# The Effect of Whole Body Horizontal Vibration in Position Sense and Dynamic Stability of the Spine

**BY**

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# Abstract

In many workplaces, workers are exposed to whole body vibration which involves multi-axis motion in fore-aft (x axis), lateral (y axis) and vertical (z axis) directions. In previous studies, our laboratory has found changes in biomechanical responses such as response time and position sense with exposure to vibration in single vertical direction. The objective of the current study was to investigate the effect of whole body, horizontal vibration on proprioception and sudden loading dynamics and to compare these results with the previously studied whole body vertical vibration experiment. Both position sense test and sudden loading test were performed in three conditions: a pre-exposure condition (pre), a post-washout condition (postw) and a post-vibration condition (postv). Subjects were exposed to the whole body horizontal vibration frequency of 5 Hz and constant acceleration of 0.284 RMS ($m/s^{-2}$) for 30 minutes. Absolute reposition sense error increased slightly after vibration exposure (relative to after quiet sitting (postw)), although the results were not significant. Times to peak muscle response and flexion magnitude were also increased after horizontal vibration exposure, suggesting a decreased stability of the spine, but again these results were not significant. Compared to the previous study of vertical whole body vibration, the effects of horizontal vibration in this study were small and not significant. This may be due to differences in the transmissibility of vertical and horizontal vibrations at the 5 Hz frequency. These results would suggest that horizontal vibration may be less of a factor in whole-body vibration induced injuries. This work was supported by University of Kansas Transportation Research Institute Grant Program.

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## List of figures

# List of Tables

# 1.0 Introduction

## 1.1 Low Back Pain

Low back pain (LBP) is the leading cause of industrial disability in the population under the age of 45 years [1]. The total cost of LBP in the U.S. is as much as 80 billion dollars per annum [2] . In the United States, one million back injuries occur per year, 100 million work days are lost each year, and LBP accounts for 20% of all work related injuries [1]. Occupational low back pain (LBP) is a great burden for industry and medicine. Kelsey & White [3] found 2% of workers in the United States have a compensable back injury. The greatest expense is from 25% of the cases, and as the LBP duration increases, the total cost accelerates. LBP is responsible for 21% of all compensable work injuries and 33% of the cost. Medical costs are 33% of the total and disability payments are the remainder [4, 5].

Pain in the soft tissues of the low back is extremely common among adults. In the United States, the National Arthritis Data Workgroup reviewed national survey data showing and found that each year some 15% of adults report frequent back pain or pain lasting more than 2 weeks [6]. Back pain is widespread in many countries, and is associated with substantial financial costs and loss of quality of life. In Canada, Finland, and the United States, more people are disabled from working as a result of musculoskeletal disorders (MSDs), especially back pain, than from any other group of diseases [7]. Although complex social and economic forces may account for part of this increase, there is suggestive evidence that working in industrialized environments may be of significance.

### 1.1.1 Risk Factors

The risk factors leading low back pain has have been investigated in different disciplines of biomechanics, psychology, psychosocial, physiology, genetics, organizational psychology and rehabilitation [8]. Epidemiologic studies clearly indicate the role of mechanical loads on the etiology of occupational LBP. McGill proposed that when biomechanical load imposed on a tissue exceeds the tissue tolerance level, injuries and disorders initiated [9]. Occupational exposures such as lifting, particularly in awkward postures; heavy lifting; or repetitive lifting are associated with LBP [4]. Gallaghar et al. found that lumbar compression increased as a result of kneeling posture [10]. McGill et al. suggested that a fully- flexed spine results in lumbar extensor muscles ineffective to support anterior shear force, which is related to low back injury [11]. Standing and sitting have specific advantages and disadvantages for mobility, exertion of force, energy consumption, circulatory demands, coordination, and motion control [4]. Workers in a driving environment are often subjected to postural stress possibly leading to back, neck, and upper extremity pain [12].. Activities involving vibration (driving vehicles and operating power tools) have been linked to increased reports of back pain [13]. Epidemiological studies as well as biomechanical research has given evidence for an elevated risk of health impairment of the lumbar spine and the connected nervous system due to long duration exposure to whole body vibration of high intensity [14]. Although ergonomic and personal risk factors are predictive of LBP, psychosocial factors can also influence LBP disability [15]. Other important patient factors are obesity, physical fitness, smoking history, height and pregnancy, age, gender which might affect tissue tolerance.

