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# Characterization of the Frequency and Muscle Responses of the Lumbar and Thoracic Spines of Seated Volunteers During Sinusoidal Whole Body Vibration

*Whole body vibration has been postulated to contribute to the onset of back pain. However, little is known about the relationship between vibration exposure, the biomechanical response, and the physiological responses of the seated human. The aim of this study was to measure the frequency and corresponding muscle responses of seated male volunteers during whole body vibration exposures along the vertical and anteroposterior directions to define the transmissibility and associated muscle activation responses for relevant whole body vibration exposures. Seated human male volunteers underwent separate whole body vibration exposures in the vertical (Z-direction) and anteroposterior (X-direction) directions using sinusoidal sweeps ranging from 2 to 18 Hz, with a constant amplitude of 0.4 g. For each vibration exposure, the accelerations and displacements of the seat and lumbar and thoracic spines were recorded. In addition, muscle activity in the lumbar and thoracic spines was recorded using electromyography (EMG) and surface electrodes in the lumbar and thoracic region. Transmissibility was determined, and peak transmissibility, displacement, and muscle activity were compared in each of the lumbar and thoracic regions. The peak transmissibility for vertical vibrations occurred at 4 Hz for both the lumbar (1.55 ± 0.34) and thoracic (1.49 ± 0.21) regions. For X-directed seat vibrations, the transmissibility ratio in both spinal regions was highest at 2 Hz but never exceeded a value of 1. The peak muscle response in both spinal regions occurred at frequencies corresponding to the peak transmissibility, regardless of the direction of imposed seat vibration: 4 Hz for the Z-direction and 2–3 Hz for the X-direction. In both vibration directions, spinal displacements occurred primarily in the direction of seat vibration, with little off-axis motion. The occurrence of peak muscle responses at frequencies of peak transmissibility suggests that such frequencies may induce greater muscle activity, leading to muscle fatigue, which could be a contributing mechanism of back pain.* [DOI: 10.1115/1.4027998]

*Keywords: vibration, muscle, spine, resonance, electromyography, transmissibility*

## Introduction

Whole body vibration has been hypothesized as a cause of back pain and injury across a wide range of individuals and exposures [1–3]. For example, low back pain has a higher prevalence among American male workers who operate vibrating vehicles, such as industrial trucks and tractors, than in workers whose occupations do not involve such exposures [1]. Also, military helicopter aviators report increased pain during deployment compared to predeployment, with between 39% and 70% reporting low back pain [2]. The typical whole body vibration exposures for train, helicopter, and bus drivers have been reported to have an amplitude between 0.02 and 1.75 $m/s^2$ over a range of frequencies between 2 and 25 Hz, and are directed vertically along the spine and in the anteroposterior direction [2–4]. Subjects exposed to vertical and anteroposterior sinusoidal vibrations report discomfort in localized regions of the body and at specific frequencies [5]; body discomfort has been reported at 4 and 8 Hz of vertically directed vibration and at 2 Hz in anteroposterior exposures, with no variation in discomfort levels with acceleration magnitude [5]. The lack of symptom dependence on acceleration and the response to specific applied frequencies suggest that vibration frequency may be an important input to the biomechanical and physiologic responses to whole body vibration. Even though these epidemiological studies strongly suggest that back pain can develop from whole body vibration and may be influenced by the frequency of the exposure, there has been little work defining the effects of whole body vibration frequency and physiological responses.

Several studies have defined the biomechanical response of the human and primate to whole body vibration in the vertical direction. A study of seated human volunteers undergoing vertical

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vibrations ranging in frequencies from 2 to 15 Hz with accelerometers on sacrum, first (L1) and third (L3) lumbar vertebrae reported resonance at 4.5 Hz [6]. Mansfield and Griffin, who used similarly seated volunteers with accelerometers on L3 and vertical vibrations ranging from 0.2 to 20 Hz with varying acceleration magnitudes, not only reported a similar primary resonance of 4–6 Hz but also detected a secondary resonance between 8 and 12 Hz [7]. Similarly, the resonant frequency of the seated primate undergoing vertically directed vibration has been reported to be between 9 and 15 Hz [8]. Although there have been a number of studies defining the biomechanics of these species during vertical vibration [6–8], only a few have studied whole body vibration directed anteroposteriorly and measured only the apparent mass at the seat level during random vibration exposures between 0.25 and 100 Hz in a seated human [9,10]. Although several studies have measured the transmissibility of the human spine [6,7], few studies have defined the mechanical effects of whole body vibration in a seated human in both the vertical [6,7] and anteroposterior directions [9,10] and none has studied the biomechanical responses of the lumbar and thoracic spines in directions other than vertically.

A limited number of studies have used EMG to study the human muscle response during whole body vibration [11,12]. For example, the muscle activity in the quadriceps increased significantly for standing subjects during vertical whole body vibration at discrete frequencies (30, 40, 50 Hz) compared to the level of muscle activity in nonvibrated subjects [11]. In addition, the time to muscle fatigue and the degree of muscle fatigue in the lumbar and thoracic spines increased for seated subjects exposed to a 5 Hz vertical whole body vibration compared to the response of those same subjects without any vibration [12]. Although the findings of increased muscle activity and fatigue with whole body vibration suggest a possible mechanism that may contribute to the development of pain and other symptoms, no studies have investigated the muscle response during whole body vibration over those frequencies relevant to the exposures of seated occupants in the lumbar and thoracic regions for both the vertical and anteroposterior directions.

