occur in vehicles (Gruber 1976; Wilder et al. 1982; R. Penzotti, personal communication, 1983). The vibration magnitude used at all frequency levels (1 m/s^2 r.m.s.) allowed for comparison of results with those of previous studies (Park 1991; Robertson and Griffin 1989; Seidel et al. 1986; Seroussi et al. 1989) and compliance with the 1-h ISO guideline on exposure to WBV (ISO 2631/1, 1985). Given the experimental conditions evaluated, this interactive response can most simply be divided into how the human system responds above and below 6 Hz and can further be divided into how the response in each of these frequency ranges is affected by pelvic orientation (APO, NPO and PPO). Table 1 provides a summary of the overall frequency range–pelvic orientation interactions found in the results. The values represent group mean data averages across the frequency ranges indicated and EMG values calculated using L1L and L1R only.

Pelvic motion exhibited effects due to both pelvic orientation and vibration frequency. These effects resulted in a “dichotomous” response based on vibration frequency. At frequencies of vibration up to and including 6 Hz there was considerably more pelvic motion than at frequencies of vibration greater than 6 Hz. The result of decreasing pelvic motion with increasing vibration frequency may be related to the use of constant acceleration levels in this study and its effect on displacement magnitude across vibration frequencies. The greater pelvic motion observed at vibration frequencies near the seated human’s primary resonance (4.5–6 Hz) is reminiscent of previous studies evaluating acceleration transmissibility at the sacrum relative to the seat pan (Hagena et al. 1985, 1986; Panjabi et al. 1986; Pope et al. 1986).

Within the frequency range up to 6 Hz, more motion occurred in the PPO than in the APO (or NPO). Changes in the trunk’s line of gravity location relative to the ischial tuberosities may have contributed to this result. As described by Keegan (1953), in the PPO the trunk’s line of gravity falls posterior to the ischial tuberosities and anterior to the flexed lumbar spine. Thus, the PPO results in a trunk mass moment arm tending to produce increased posterior pelvic rotation and lumbar spine flexion, which may facilitate stretch of the lumbar erector spinae. This may account for the increased erector spinae peak-to-peak EMG and increased pelvic rocking observed in the PPO near the resonant frequencies (4.5–6 Hz). In the APO, the trunk’s line of gravity falls anterior to the ischial tuberosities and through or posterior to the extended lumbar spine. Seat pan vertical motion in this posture should facilitate further flexion of the pelvis relative to the hip and increased lumbar spine extension. Both of these movement patterns may be affected by passive restraint mechanisms (pelvis, hip joint capsule; lumbar spine, vertebral facets), tending to reduce the pelvic motion relative to the seat pan and resulting in a more stationary pelvis (Schoberth 1962).

Trunk acceleration transmissibility (TAT) was also affected by both frequency range and pelvic orientation. In general, TAT was less at frequencies up to and including 6 Hz than at frequencies greater than 6 Hz.