## 1.2 Whole Body Vibration and Low Back Pain

The WBV exposure is defined as the vibration transmitted to the whole body from a vibrating seat or standing platform and measured at the interfaces between the machine and the operator i.e. at the driver's seat. The spinal health risk may arises from a mechanical damage of anatomical structures due to forces acting on those structures (internal load) [16]. Increases in these these internal forces can be a direct product of the vibration or may be an indirect result of the vibration altering the dynamic control and motion of the lumbar spine. Professional drivers have been found to be at high risk of developing such low back injuries due to prolonged sitting and vibratory exposure [17-19]. A number of studies have examined these problems in drivers of different types of vehicles such as from trucks, urban transit buses, taxis and rally cars [3, 19, 20]. Backman et al. found that 40 % of the bus drivers studied had LBP, with the occurrence increasing with age [21]. Schmidt et al. compared drivers of heavy trucks and bank employees and found that 75% of the truck drivers had pathological changes of the spine compared to 61.1% of the bank employees [2]. Driving different types of vehicles may impose different stresses to the body, as the driver's seat, control mechanisms and vibration generated may vary [19]. Intense, long term, whole body vibration can adversely affect the spine and can increase the risk of low back pain. Christ and Dupius found that, of those with more than 700 tractor driving hours per year, 61% had pathologic changes of the spine; of those with 700-1200 hours, 68% were affected; of those with greater than 1200 driving hours, 94% were affected [22]

Literature on whole body vibration has related occupational vibration exposure to a high risk of LBP [23]. Frymoyer et al. found that patients with LBP tended to have occupations that involved vehicular vibration [24]. Magnusson et al. studied bus and truck drivers both Sweden and USA and found a significant correlation between low back pain and history of whole body vibration exposure [25]. Bongers et al., demonstrated that, workers exposed to a high degree of vibration at work had a 32% increase in back-related disability compared to a control group[26]. Cremona et al. reported that the occurrence of low back pain is a 70% in drivers of heavy earth moving equipment [27, 28].

Efforts have been made to quantify the effect of vibration magnitude, frequency, duration and other confounding factors on low back disorder in terms of dose response relationship. The vibration dose value (VDV) provides a convenient measure for assessing the severity of vibration on human health. The VDV is given by the fourth root of the integral with respect to time of the fourth power of the frequency weighted acceleration and it can be expressed mathematically as follows:

$$\mathrm{VDV}= \int_0^T a^4(t)\, dt$$

Some studies have examined the direct correlations of vibration magnitude and duration of vibration dose value and LBP prevalence [12]. From a survey of occupational drivers (of many types), Schwarze et al. [29] concluded that with increase vibration dose, LBP incidence increases. Robb et al. suggested that manual material handling and seat discomfort in truck drivers affect dose-response relationship [30].

### 1.2.1 Frequency Dependence of Whole Body Vibration

#### Vertical vibration and frequency response

Experimental whole body vibration studies have demonstrated a consistent pattern for the vertical response of the seated human body exposed to whole body vibration. At resonance aA system oscillates at maximum amplitude at certain frequencies, known as the system's resonant frequenciesy. Resonant frequencies to vertical seat vibration have been reported to occur between 4 to 6 Hz. This is usually attributed to the upper torso vibrating vertically with respect to the pelvis [2] At resonant frequency, there is greater movement of the spine and studies of comfort with vibration exposure have indicated these frequencies are the most uncomfortable to be exposed to [31-35].

Griffin et al. [32] suggested that, with vertical vibration exposure the equivalent peak dip of the comfort contour (a measure of the maximum amplitude of vibration that is comfortable for a given frequency) falls in the range of frequency from about 2 or 3 Hz to about 5 or 6 Hz. Dupis et al.[33] and Jones and Saunders [31] showed that this comfort contour rises in proportion to frequencies frequency above 5 Hz. In sitting subjects, resonance occurs at the shoulders at 5 Hz and also, to some degree, at the head and significant resonance between shoulder to head can occurs at approximately 30 Hz [4]. Using accelerometer and pins implanted in the lumbar region Dupuis, Panjabi et al., and Pope et al. all demonstrated that the resonant frequency in the human lumbar region of the vertically vibrated seated operator was approximately 4.5 Hz [22] . Finally Pope et al. found that the greater intervertebral rotations and translations occurred at 5 Hz, confirming the effect of natural frequency [2, 34].

## Horizontal vibration and frequency response

In many travel environments (e.g., road vehicles, off-road vehicles, trains, boats, aircraft) there are substantial low frequency (i.e., 0.1–2.0 Hz) motions that may influence the comfort of passengers and operators. A study over the frequency range 0.5–300 Hz suggested that the discomfort arising from exposure to acceleration in the horizontal axes was independent of both frequency and axis between 0.5 and 1.0 Hz [36]. In the frequency range 1–30 Hz, Rao and Jones [37] concluded that sensitivity to both lateral and fore-and- aft acceleration was greatest at 2 Hz, with decreasing sensitivity above and below this frequency. Corbridge and Griffin [38] investigated discomfort caused by lateral vibration at frequencies between 0.5 and 5 Hz and found that lateral acceleration caused most discomfort at frequencies between 1.6 and 2.0 Hz, with a gradual reduction in sensitivity at lower frequencies. Finally, Fairley et al. [39] assessed the response characteristics of seated occupants using apparent mass frequency functions in the fore-and-aft direction. The apparent mass measure is a ratio of the horizontal force transmitted at the occupant and seat interface to the acceleration measured between the occupant and seat interface. The results without a backrest exhibited two resonance modes in the fore-and-aft direction, with a primary resonance peak at 0.7 Hz and a secondary resonance peak at 2.5 Hz.