The aim of this study was to characterize the frequency and muscle responses of seated human volunteers during whole body vibration exposures along the vertical (*Z*-direction) and anteroposterior (*X*-direction) directions. Specifically, the transmissibility, which is the ratio of the output acceleration to the input acceleration, of the lumbar and thoracic spines was defined for vibration directed along the spine's long-axis (vertically) and in the anteroposterior direction (front–back), in order to define the frequency response of the human. Because resonance occurs at a transmissibility ratio greater than one and maximum resonance occurs at the system's resonant frequency [7], it presents the greatest displacements on anatomical structures. In addition, EMG also was used to measure the corresponding muscle activity in the lumbar and thoracic spines during those imposed vibrations in order to compare the corresponding muscle and resonant responses. Based on prior transmissibility studies [6–10] and known exposures of workers exposed to whole body vibration [2–4], each exposure consisted of a sine sweep between 2 and 18 Hz in the *Z*-direction and *X*-direction.

## Methods

The analyses presented here are performed using unanalyzed datasets from experiments performed in 1994 and 1995 [13], of which a portion of the data were used to develop a set of repeated jolt health hazard assessment criteria for soldiers [13,14]. All procedures were USAMRMC IRB-approved and performed with informed consent. Prior to participation in the experiments, subjects underwent a focused medical examination. Subjects with a history of back pain or strain, recent trauma or surgical procedures, presence of internal or external prosthesis, and disorders of the muscular–skeletal system were not included in these experiments. Five, healthy, male volunteers ($24.8 \pm 2.2$ yr; $73.9 \pm 7.3$ kg; standing height $1.80 \pm 0.05$ m) participated in the study.

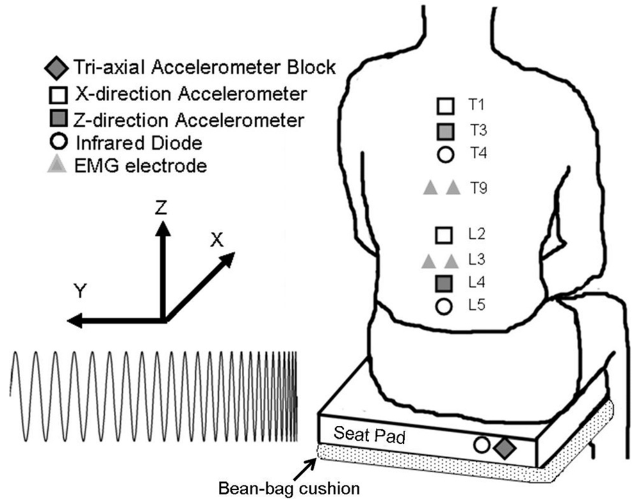


**Fig. 1 Schematic illustrating the experimental setup showing placement of accelerometers, infrared diodes, and EMG electrodes on the lumbar and thoracic regions of the spine and seat pad. The *Z*-(vertical), *X*-(anteroposterior), and *Y*-(lateral) directions, and the seat pad acceleration profile are also indicated.**

Each subject underwent separate exposures on different days of whole body vibration directed vertically (*Z*-direction) and horizontally in the anterior–posterior direction (*X*-direction) (Fig. 1). Subjects were seated on a solid metal seat securely mounted on a shaker table (Multi-axis Ride Simulator; Schenk Pegasus) with a bean-bag cushion taped to the top of the metal seat. The bean-bag cushion was designed to distribute the subject's weight without altering the input acceleration signal. A molded epoxy seat pad was placed between the subject and the bean-bag cushion and firmly secured with tape to ensure that acceleration due to vibration of the metal seat structure or damping effects of bean bag cushion would not affect the accuracy of the input accelerations reported (Fig. 1). The seat was not equipped with a backrest since most drivers and occupants of tactical ground vehicles did not utilize a backrest at the time this study was conducted [13]. The seat was adjusted such that the subject's feet rested comfortably with the hips and knees at approximately 90 deg. Subjects were asked to sit in a comfortable upright position facing forward, with their hands placed on their knees for the duration of the test. The seat pad was equipped with three, single-axis accelerometers (EGAX-25; Entran Devices) positioned in a tri-axial accelerometer block that measured the input acceleration (Fig. 1). Single-axis accelerometers (EGAX-25; Entran Devices) were attached to each subject's skin by a small square ($<1$ cm$^2$) of two-sided adhesive tape placed over the vertebral processes in the lumbar and thoracic regions (Fig. 1). The accelerometers at L4 and T3 were oriented to measure the accelerations along the *Z*-direction and those at L2 and T1 were oriented at 90 deg to measure accelerations along the *X*-direction (Fig. 1). In addition, infrared emitting diodes embedded in a plastic coating were attached to the seat pad and the skin over the vertebral processes at L5 and T4 using two-sided adhesive tape, and were tracked using a three-camera system (OPTOTRAK/3200; Northern Digital) positioned 2 m behind the subject. The motions of those lumbar and thoracic markers were measured in all three directions to provide the corresponding 3D displacements of the seat and the spinal regions during each vibration exposure (Fig. 1). Accelerometer signals were low-pass filtered at 220 Hz for anti-aliasing prior to being sampled at 500 Hz using a VAX 4000 computer system, and motion tracking data were acquired at 200 Hz using an Optotrak Data Acquisition Unit.