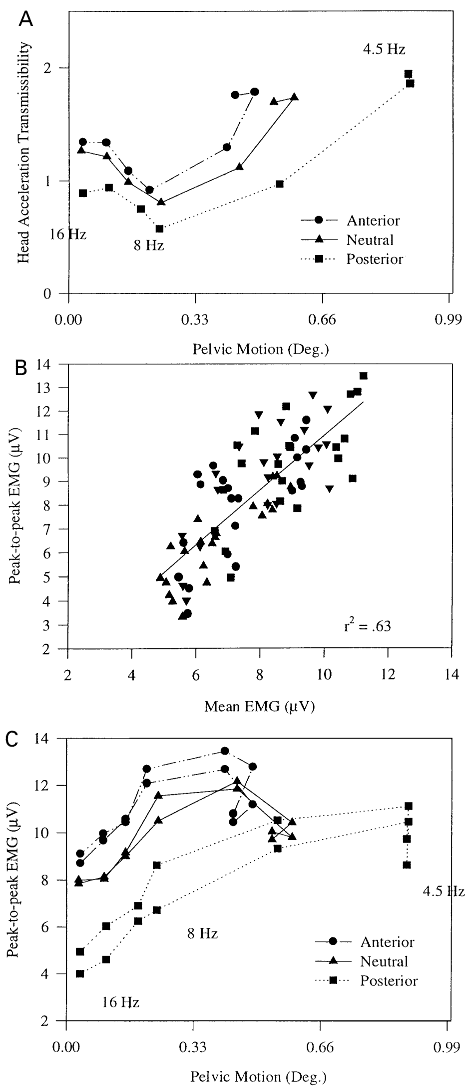


Fig. 13 A Pelvic motion vs head transmissibility; **B** mean vs peak-to-peak EMG; **C** peak-to-peak EMG vs pelvic motion

TAT values in the study appear to be similar to those previously reported for skin-mounted micro accelerometers (Hinz et al. 1988; Seidel et al. 1986). However, the value of these earlier studies is limited insofar as only one subject was evaluated at two frequencies, 4.5 and 8 Hz (Hinz et al. 1988), and reported data for frequencies beyond 8 Hz are lacking (Seidel et al. 1986). In the present study, as well as in previous TAT studies, measurement artifacts due to movement of the skin, harness, and shoulders are a potential problem. The non-sinusoidal appearance of the response curves across all frequencies and pelvic orientations may be due to movement artifacts or the influence of pelvic rocking and changes in the lumbar spine configuration. With these potential limitations of the data in mind, the results of the present study indicate that, like pelvic motion, TAT is dependent on both vibration frequency and pelvic orientation. If increased transmissibility is related to decreased performance, then the influence of pelvic orientation shown in the present study would apparently be of importance for determining appropriate vibration exposure guidelines.

Head acceleration transmissibility's (HAT) primary resonance frequency (4.5–5 Hz) and magnitudes are in agreement with previous literature (Messenger and Griffin 1989; Sandover 1982; Wilder et al. 1982; Dieckmann 1957; Coermann et al. 1960; Griffin et al. 1979; Griffin 1975; Paddan and Griffin 1988; Seidel et al. 1986). As with several other measures in this study (Table 1), HAT exhibited differing responses dependent on pelvic orientation and frequency range. Generally, at frequencies higher than 6 Hz, APO resulted in significantly greater HAT than PPO (or NPO). At frequencies of 4.5 and 5 Hz, the PPO resulted in increased transmissibility relative to APO (or NPO). Findings that are similar to the trends observed in the pelvic motion response at frequencies up to and including 6 Hz.

Effects of changes in posture on HAT have been investigated by several other researchers. Their investigations would indicate that: (1) "stiff" postures result in increased HAT at frequencies above 6 Hz and decreased HAT at frequencies below 6 Hz (Griffin et al. 1979); (2) increased APO (85–105° of pelvic femoral

Table 1 "Dichotomous" nature of dependent variable responses

Dependent variables	Less than or = to 6 Hz			vs.	Greater than 6 Hz		
	Anterior		Posterior		Anterior		Posterior
Pelvic motion	0.47°	<	0.74°	>	0.15°	=	0.16°
Trunk acceleration transmissibility	1.76°	>	1.45°	⩽	1.86°	>	1.52°
Head acceleration transmissibility	1.61°	=	1.59°	>	1.17°	>	0.79°
Mean erector spinae EMG	9.55 µV	>	6.11 µV	=	9.55 µV	>	6.11 µV
Peak-to-peak erector spinae EMG	11.58 µV	=	9.89 µV	>	9.87 µV	>	5.74 µV