### 1.2.2 Whole Body Vibration and Muscle Fatigue

The development of muscle fatigue can be studied by analyzing the electromyographic (EMG) signals from a contracting muscle. Several studies have reported an decrease in mean or median frequency of the EMG signal during muscle fatigue [40-43]. Pope and Wilder et al. demonstrated that after an exposure of 5 Hz whole body vibration and at 0.2 g r.m.s. acceleration. The, the mean frequency and r.m.s. value of the EMG signal decreased with time after the exposure to vertical whole body vibration [43]. These researchers also found a significant increase in time to response of vertebral lumbar muscles to a sudden unexpected load applied to the upper trunk. Thus, truckers who unload their truck right away after longer driving might have a higher risk of soft tissue or muscle injury due to delays in response and a resulting overload of the soft tissues. This could beis increased if the vibration is asymmetric or the specimen spine is preloaded. Seated whole body vibration in a position that ensured muscular activity of the erector spinae muscles caused faster and more pronounced muscular fatigue in the lumbar erector spinae muscles when compared to the absence of vibration [25]. Zimmermann et al. [44] conducted an experiment where subjects were exposed to 1 $m/s^2$ R.M.S. sinusoidal vibration at frequencies 4.5, 5, 6 8,10,12 and 16 Hz at L1 and L3 level Erector Spinae muscles.. They suggested that the response of the muscular system is dependent on both pelvic orientation and vibration at 4.5- 6 Hz of vibration frequency.

Both static sitting and seated whole body vibration caused increased height loss in subjects, suggesting increased spinal loading [4]. These factors would likelymight contribute to more rapid degenerative changes in the lumbar spine. Pope et al. [43] also commented that when the spine is loaded axially for a prolonged period, the back muscles could become fatigued and the discs are being compressed, this would result in an inability poorer condition to sustain larger loads; thus when there is any suddenly applied load such as a sudden stopping of the vehicle, there may be an increased risk of sustaining serious injuries to the spine.

## 1.3 Low Back Pain and Stability of Lumbar Spine

Biomechanical etiology of low back pain is not precisely known. Over the past 40 years, both field surveillance and laboratory studies have focused on the relationship between back injury and compressive force experienced by the lumbar spine. However, epidemiologic research examining the association between spinal compression and occupationally- related low back pain, has been unable to demonstrate a strong complete causative relation between the two. A number of authors have proposed dynamic stabilization of the low back pain as a potential factor in low back disorder [45, 46]. One important function of the lumbar spine is to support the upper body by transmitting compressive and shearing forces to the lower body during the performance of everyday tasks. However, the isolated thoracolumbar spine buckles under compressive loads exceeding 20 N and the lumbar part of the spine buckles under approximately 90 88 N. In vivo, a spine may experience compressive loads ranging from up to 6000N for more demanding everyday tasks and up to 18000 N during competitive power lifting, far above the 90 N buckling load, which clearly exceed the tissue tolerance limit [45]. To maintain these compressive loads the spine musculature and neuromotor system muscle stabilize the spine. Even the weight of the upper body can exceed the 90 88 N buckling load. This has led several researchers to consider stability as a potential factor in the risk of low back injury.

Stability is one of the most fundamental concepts to characterize and evaluate any system. In terms of the spine, stable behavior is critical for the spine to bear loads, allow movements and at the same time avoid injury or pain. Several researchers have applied stability analysis to the spine by evaluating the potential energy of the system and have yielded a number of important insights including the requirement for stiffness from trunk muscle to maintain spine stability [47].

Stability was defined by McGill (2001) as a function of potential energy which is a function of stiffness and storage of elastic energy. He suggested that active muscle acts like a stiff spring and the greater the activation of the muscle, the greater the stiffness. The motor control system is able to control stability of the joints through coordinated muscle coactivation and/ or by placing joint in positions that modulate passive stiffness [48]. Stability is an estimation of musculoskeletal injury tolerance represented by euler buckling or systems analyses of the neuroanatomic structure [45, 46, 49, 50]. When the spine is stable under a given load, the small neuromuscular or vertebral movement errors are automatically corrected without tissue damage. Conversely, if the spine is unstable, then a small neuromuscular error can be amplified by the biomechanical forces, causing sudden undesired vertebral motion [46]. These buckling movements may impose acute strain on intervetebral tissues or stress on the nerve root foramen. To control stability, well-orchestrated neuromuscular control is necessary [48].