In addition, for each subject, muscle activity in each spinal region was measured. EMG activity was measured using bipolar surface electrodes (Telemg; Bioengineering Technology Systems) placed symmetrically over the left lumbar (L3), right lumbar (L3), left thoracic (T9), and right thoracic (T9) paraspinal muscles (Fig. 1). A reference electrode was placed on the lower ribs of each subject on the right side of the back. EMG signals were preamplified 100× to reduce the effect of cable motion by improving the signal-to-noise ratio. EMG data were acquired at 500 Hz. In addition, an electromyographic amplifier with a high-pass filter set to 10 Hz and a low-pass filter 4th order Bessel filter set at 200 Hz was used for all EMG measurements to minimize motion artifact and prevent signal aliasing. On a separate day prior to vibration testing, the maximum voluntary contraction (MVC) for each subject was measured. The pelvis was stabilized during the contractions using two canvas cargo straps and a seat belt, and each subject wore a climbing harness attached to a force transducer at chest level (Model U4000, ±100 kg; Maywood Instruments, Ltd.). Force and EMG data were recorded while the subject performed a maximal isometric trunk extension in which each subject extended his trunk at the waist against the resistance of the chest harness. The force during extension was recorded during the 100% MVC. On the day of testing, immediately prior to the vibration exposure, each subject underwent EMG calibration trials by maintaining trunk isometric extension at 40% of the maximum force measured during the prior MVC testing by requiring the subject to sustain brief static contractions at 40% of the maximum force achieved during orientation using real time visual feedback. This was done in order to eliminate the potential muscle fatiguing effect of attempting an MVC prior to testing. The 40% maximum force contraction was used for EMG normalization in analysis. After data collection and prior to additional analysis of EMG signals, a 30 Hz high-pass 8th order zero phase digital Butterworth filter was applied to each subject's EMG data using MATLAB signal processing tools to remove artifacts induced by motion with a bandwidth of 2–18 Hz. Both the data collected during sinusoidal vibration testing and during MVC were filtered to ensure that each subject's EMG response was normalized to MVC EMG data with the same frequency band.

Each vibration exposure included a sinusoidal sweep ranging from 2 to 18 Hz, with a constant amplitude of 0.4 g and lasting for a duration of 70 s. The hydraulically driven mechanical shaker table was used to impose a sinusoidal sweep in each direction separately. During each test in the $X$-direction and $Z$-direction, acceleration, image, and EMG data were acquired. Only a single direction was tested on a single day, with at least 48 h of rest between the test days. For both test directions, each of the acceleration, displacement, and EMG signals was segmented into 2-s nonoverlapping epochs corresponding to distinct seat pad acceleration input frequencies in order to determine the transmissibility, peak-to-peak displacement, and muscle response. The spinal and seat pad acceleration data in both directions were high-pass filtered at 0.5 Hz and low-pass filtered using a 4th order Butterworth filter at 25 Hz, since analysis of the frequency response of the seat and spine accelerometers in both the $X$ and $Z$ directions indicated signal content primarily in the frequency range of seat inputs from 2 to 18 Hz. Prior to performing analysis of the accelerometer data, the accelerometer data for each subject were corrected for any movement of the skin relative to the underlying spinous processes using a linear modeling approach to identify a unique corresponding transfer function for each sensor and subject [15,16]. Those data corrections were only performed for vibration in the $Z$-direction since the $X$-direction accelerometers measured motion perpendicular to the skin's surface and the effects of skin and subcutaneous fat were assumed to be negligible based on prior work [13,16]. Specifically, before each subject underwent any vibration exposure, the skin surface directly below each spinal sensor was perturbed in the $Z$-direction and released, and the resulting acceleration response was recorded. The natural frequency and damping ratio were determined from the Fourier transform of that response, and those two parameters were used to calculate a skin transfer function specific to each sensor on each subject, as previously reported [16]. Following the vibration exposure, the skin transfer function for each sensor and subject were applied to the recorded accelerations. Transmissibility was then calculated using the corrected data.

For $X$-direction and $Z$-direction, transmissibility was calculated as the ratio of the root mean square (RMS) of the spinal acceleration to the RMS of the seat pad acceleration in the corresponding applied direction [17]. For each test, the frequency of the seat pad was calculated using the accelerometer data and spectral analysis in MATLAB. Although seat pad and spinal displacements were measured through the entire exposure frequency range, only the displacements larger than the system resolution of 0.5 mm (occurring during exposures between 2 and 5 Hz) were analyzed using OPTOTRAK MOTION analysis software (Northern Digital). The muscle response for each exposure was determined by calculating the RMS EMG value for each epoch. The muscle responses were normalized by dividing the RMS EMG values measured during testing by the RMS EMG value of the 40% MVC static calibration test and multiplying by 0.4 to scale to the 100% MVC recorded during subject orientation. The normalized RMS EMG ranged between zero, which indicated no muscle activity, and one corresponding to MVC contraction. The left and right RMS EMG values were compared using a paired $t$-test at each frequency and at each spinal level to test for differences in the responses based on side. Since there was no difference between left and right RMS EMG values ($p > 0.3$), bilateral symmetry was assumed and the left and right RMS EMG data for each spinal region were averaged together for each vibration exposure.