angle) results in decreased transmissibility at 4 Hz and increases of 40–60% between 6 and 40 Hz (Messenger and Griffin 1989); (3) increased hip flexion in conjunction with an increased anterior trunk mass moment arm (bending forward at the hips and trunk to obtain a 30° flexion posture) results in decreased transmissibility of 60–70% between 6 and 35 Hz compared with normal upright, a response similar to the PPO in the present study (Messenger and Griffin 1989); and (4) a 20° flexion posture, obtained through the maintenance of pelvic position and forward trunk flexion, results in a 30–40% transmissibility decrease at frequencies between 6 and 35 Hz compared with a normal upright posture in both supported and unsupported sitting (Messenger and Griffin 1989). Messenger and Griffin's (1989) second and third mechanisms for postural change differ from that used in the present investigation, since increased mean EMG was associated with an increased trunk mass moment arm and decreased coupling of the posterior elements of the vertebrae. This decreased transmissibility associated with increased muscle activity seems to suggest a greater relative importance of lumbar spine posture and pelvic orientation versus mean erector spinae EMG in determining acceleration transmissibility. These findings, in conjunction with the results of the present study, indicate that coupling of the posterior elements of the lumbar vertebrae is a major factor in determining HAT. Also, the mean EMG associated with a posture may be less important than the coupling of the lumbar vertebral posterior elements, orientation of the vertebral bodies or orientation of the pelvis in determining HAT at frequencies greater than 6 Hz.

Mean EMG activity is indicative of the muscle activity required to maintain a given posture. Any changes in mean EMG activity should reflect changes in posture (Assmusen 1960; Floyd and Silver 1955; Portnoy and Morin 1956) or co-contraction of the agonists and antagonists about a joint to facilitate stability (Lacquaniti and Maioli 1987). In contrast to the results discussed in the previous sections, the mean erector spinae EMG response was statistically unchanged for all vibration conditions within each pelvic orientation. Since the mean moment generated by the body mass above a given vertebral level is constant with unchanged posture, the results reported here should be expected (Zimmermann et al. 1993). However, increases in mean EMG in vibratory versus non-vibratory environments, while seated posture remains unchanged, have been reported (Seroussi et al. 1989; Robertson 1986; Robertson and Griffin 1989). In the study conducted by Seroussi et al. (1989), the subject's posture, anterior trunk lean and lordotic lumbar spine, was controlled by rigid and semi-rigid external bodies in contact with the subject's lumbar spine and chin. The differences in mean EMG activity may be a result of force exerted against these objects in the subject's attempt to maintain tactile feedback during vibration exposure. Robertson and Griffin (1989), in an evaluation of long-term vibration exposure, reported increases in the mean EMG 1.5 h of exposure, which may be fatigue related. In a short-term exposure study, Robertson (1986) reported an increase in the minimum EMG activity above resting baseline during vibration exposure at 16 Hz, which he attributed to slight changes in posture or co-contraction of the trunk flexors and extensors.

The current study's mean EMG changes associated with changes in pelvic orientation or posture are consistent with those of Zimmermann et al. (1993). Two earlier investigations (Seroussi et al. 1989; Robertson and Griffin 1989) evaluated mean erector spinae EMG in a single posture. However, the significant differences across pelvic orientations noted in the present investigation's results point out a limitation of single posture evaluations. Thus, if the posture in the work environment is different from that being evaluated, the application of results is limited.

The peak-to-peak EMG response magnitude trend in the PPO [greater magnitude at lower vibration frequencies ($\leqslant 6$ Hz) than at higher vibration frequencies (> 6 Hz)] is similar to those previously reported for pelvic motion and HAT. The identification of a peak-to-peak EMG response in the PPO may be attributed to the fact that the PPO adopted in the current study was not maximal posterior pelvic rotation and allowed the erector spinae to remain at a length shorter than that observed during the erector spinae's "silent period".

This investigation identified no significant differences in peak-to-peak EMG between any pelvic orientations at vibration frequencies below 8 Hz, suggesting that the role of pelvic motion as a stimulus for the peak-to-peak EMG response has greater influence than mean EMG activation at frequencies below 6 Hz. However, at higher vibration frequencies (> 6 Hz) there was significantly greater peak-to-peak erector spinae EMG in the APO than the PPO, which would indicate a frequency dependence of the role of mean EMG on the peak-to-peak erector spinae response. Zimmermann et al. (1993) reported similar differences in the peak-to-peak erector spinae EMG response as a result of changes in posture.