Panjabi et al. depicted the concept of spinal stability by differentiating in terms of “Mechanical stability” and “Clinical stability”. Mechanical instability defines the inability of the spine to carry spinal loads without buckling type deformation, while clinical instability includes the clinical consequences of neurological deficit and/ or pain. The spinal stabilizing system of the spine was conceptualized by Panjabi to consist of three subsystems: spinal column providing intrinsic stability; spinal muscles, surrounding the spinal column, providing dynamic stability; and neural control unit evaluating and determining the requirements for stability and coordinating the muscle response. Under normal conditions the three subsystems work in harmony and provide the needed mechanical stability. The various components of the spinal column generate information about the mechanical status of the spine, such as position, load and motion of each vertebra, in a dynamic fashion. The neural control unit computes the needed stability and generates appropriate muscle pattern for each instance. Thus neuromuscular control of spinal stability may play a significant role in the etiology and prevention of low back pain [51].

According to Crisco et al. the critical buckling load for the ligamentous lumbar spine column alone is only 88 N [36], which is much smaller than the estimated in vivo spinal load of 1500 N and above [12]. Activated muscles surrounding spinal columns, therefore, must acting as a guy wire to stiffen the spine and provide significant spinal stability as well as increase the critical load of lumbar spine columns up to 2600 N [34, 39]. The muscle – tendon subsystem or muscle stiffness is therefore a very important contributor in spinal stability. A stiff trunk can constrain a movement of each vertebra and help it withstand forces placed upon it. Increase in muscle stiffness is associated with increased in muscle activation [39]. Reduced muscle activation is a risk factor of low back pain. Using a detailed model of the spine, and incorporating experimental actual electromyographic (EMG) data, Cholewicki and McGill found that the stiffness and stability of the spine was increased during more demanding tasks, but diminished during periods of lower muscular activity, such as picking up a pencil from the floor [45].

Additional evidence for spinal stability as a mechanical injury mechanism comes from epidemiology. Brumagne et al have shown that low back-injured patients have reduced proprioception, a component of neural feedback [52]. Marras et al. and Granata et al. have shown that experienced workers have greater co-contraction, which would increase their spinal stability [53, 54]. Orishmo et al have also shown that under unstable loading conditions, subjects showed increased co-contraction, increasing their stability [55]. Finally individuals with low back injuries have also been found to have delayed trunk muscle response to sudden loading, suggesting decreased spinal stability.

The various components of the spinal column generate information about the mechanical status of the spine, such as position, load and motion of each vertebra, in a dynamic fashion. The neural control unit computes the needed stability and generates appropriate muscle pattern for each instance [56]. Thus neuromuscular control of spinal stability may play a significant role in the etiology and prevention of low back pain [55].

## 1.4 Proprioception- A Component of Spinal Stability

Proprioception is an important component in dynamic stabilization of the spine. Any reflex or voluntary response to a sudden perturbation requires first a detection of the change in joint orientation. This detection is provided by the muscle spindle organs, ligamentous receptors, golgi tendon organs, cuteneous sensors and vestibular organs that play a role in trunk position sense and proprioception [57, 58].

There are different types of sensory receptors. Exteroceptors are the conscious senstation that respond to light, sound, odor, heat, touch, pain, acceleration and so on. The receptors which are not typically used for conscious sense of the external world, and provide feedback of the body internally are known as proprioceptors. Proprioception is the measure of subjects ability to detect an externally imposed passive movement or the ability to reposition a joint to a predetermined position. This afferent neural information originates from joint muscles, tendons and deep tissue mechanoreceptors and frequently used in rehabilitation. Proprioceptive information is processed at three different motor control centers of the central nervous system (CNS) – spinal, lower brain (brainstem, cerebellum) and cortical levels [59-62]. The CNS then integrates these afferent signals to regulate motor commands controlling voluntary muscle activation that contribute to joint stability [62] The proprioceptors associated with muscles are the spindle organs and the golgi tendon organs. They modify and even control many aspects of muscle behavior. Muscle spindle organs are known as stretch receptors because they are ordinarily attached at both ends of the main muscle mass and experience the same relative length changes as the overall muscle, and can therefore work as a strain gauge [63]. Another important proprioceptor is the golgi tendon organ. It is found very close to the junction between tendon and muscle fibers. The golgi tendon organ acts like a force transducer for the muscle [63].

Proprioceptive information is responsible for generating preprogrammed motor commands to achieve a desired outcome [64-66]. In addition, these same sensory organs may be connected at spinal cord levels to reflex loops, allowing quick corrective actions to take place. [63]. The muscle spindle organs are considered to be the primary sensors for position and movement detection and sensitive to change in muscle and velocity of the muscle length changes [67, 68]. The error in proprioception may be increased due to dysfunction of muscle spindle organs which in turns results in spine instability and low back injuries.

### 1.4.1 Effect of Vibration on Proprioception

Proprioception plays an important role in appropriate sensation of spine position, movement and stability. Previous studies [69-72] have observed that increased trunk reposition error a measure of person's ability to replicate a predetermined position, occurs in the low back pain population. Altered spinal reposition sense has been reported to be related to factors of age, vibration, muscle fatigue, asymmetric lifting posture, gender, trunk position as well trunk motions[72].