The acceleration, displacement, and EMG data from all subjects were averaged for each epoch, and the transmissibility was determined as a function of the input frequency. The resonant frequency for the lumbar and thoracic spinal regions was determined using the corresponding acceleration data in that region for each direction. The normalized EMG values for the lumbar and thoracic spinal regions were also compared to the input frequency in each direction of imposed vibration in order to identify those frequencies eliciting the greatest muscle activity. To measure the primary and associated motions of the spine during vibration and to provide context for the muscle response data, the peak-to-peak displacements in all three-dimensions also were calculated in each spinal region for each subject. For each direction of imposed vibration, separate one-way repeated measures analysis of variances (ANOVAs) ($p < 0.05$) compared the transmissibility and muscle response at each frequency to determine which frequencies produced the greatest responses.

## Results

The lumbar and thoracic spinal regions exhibited similar transmissibility responses for these young, male volunteers in the seated position exposed to vibration in both the $Z$-direction and $X$-direction over the range of frequencies (2–18 Hz) tested (Figs. 2 and 3). In general, the transmissibility of the lumbar spine was higher than that of the thoracic spine, but both regions exhibited the greatest transmissibility at frequencies below 5 Hz (Fig. 3). In particular, for vertically oriented ($Z$-direction) vibration, the transmissibility ratio was greater than 1 and significantly higher at 3 Hz ($p < 0.04$) and 4 Hz ($p < 0.02$) than at any other frequency in both the lumbar and thoracic spinal regions (Fig. 3). The peak transmissibility was observed at 4 Hz for the lumbar (transmissibility of $1.55 \pm 0.34$) and thoracic (transmissibility of $1.49 \pm 0.21$) regions. However, there was no statistical difference between the transmissibility at 3 and 4 Hz for the lumbar region or between 2, 3, and 4 Hz for the thoracic region (Fig. 3). Although a possible second dominant frequency was evident in the lumbar spine between 7 and 9 Hz, it was not statistically different from all other frequencies. For the $X$-direction vibrations, the transmissibility ratio in both spinal regions was significantly higher at 2 Hz

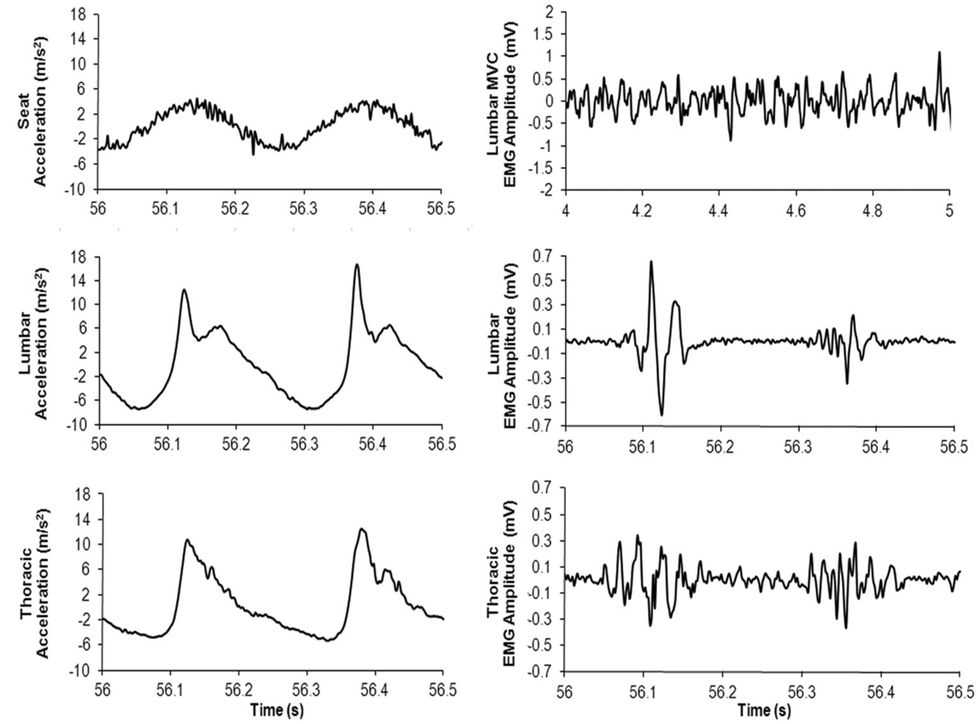


**Fig. 2 Representative data from a single subject undergoing a vertical seat vibration at 4 Hz, showing the acceleration signals acquired from the seat and at L4 (lumbar spine) and at T3 (thoracic spine). Also shown are the corresponding EMG signals acquired from that same subject during the MVC and in the lumbar and thoracic spinal regions during the vibration.**