The effect of vibration frequency on the peak-to-peak erector spinae EMG response has been studied by several investigators (Robertson and Griffin 1989; Seidel et al. 1986; Seroussi et al. 1989). Seroussi et al. (1989) reported a significant increase in the peak-to-peak erector spinae response magnitude in the vibratory environment compared to static sitting but did not perform peak-to-peak EMG comparisons across vibration frequencies. Figure 5a in their publication, however, indicates that the greatest peak-to-peak response appeared at frequencies below 6 Hz, with a plateau occurring at frequencies higher than 6 Hz. These results are similar to those observed in the NPO

(Fig. 12). Robertson and Griffin (1989), in a study of upright seated vibration exposure at frequencies of 1, 4, 8, 16, and 32 Hz and increasing vibration intensities, reported a decreased phasic or peak-to-peak magnitude of the EMG response as the vibration frequency increased or decreased from 4 Hz. Results similar to the changes associated with the 6-Hz region in the present investigation. This difference in frequencies may be related to their EMG processing techniques, which included high-pass filtering at 80 Hz and smoothing with a 100-ms time constant. Seidel et al. (1986) evaluated changes in the estimated normalized moments at the minimum and maximum values of seat pan acceleration. Their EMG response pattern at maximum seat pan acceleration follows the pattern observed in the present study's results (Fig. 12).

Although not as evident as in the acceleration transmissibility and pelvic motion results there was a peak-to-peak EMG "dichotomous" response (Table 1). Near resonance (4.5–6 Hz) there were fewer differences across pelvic orientations, with increasing vibration frequency producing more prevalent differences across pelvic orientations. Also, there was a decline in the peak-to-peak EMG magnitude at frequencies greater than 6 Hz within pelvic orientations, as noted previously. This would indicate a greater effect of increased head, trunk, and pelvic motion than mean erector spinae activity (or pelvic orientation) on the peak-to-peak erector spinae response magnitude, near resonance.

Pelvic motion and acceleration transmissibility at the head

Sandover (1982) and Seroussi et al. (1989) have commented on the potential effects of "pelvic rocking" on HAT. Sandover attempted to isolate the motions and contact surfaces of seated persons exposed to vibration and concluded that pelvic rocking was responsible for the resonance observed at 4.5–6 Hz. Therefore, the association between pelvic motion and HAT was evaluated using a first-order linear regression. The condition in which the r^2 was greater (PPO) was also the region of maximal transmissibility on the frequency spectrum and the pelvic orientation with the least mean EMG. It is interesting that the strongest association occurs in these conditions. (However, only three points were used in the regression calculations at frequencies up to and including 6 Hz, which may have contributed to the higher values.) In the condition with the greatest mean EMG and most rigid spinal system (APO), the association across all frequencies was the poorest. This poor association was a result of decreased pelvic motion and decreased peak-to-peak erector spinae EMG variability within the APO, leading to a lack of data spread and poor associations.

Support for the association between pelvic motion and HAT is evident in the previous WBV literature. Hagena et al. (1986) reported decreased transmissibility (sacrum to C7) between the frequencies of 4 and 6 Hz and transmissibility curves for frequencies beyond 6 Hz that appear very similar in shape and magnitude to those reported for HAT in the present study. This decreased transmissibility indicates that trunk motion relative to the sacrum is minimized in this frequency range, which is the same as the frequency range in which maximum pelvic rocking occurred in the present study. The identification of maximum head and minimum sacral-to-cervical acceleration transmissibility values within this frequency range points to pelvic rocking's influence on HAT (Sandover 1982).

Other investigators (measuring vertical vibration transmissibility between the seat pan and sacrum or ilium) report that maximum vertical motion of the pelvis occurs between 4.5 and 6 Hz (Hagena et al. 1985; Panjabi et al. 1986; Pope et al. 1986). Most have not reported measures of rotational pelvic motion, but M. H. Pope, (personal communication, 1995) has commented on the presence of significant anterior–posterior pelvic acceleration during seated WBV exposure. Based on the anatomical configuration of the pelvis, anterior–posterior acceleration of the pelvis is most probably associated with rotation of the pelvis about the ischial tuberosities, contributing to the acceleration peak observed in the region of 4.5–6 Hz.