Muscle Muscle-tendon vibration and microneurography studies have demonstrated a major roll of muscle spindles in proprioception. Muscle- tendon vibration is a powerful stimulus of the muscle spindle primary afferents. The effect of vibration is to introduce a bias into the muscle spindle output. Inglis et al. illustrated [76] that mechanical vibration (95 Hz, 2 mm) when applied to the antagonist muscle or muscle that is stretched by voluntary movement, causes a systematic distortion of human position sense. The vibrated muscle is usually perceived to be longer than it actually is [73].

Roll et al. conducted research in the area of vibration and its effects on proprioception on extremities [74]. Mechanical vibrations applied to the arm muscles of human subjects, with no visual information available, induced the illusory sensation. This illusory sensation is the sensation that the muscles are lengthened, when they are actually not during the vibration exposure. It is known that vibration frequencies ranging from 20 to 120 Hz result in lengthening illusions during vibration [74]. These lengthening illusions may lead to proprioception alterations.

Burmagne et al. observed a significant muscle lengthening illusion during multifidus muscle vibration, which caused the subjects to undershoot their target position [67]. In another research, it was observed that patients with low back pain had a significantly lower proprioceptive acuity than the controlled healthy volunteer [83].

Proprioception is modified not only during the exposure of vibration; it remains altered even after the stimulus is removed. Zhang et al. [75] and Arashanapalli et al. [76] demonstrated that the vibration-induced errors persist after the termination of the vibration without the illusory sensation. According to Wierzbricka et al. [77], it was not only during vibration that sensory information was changed. Proprioception remained altered for certain amount of time after a vibration stimulus was terminated. The sustained sensory input evoked by vibration has a powerful post-vibration effect on the motor system at the postural level.

A recent experimental study demonstrated an association between proprioception and neuromotor response to a sudden perturbation, when vibration was applied to the paraspinal musculature. Increased delays in neuromotor response and increased flexion after a sudden perturbation were observed during paraspinal muscle vibration, specifies the role of proprioceptive system in dynamic stabilization [76, 80]. In another study, paraspinal muscle vibration has been shown to increase center of pressure (COP) path length in seated sway, demonstrates losses in dynamic stabilization of the trunk [84].

### 1.4.2 Proprioception Test- Detection of Resposition Sense Error

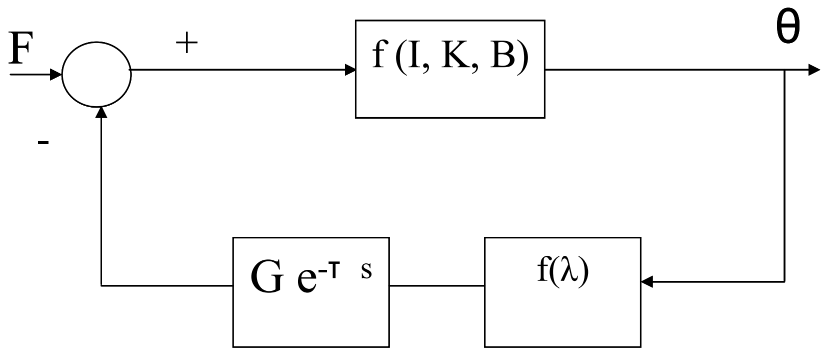
Different testing methods including kinesthesia testing, sense of effort testing and joint position testing were discussed to evaluate proprioceptive sensation. Kinesthesia testing is a method of proprioception testing that focuses on an individuals ability to detect movement of peripheral body segments. This can be conducted in their active or passive modes. Sense of effort testing focuses on an individual's ability to replicate torque magnitudes produced by a group of muscles under varying conditions [81]. Wilson et al. developed a low back reposition sense testing paradigm to examine lumbar reposition sense error in different studies [78]. Errors in the ability to sense joint position can be measured using a reposition sense protocol in which accuracy in joint positioning is measured by positioning the joint at a certain angle then asking the subject to reproduce that angle.

Different measures of reposition sense error suggested that a number of low back pain factors including whole body vibration, a history of low back pain and trunk posture will increase reposition sense error [69, 78, 80]. Using the reposition sense test protocol

Gade et al. and Wilson et al. found that higher reposition error also occurs in flexed and asymmetric posture [79]. In another study, Arashanapalli et al. demonstrated the increased position sense error both during vibration exposure and instant after vibration exposure [76]. In addition, in a recent study of vibration induced changes in position sense, Li and Wilson demonstrated that increase errors in reposition sense with vibration exposure was associated with increase delays in neuromotor response and increased flexion in sudden loading experiments [80].

### 1.4.3 Computational Model of Proprioception and Stability

A simple inverted pendulum model for trunk was developed by Wilson et al. [82] and effect of position sense on dynamic stability and the response to a sudden perturbation was studied . The model considered trunk motion as a function of trunk inertia (I), intrinsic stiffness (K), damping (B), response magnitude (G), response delay (τ) and detection threshold (λ) [Figure 1]. In this model, Wilson et al. predicted that as the response threshold increases (i.e. decrease in proprioception), the time delay and response magnitude increased which in turns decrease the stability of the trunk. The model also predicted that increased proprioception would result in increased flexion magnitude, and thereby increased muscle activity to stabilize the spine. Based on the results from the model, and the experimental findings, researchers proposed that such increased time delays for trunk muscle activation could impair trunk stiffness required for torso stabilization.