($p < 0.001$) and 3 Hz ($p < 0.001$) than at any other frequency but never exceeded a value of 1 (Fig. 3). In addition, in both the lumbar and thoracic regions the transmissibility ratio was significantly different at 4 Hz ($p < 0.004$), but not higher than 2 Hz and 3 Hz

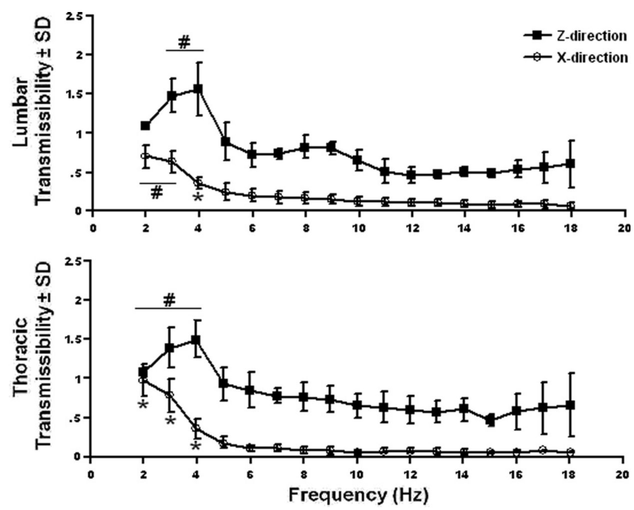


**Fig. 3 The mean transmissibility responses of the lumbar and thoracic spines for *Z*-direction and *X*-direction sinusoidal vibration sweeps imposed between 2 and 18 Hz. The transmissibility in the lumbar region in the *Z*-direction at 3 Hz and 4 Hz and in the *X*-direction at 2 Hz and 3 Hz is significantly different (#) from all other frequencies but not different from each other. In the thoracic region, the transmissibility in the *Z*-direction at 2 Hz, 3 Hz, and 4 Hz is significantly different (#) from all other frequencies but not each other. In addition, the transmissibility in the *X*-direction at 4 Hz in the lumbar region, as well as 2 Hz, 3 Hz, and 4 Hz in the thoracic region are significantly different (*) from all other frequencies.**

(Fig. 3). Even though the transmissibility was a maximum at 2 Hz for both the lumbar (transmissibility of $0.70 \pm 0.15$) and thoracic (transmissibility of $0.98 \pm 0.21$) spines, transmissibility was only significantly higher at 2 Hz ($p < 0.044$) in the thoracic region and there was no statistical difference between 2 Hz and 3 Hz in the lumbar region (Fig. 3).

The muscle responses exhibited similar patterns to the transmissibility responses in both spinal regions (Figs. 2 and 4). In fact, the peak muscle response in both spinal regions was detected at the frequency corresponding to the peak transmissibility in each region for both vibration directions (Figs. 3 and 4). The peak muscle response in both spinal regions was detected at 4 Hz for the *Z*-direction vibration and at 2–3 Hz for the *X*-direction vibrations (Fig. 4). In particular, for a *Z*-direction vibration, the muscle response at 4 Hz and 5 Hz was significantly higher ($p < 0.03$) than any other frequency in both the lumbar and thoracic spinal regions, but not different from each other in either regions. In response to an *X*-directed vibration, the muscle response at 2 Hz and 3 Hz was significantly higher ($p < 0.04$) in both spinal regions (Fig. 4). However, the muscle responses at 2 Hz and 3 Hz were not different from each other in either the lumbar or thoracic spines. Interestingly, in the lumbar region the muscle response at 4 Hz was significantly different ($p < 0.01$) than all other frequencies for the *X*-direction test, but was not higher than 2 Hz or 3 Hz.

The displacements of the motion markers on the spine for vibrations in the range of 2–5 Hz indicate that the primary motion of each spinal region was in the direction of the imposed seat pad vibration in all tests (Fig. 5). In both vibration directions, there was little motion in the directions that were perpendicular to the primary direction of seat pad vibration (Fig. 5). Although the spinal displacements along the primary axis (*Z*-axis) during the *Z*-direction test follow the seat in both spinal regions, there is a slightly greater difference in the lumbar region. Moreover, during a *Z*-direction vibration, the corresponding *Z*-direction spinal displacements were greater than the seat displacements at each frequency in both spinal regions (Fig. 5). However, the spinal