Mean and peak-to-peak erector spinae EMG

If the peak-to-peak erector spinae EMG response is reflex based, a positive association might be expected between the magnitude of the phasic component and the mean activation level. A significant positive association ($r^2 = 0.55$) was identified between the mean EMG and the peak-to-peak EMG across all FPOCs (Fig. 13). The identification of an association between mean and peak-to-peak erector spinae EMG magnitude in the present study differs from the results of Robertson and Griffin (1989), who indicated that during prolonged WBV exposure there was a mean EMG activation level increase at approximately 1.5 h of exposure, with no associated change in the magnitude of the peak-to-peak or phasic component of the muscle response even after 2.5 h of exposure. The positive association was identified in the current study, which leads one to question Robertson's findings of adaptation or fatigue.

The significance of this association involves the potential for greater compression of the intervertebral disc with increased muscular activation. Several mechanisms can result in increased mean erector spinae EMG during sitting. These include changes in seated posture, changes in pelvic orientation, application of external forces, and voluntary and involuntary co-contraction of agonists and antagonists to increase joint stability. The mechanism evaluated in this study, changes in the orientation of the pelvis while the seated posture remains relatively unchanged, resulted in significant differences between the mean EMG for each pelvic orientation, and yet the association between mean and peak-to-peak EMG remained significant across all pelvic orientations.

Peak-to-peak erector spinae EMG and pelvic motion

In an attempt at further definition of the stimulus for the erector spinae peak-to-peak response the association between peak-to-peak EMG and pelvic motion magnitude was evaluated across all frequencies and pelvic orientations and found to have a generally positive slope. There was a significant change in the association's slope at frequencies near resonance (4.5 and 5 Hz). As the vibration frequency increased from this range, the association became more consistent with significant improvement in the r^2 values. The limited resolution of the potentiometers incorporated in the pelvic motion monitor, and the apparent lack of significant differences in pelvic motion at vibration frequencies greater than 6 Hz, limits the strength of any arguments that may be proposed related to the association between pelvic motion and peak-to-peak erector spinae EMG. However, if the proposed interaction between pelvic motion and mean EMG activity in determining the magnitude of the VSR is accepted, then the data are consistent. As the pelvic motion decreased, the association between pelvic motion and peak-to-peak EMG declined while the association between mean and peak-to-peak EMG became more significant.

Changes in seated pelvic orientation produce significant effects on the response of the body to seated WBV. The response of the system is generally dichotomous in nature and dependent on both pelvic orientation and vibration frequency. The dichotomy of the human response occurs between vibration frequencies up to and including 6 Hz and those above 6 Hz.

At frequencies of 6 Hz and above, the human responds as a typical mechanical system. System changes that produce increased stability (APO, increased mean EMG) result in increased stiffness, higher resonance frequencies, and increased transmissibility, and changes in the system which decrease stability (PPO, decreased mean EMG) result in decreased stiffness, lower resonant frequencies and decreased transmissibility. At frequencies below 6 Hz, the human behaves in a somewhat different manner. Changes in the system that produce increased stability (APO increased mean EMG) result in increased stiffness and higher resonance frequencies, but decreased transmissibility, while changes that decrease stability (PPO, decreased mean EMG) result in decreased stiffness, lower resonant frequencies and greater transmissibility.

This dichotomy is also evident in the EMG and pelvic motion results. Peak-to-peak EMG differences across postures present at frequencies beyond 6 Hz are not present below 6 Hz, and significant differences in pelvic motion across postures occurring at frequencies up to and including 6 Hz are not present at frequencies beyond 6 Hz.

These differences indentify the importance of both vibration frequency and pelvic orientation when describing the effects of seated WBV exposure. If the intent of vibration exposure guidelines is to decrease the incidence of low back pain associated with exposure to WBV, then the issue of driver posture needs to be addressed. On the basis of the results of the current investigation it does not appear to be sufficient to propose standards that do not address occupant posture during vibration exposure.

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