F: Trunk disturbance force

I: Trunk inertia

K: Muscle stiffness

B: Damping

G: Response magnitude (Gain)

$\tau$ : Time delay

$\lambda$ : threshold response function

**Figure 1. Control theory model of torso motion (θ) in response to a disturbance force (F) developed by Wilson et al. Stability of the system is a function of trunk inertia (I), muscle stiffness (K). damping (B), response magnitude, time delay and detection threshold. When the response threshold increases (decrease proprioception), time delay and response magnitude increases, which in turns decrease stability of the spine [80, 82].**

## 1.5 Response to Sudden Load and Stability of the Spine

Sudden load is commonly assessed experimentally as a research measurement of spinal stability. Material handling has been cited as one of the most frequent causes of back injuries, although sudden (forceful, unexpected) movements have been associated with most costly injuries [5, 83]. Sudden exertion can occur due to slips or falls, lifting of unstable loads or failed two- person lift. Nurses handling patients[84] and physical therapists working with patients are especially prone to this problem [85]. Drivers unloading their trucks or cars are exposed to catching materials when loading or unloading goods. There are other less obvious unexpected sudden load conditions, many occurring while sitting: vertical impact in a high speed boat, lateral impact in a train or subway, a pothole strikes by a vehicle with stiff tire or stiff suspension. Unexpected load conditions can occur in the fore-aft direction because of the “slack action” resulting from velocity changes in trains, subways etc [83].

While unexpected force or load is exerted on the body, CNS must respond rapidly in order to restore postural stability.. But there is a tendency for the “CNS” to overshoot in response to these events. This overshoot is characterized by an increase in the number of muscles activated, the onset rate of muscle activity and the magnitude of the muscle activity [86]. Unexpected perturbation in lumbar spine, which results in increased trunk muscle activity, creates greater compressive loads on the spine and is a potential mechanism for injury to contractile and non-contractile spine structure.

Sudden loading paradigms are designed to investigate the neuromuscular preparation and response to biomechanical trunk perturbation. Thomas et al. [86] investigated the effects of sudden loads applied to the lumbar spine. During unexpected sudden loading increased peak muscle activity, greater displacement of the trunk and synchronized peak response of the anterior posterior trunk muscles were observed, which in turn might cause an increase in the compressive load on the spine [86]. Granata et al. demonstrated that dynamic stability relies greatly on neuromuscular feedback and this neuromuscular response rate and magnitude may contribute to LBD risk [87]. Cholewicki et al. found that patients with low back pain, in contrast to healthy subjects, demonstrated a significantly different muscle response pattern in response to a sudden load release. They attributed these differences as either a predisposing factor for low back injuries or a compensation mechanism to stabilize the lumbar spine [88]. Wilder et al. found that patients with low back pain had longer reaction times compared to healthy subjects [89]. Magnusson et al. also found that chronic low back pain patients have less ability to protect themselves from sudden loads [90].

Although research on vibration is extensive, few have been conducted related to vibration-induced changes in proprioception of lumbar spine. Brumagne et al. found changes in reposition sense (proprioception) with exposure to 70 Hz back vibration at 0.5mm amplitude for only 5 seconds [52]. This application, however, was at a much higher than the principal resonance frequency of the human body at about 5 Hz in a seated posture. In a recent study Li [80] combined both reposition tests and sudden loading tests to assess the 5 Hz whole body vertical vibration induced changes. It was observed that the lumbar reposition sense accuracy decreased with exposure to vibration which in turns increased the sudden load response time which may contribute to the low back pain.

## 1.6 Effect of Whole Body Vertical Seat Pan Vibration

Occupations involving vibration (driving vehicles and operating power tools) have been linked to increased reports of low back pain [1-4]. Previous studies have suggested a strong relationship between exposure to whole body vertical vibration and low back pain [5]. Epidemiological studies as well as biomechanical research has given evidence for an elevated risk of health impairment of the lumbar spine and the connected nervous system due to long duration exposure to whole body vibration of high intensity [6]. Studies have reported that a resonance of the seated person to vertical seat pan vibrations of 4- 6 Hz [7-9]. A direct muscle and tendon vibration has been shown to effect in altered proprioception and kinesthetic illusions. Proprioception is the process of presenting the central nervous system with peripheral data related to joint position, motion, and force that is subsequently processed at conscious and subconscious levels in order to initiate appropriate motor responses. Proprioceptive information is a critical source of sensory information for optimal and efficient motor performance [10-12]. The muscle spindle organs, a primary element of joint proprioception, have been shown to be sensitive to vibration both during and after vibration.