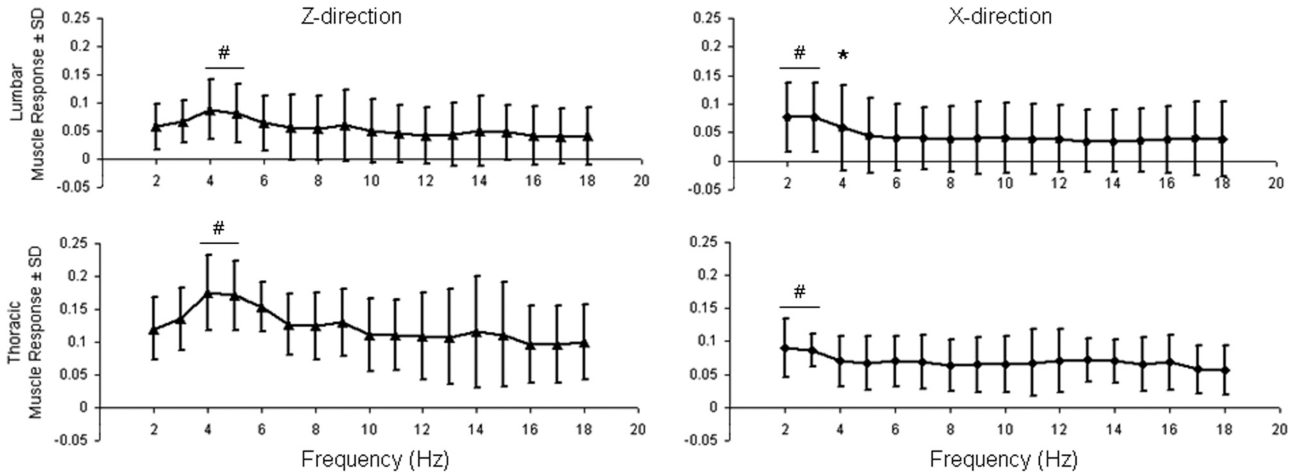


**Fig. 4 The mean muscle responses in the lumbar and thoracic regions for the *Z*-direction and *X*-direction vibration exposures between 2 and 18 Hz. In both spinal regions, the mean muscle activity at 4 Hz and 5 Hz for the *Z*-direction vibrations is significantly different (#) from activity at all other frequencies but not different from each other. Similarly, the muscle activity at 2 Hz and 3 Hz for both spinal regions during *X*-direction vibrations are significantly different from all other frequencies (#) but not different from each other. For an anteroposterior (*X*-direction) vibration sweep, the muscle response at 4 Hz in the lumbar region also is significantly different (*) from all other frequencies.**

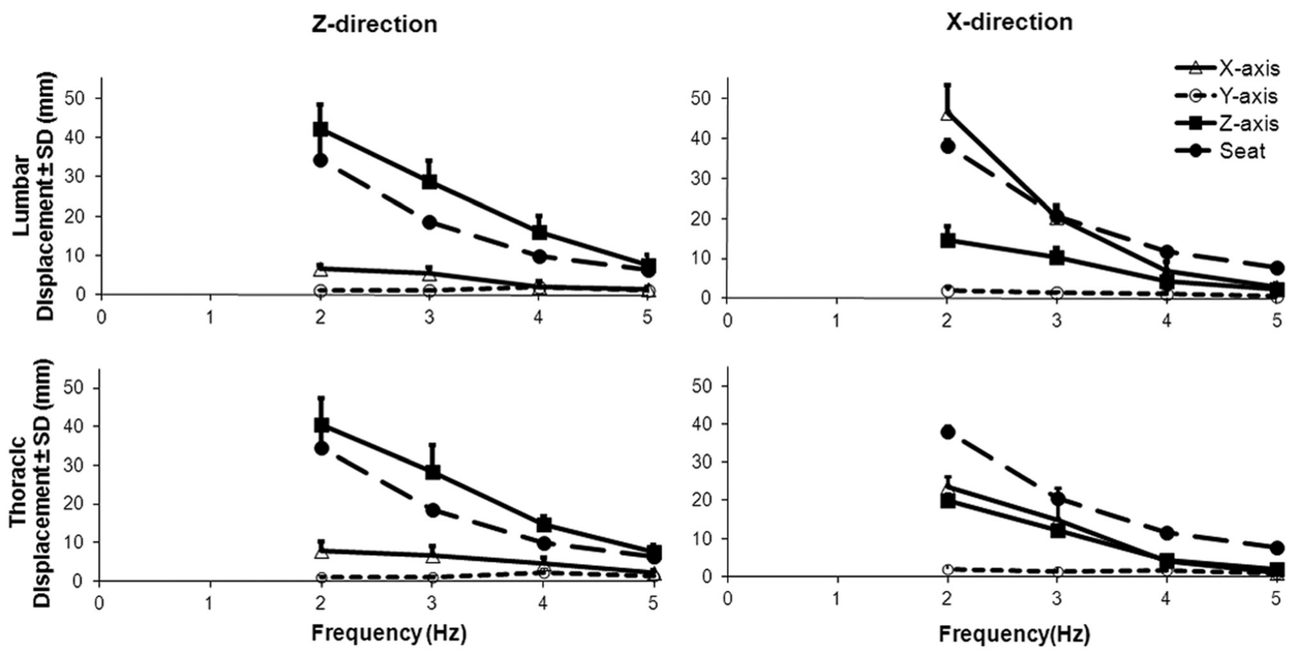


**Fig. 5 The mean peak-to-peak displacements along all directions of the seat and the markers for the lumbar and thoracic regions of the spine for vibrations in the *Z*-direction and *X*-direction between 2 and 5 Hz, showing the primary direction of motion is along the corresponding dominant motion of the seat, with laterally directed (*Y*-axis) motions being the smallest for all exposures**

displacements in the *X*-direction were only greater than the seat displacements in the lumbar region at 2 Hz (Fig. 5). For exposures directed along the *X*-axis, there was greater associated off-axis *Z*-direction motion in both regions of the spine than were observed for off-axis *X*-direction motion during the *Z*-direction exposures (Fig. 5). For both directions of seat vibration, the lateral motions remained negligible (Fig. 5).