One potential mechanism is altered proprioception leading to inappropriate stabilization of the lumbar spine. Measuring spinal stability is important in studying the etiology of low back pain [13]. Response to sudden loading can be used as a measure of dynamic stability of the spine. Many researchers [14] have used response to a sudden load to study the dynamic stabilization of the spine. Response to sudden load may change with exposure to vibration. Wilder et al. have shown that the response time to sudden loading increases with vibration exposure [13]. Modeling suggests that, changes in proprioception will lead to increased delays in sudden load muscle response latency and decreased effective trunk stiffness [15].

In a recent experimental study, Zhang [16] and Arashanapalli [17]demonstrated an association between proprioception and neuromotor response to a sudden perturbation, when vibration was applied to the paraspinal musculature. Increased delays in neuromotor response and increased flexion after a sudden perturbation were observed during paraspinal muscle vibration, specifies the role of proprioceptive system in dynamic stabilization. In addition, in a recent study of whole body vertical vibration induced changes in position sense, Li demonstrated that increase errors in reposition sense with vertical vibration exposure was associated with increase delays in neuromotor response and increased flexion in sudden loading experiments [18].

## 1.7 Effect of Whole Body Horizontal Vibration

Seated human occupants responses to whole body vibration have been widely investigated. The majority of these studies focus on response analysis of seated body exposed to vertical vibration. However, a number of studies have found that, in a number of occupations, vibration exposure includes strong components of vibration in horizontal (fore-aft), lateral, and rotational directions [39, 91, 92] . Vehicles such as trucks, earth moving machinery, industrial vehicles such as forklift trucks, and port cranes exhibit large amounts of fore-aft seat vibration [32]. For off-road tractors when ploughing, harrowing or drilling, the magnitudes of frequency-weighted horizontal vibration was found to be either comparable to or exceeding that of the vertical vibration [93, 94]. Mansfield and Lundstrom et al. [95] studied 56 construction vehicles for multi axis vibration magnitude and observed that 13 of the vehicles exhibited more weighted vibration in the horizontal or fore-aft direction and the remaining vehicles in the horizontal direction exhibited at least 90% of that reported for vertical motion. Since the magnitudes of vertical vibration encountered in road and off- road vehicles are generally believed to be higher than those along the other axis and cause more detrimental effects in view of operator health and safety [91]. While the nature of horizontal vibration transmitted along the horizontal axis is also known to be severe, particularly for many off- road vehicles, only a few studies have investigated along this axis [94].

Vibration in work environment is almost always multi-axis, most measurements of biomechanical response of the seated person have only considered single- axis vibration [96]. Mansfield described this is due to practical difficulties of the requirement for laboratories with a multi-axis capability, the complexity of experimental design with multi-degrees of freedom for independent variables and for linear system, single-axis data should be applicable in multi-axis environments. But some researchers demonstrated that a cross axis effect, vibration in one axis causes a response in another axis due to complex motion of human body, which in turn cause some discrepancy in data, measured using single-axis and multi-axis vibration [17, 92, 97].

The seriousness of vibration effects on the human body has led some investigators to extend their studies to the effect of axial, transverse and torsional vibrations. Pope et al. and Wilder et al. have reported that low back pain could be due to both vertical and bending motions [98]. Sjoflot and Suggs have shown that human is more affected by transverse angular vibration than by vertical vibration alone [99]. Qassem et al. developed a human mechanical model and analyzed the effect of horizontal and vertical vibration on human body. He suggested that the vibration force comes from hand, seat or a combination of the two and found that the body segments are affected by horizontal vibrations more when the input force comes from both hand and seat than when a the input force comes from the seat alone [100]. Therefore the response characteristic of the human spine to fore-and-aft vibration requires the assessment of vibration frequency and amplitude on the transmission functions between seat and low-back flexion-extension and neuromuscular system.

It was noted that a number of studies have demonstrated neuromotor effects such as increased position sense error, muscle fatigue, and increased delays in neuromotor response with exposure to vibration [1, 45, 75, 76, 80]. It has been suggested that such effects may alter spinal stability leading to an increased risk of low back injury. This, however, has not been investigated for exposure to vibrations in other directions expect vertical such as fore-aft, horizontal vibrations. It is therefore, unknown, whether such vibrations may also alter stability and potential increase risk of injury.

## 1.8 Objective and Hypothesis

Different studies reporting biodynamic responses of human occupants exposed to vertical vibration have provided considerable insights into the resonant behavior of the biological system. However the effect of horizontal (fore- and –aft) whole body vibration has not been extensively investigated. In the previous study of vertical vibration, reposition sense test and sudden loading test were combined to assess the effect at 5 Hz of vertical whole body vibration, as in these mechanisms , vibration causes higher incidence of low back injuries. Although there are few studies conducted on biodynamic interaction of human on response to whole body horizontal vibration, there are no studies that examine the proprioception (lumbar position sense) change and effect of sudden loading on lumbar spine after exposure to vibration on horizontal (fore-and-aft) direction. The objective of the current study is therefore to investigate how the whole body, horizontal seatpan vibration affects muscle response and to compare these results with the previously studied whole body vertical vibration. It was hypothesized that the change in lumbar position sense accuracy would decrease and the time to peak muscle response would increase with the exposure to horizontal vibration as it was observed in vertical seat pan vibration.