## Discussion

This study found that the peak transmissibility of the lumbar (2–4 Hz) and thoracic (2–4 Hz) spinal regions occur at similar frequencies for the seated human regardless of the direction of the imposed vibration (Fig. 3). Moreover, the peak of muscle activity was detected at frequencies at or below 5 Hz for both directions (4–5 Hz in the *Z*-direction and 2–3 Hz in the *X*-direction) (Fig. 4). Together, these findings suggest that vibration frequency is an important factor in determining the biomechanical and physiological responses of the seated human to whole body vibration.

The resonant frequency for a vertically vibrated seated volunteer in this study was found to be similar in both the lumbar (3–4 Hz) and thoracic (3–4 Hz) spinal regions (Fig. 3). This resonance is comparable to other reports, in particular for the lumbar spine for which the resonance in the vertical direction has been reported to be between 4 and 6 Hz [6,7]. The transmissibility in both spinal regions is greater at 4 Hz ($1.55 \pm 0.34$ lumbar; $1.49 \pm 0.21$ thoracic) than 3 Hz ($1.47 \pm 0.22$ lumbar; $1.39 \pm 0.26$ thoracic), although this difference is not significant (Fig. 3).

Interestingly, a second dominant frequency has been reported in seated humans between 8 and 10 Hz [7]; this relative increase is also evident in our study for the lumbar spine around 8–9 Hz (Fig. 3) but it is not significantly higher than other frequencies.

The transmissibility response for seat vibrations in the anteroposterior direction (*X*-direction) is also similar between the thoracic and lumbar regions but peaks at lower frequencies than does the transmissibility response to vertical vibration (Fig. 3). The peak transmissibility in the lumbar (2–3 Hz) and thoracic regions (2 Hz) falls within in the range of frequencies (0.7–5 Hz) that produce the largest apparent mass measured at the seat during whole body vibration [9,10]. Although the peak transmissibility ratio was seen at 2 Hz in both the lumbar ($0.70 \pm 0.15$) and thoracic ($0.98 \pm 0.21$) spinal regions, it never reached a value greater than 1 (Fig. 3). Our data demonstrate that it produces lower transmissibilities than a vibration along the spine's long-axis. Of note, the muscle responses were also generally lower for vibrations in the *X*-direction (Fig. 4). This study did not impose vibrations at frequencies below 2 Hz, so it is not possible to definitively assert that the maxima at 2 Hz for the *X*-direction sweeps are indeed a peak. Further work is needed to explore the transmissibility between 0 and 2 Hz.

The peak muscle response in both spinal regions also occurs at the primary peak frequency in both directions of vibration (*Z*-direction at 4–5 Hz; *X*-direction at 2–3 Hz) (Figs. 3 and 4). These peak responses are consistent with those frequencies reported to have the highest discomfort levels during vertical vibration (4 Hz and 8 Hz) and in anteroposterior vibration exposures (2 Hz) [5]. In particular, a peak muscle response for a vertical seat vibration in the lumbar (4 Hz) and thoracic (4 Hz) spines corresponds to the peak of the transmissibility ratio that occurs in the lumbar (3–4 Hz) and thoracic (2–4 Hz) spines (Figs. 3 and 4). Similar trends were also observed in the *X*-direction (Figs. 3 and 4). For these experiments, the exposure protocol imposed separate unidirectional accelerations in each of the vertical (*Z*) and anteroposterior (*X*) directions (Fig. 1), which does not fully simulate the more complicated and multidimensional "real world" exposures that train and bus conductors and helicopter pilots undergo [2–4]. For frequencies lower than 4 Hz, the multidirectional vibration of a seated human has been shown to increase the mechanical impedance at the contact between the seat and the trunk relative to that of a uni-axial anteroposterior exposure; no such difference occurs with respect to vertical vibration [9]. Although only inferential based on the uni-axial data in our study and the reports in the literature, it would be expected then that our findings may underestimate the severity of these responses if there were multidirectional seat vibrations imposed. The seat in this study did not have a backrest, which has been shown to alter the biomechanical response [18], but drivers and occupants of tactical ground vehicles use a range of seating configurations, with and without backrests [13]. Posture is likely to affect muscle activity. In fact, a previous study measuring the muscle response at L4 showed that EMG activity of erector spinae increases during an anterior lean compared to a neutral position or posterior lean during vibration [19]. Nonetheless, the goal of our study was to simulate those frequencies that are experienced by seated occupants in ground vehicles and further studies are needed to define the biomechanical and muscle responses to such inputs.

The muscle response data exhibit substantial variability, especially in the lumbar region (Fig. 4). The muscle activity recorded in our study captures the total response using surface electrodes and groups the responses on the right and left sides (Figs. 1, 2, and 4). In addition, this technique does not enable separation of individual muscles to determine which are active. Therefore, the data only distinguish when muscles are active and responding at certain frequencies. Although there has been some debate in the literature about whether EMG responses recorded during whole body vibration are motion artifacts or physiological responses [20,21], recent studies indicate that there is a cyclic physiological EMG response to vibration due to excitation of the stretch reflex [21,22]. In addition, in this study all EMG signals were preamplified and high-pass filtered in an effort to reduce any motion artifacts. This technique has a further limitation in that it does not enable identifying the line of action along which a muscle may be acting. While it is possible that the specific muscles that are activated by the vibration may also vary between subjects, the evoked muscle activity for each subject was normalized to the MVC for each subject. A linear approximation was used to normalize RMS EMG from zero to one (related to MVC muscle contraction), which may be an incorrect approximation and may have an effect on the muscle response presented. Regardless, this methodology accommodates any differences, such as muscle strength, that may be present between individuals. The EMG signal was low-pass filtered at 200 Hz, which may cause part of the EMG signal to be truncated. As such, plots of the EMG frequency content (data not shown) revealed that the frequency content peaks around 25–30 Hz and is largely reduced by 100–150 Hz, suggesting no signal was lost by using the 200 Hz low pass filter. The original experiments did not specify the distance between electrodes [13], which can cause a shift in frequency content [23]. However, the EMG frequency content peaked around 25–30 Hz for all subjects, suggesting a substantial interelectrode distance. Given these limitations, any correlation between spinal and muscle activity may be oversimplified. Nonetheless, this study did determine an obvious peak of muscle activity that corresponds well with the resonant frequency during whole body vibration (Figs. 3 and 4).

The coordinated responses between the spinal biomechanics and muscle activity imply that particular frequencies may require more muscle activity to stabilize the spine. The increase in muscle activity may be explained by the associated increased spinal motions that occur at those same frequencies, particularly in the vertically oriented vibrations (Figs. 4 and 5). Both the lumbar and thoracic motions along the *Z*-axis exceed those of the seat (Fig. 5), which also corroborates the resonance frequency being in that range. In contrast, for anteroposterior vibrations, the associated primary (*X*-axis) motions are consistently lower than or the same as the seat motion (Fig. 5), suggesting vibration in the *X*-direction may be adequately dampened. Although it is possible that the seat and bean-bag cushion could alter motions in the *X*-direction due to dampening from the sliding of individual granules in the cushion, it is likely that any dampening may be due to the anatomy and physiology of the spine. Of note, posture has been shown to play a significant role in the muscle response and also vibration response; a subject exposed to vibration in a relaxed posture exhibits greater spinal motion at lower frequencies than when in a tensed erect posture [24]. Since this study tested subjects in a relaxed posture, the muscle response, spinal motions, and transmissibilities reflect that condition, which may vary across subjects. Nonetheless, the transmissibility responses exhibit little variation (Fig. 3) and a relaxed posture was used to simulate the typical occupant setting.

The amount of off-axis motion along the *Z*-axis is not negligible during the anteroposterior vibration exposure (Fig. 5). This may be caused by postural changes, such as bending forward or backward in response to the anteroposterior vibration. Of note, the Optotrak system is only able to measure off-axis motion that is 5 mm or greater, which may also explain the small off-axis displacements detected in the nondominant directions of exposure. Accordingly, it is possible that the displacements measured in the *Y*-direction, which are lower than that limit, may be underestimating the true lateral motion (Fig. 5). However, it is unlikely that motions in those directions are major contributors to the subjects' responses.

Although muscle fatigue and pain were not assessed here, the increase in muscle activity observed at the resonant frequency (Figs. 3 and 4) suggests a possible mechanism by which muscle fatigue may be more likely (or even more severe) and may lead to the development of pain [3]. The transmissibility response in the vertical direction is consistently above 1, but this is not the case for vibration in the anteroposterior direction (Fig. 3). Since a transmissibility ratio greater than 1 is considered a resonance frequency because an amplification of the input occurs [7], these data suggest that spinal loading along its axis may have injurious effects on the spine and lead to physiological dysfunction. However, additional work is needed to define whether the injury risks vary with direction of loading. Indeed, increased muscle activity is similarly observed in these cases and suggests a potential mechanism for injury either to the muscle directly or to the spine indirectly. Studies incorporating other physiological outcomes and more complex loading scenarios may provide important insight into the pathobiomechanical mechanisms of the common painful injuries that develop from repeated whole body vibration. In fact, a study using an animal model has reported changes in pain-related neuropeptides and pain at the corresponding resonant frequencies [25]. However, a rodent model of vibration along the spine also reports persistent pain to be generated at a frequency (15 Hz) that is not the resonance [26], suggesting that the relationship between whole body vibration and pain may be more complex. Regardless, having identified certain frequencies that increase the transmissibility and corresponding muscle responses for seated humans may provide important information for designing better seats to minimize injury and discomfort for seated occupants in at-risk populations. Future work coordinating animal and human biomechanical studies that also integrate neurophysiological assays may improve our understanding of whole body vibration injury and the relationship between vibration exposures and tissue biomechanical and physiological responses.

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