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# 2.0 Method

19 adult subjects including 10 male (age 25 ± SD 3.91 years, height 1.76 ± SD 0.1 meter and weight 83 ± SD 10.57 kg) and 9 female (age 21 ± SD 3.01 years, height 1.68 ± SD 0.1 meter, weight 61 ± SD 8.13 kg) participated in this study which was approved by the Human Subjects Committee University of Kansas, Lawrence KS. Before the study, all subjects signed a consent form approved by the Human Subjects Committee, University of Kansas and filled out medical questionnaires. Subjects with previous history of low back pain or musculoskeletal injury were excluded from the experiment.

## 2.1 Experimental Protocol

The experiment protocol consisted of two tests, A ***Reposition sense test*** (ability to measure position sense and detect lumbar posture) and a **S*udden loading test*** (used as a measure of dynamic stability of spine). Both the reposition sense tests and sudden loading tests were conducted in three conditions: before exposure (pre condition), after vibration (postv condition) and after washout (postw condition). The possible orders of these conditions (pre-postv-postw and pre-postw-postv) were randomized between the subjects. Postw followed a 30 minute period of quiet sitting and postv followed a 30 minute period of vibration exposure.

Voluntary, isometric, maximal, muscular contraction was performed to normalize electromyographic (EMG) signals. The subjects were instructed to lie down on their abdomen with face down on a bench. Their leg and hip was strapped to the bench to prevent motion of their lower body. They were instructed to raise their chest off the bench while their shoulders were kept in place by an investigator. In this position, maximum EMG activation of the Erector Spinae (ES) muscle groups were achieved. Similarly to determine the maximum EMG activation of the Internal Oblique (IO) and External Oblique (EO) muscle group, the subjects laid down on the same position. They were instructed to twist their torso, raising their right shoulder off the bench (clockwise) to achieve maximum EMG activation for the right EO and left IO. Similarly by twisting their torso and raising left shoulder off the bench (counter clockwise), maximum EMG activation for left EO and right IO was achieved. The subjects were then instructed to lie down on their back with face upward on the bench. Maximum EMG activation of Rectus Abdominus (RA) was achieved while the subjects perform a sit up with their shoulder kept in place by an investigator.

For the vibration exposure, the subjects were seated on a shaker table and exposed to a 5 Hz, 0.284 RMS ($m/s^{-2}$) constant acceleration, horizontal (fore-aft) vibration for 30 minutes. For the washout period, the subjects were asked to seat and relax in a chair for 30 minutes. The order of the vibration exposure and the washout exposure were randomized [Figure2].

<table>
<tr><th colspan="9">Order 1</th></tr>
<tr><th colspan="3">Pre</th><th></th><th colspan="2">PostW</th><th></th><th colspan="2">PostV</th></tr>
<tr><td rowspan="2">Maximum muscle activity</td><td>RST</td><td></td><td rowspan="2">30 min washout period</td><td>RST</td><td></td><td rowspan="2">30 min vibration exposure</td><td>RST</td><td></td></tr>
<tr><td></td><td>SLT</td><td></td><td>SLT</td><td></td><td>SLT</td></tr>
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<table>
<tr><th colspan="9">Order 2</th></tr>
<tr><th colspan="3">Pre</th><th></th><th colspan="2">PostW</th><th></th><th colspan="2">PostV</th></tr>
<tr><td rowspan="2">Maximum muscle activity</td><td>RST</td><td></td><td rowspan="2">30 min vibration exposure</td><td>RST</td><td></td><td rowspan="2">30 min washout period</td><td>RST</td><td></td></tr>
<tr><td></td><td>SLT</td><td></td><td>SLT</td><td></td><td>SLT</td></tr>
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**Figure 2. Time line for three different conditions – pre, postw and postv. In Pre condition Reposition sense test (RST) and sudden loading test (SLT) were performed. After the pre condition Postw or Postv conditions were performed followed by 30 minutes of washout period or 30 minutes of vibration exposure respectively**

## 2.2 Data Acquisition

Eight surface electromyographic (EMG) electrodes (Delsys, Boston, MA) were attached to the skin over 8 trunk muscle groups. Two electrodes were placed bilaterally over the right and left Erector Spinae muscles at L2/L3 level of the spine with 4 cm inter-electrode spacing. For the Rectus Abdominus muscles, electrodes were placed bilaterally 3 cm lateral and 2 cm superior to the umbilicus. Another two electrodes were placed 10 cm lateral to the umbilicus with an orientation of 45° to vertical over External Oblique muscles. The last two electrodes were placed 8 cm lateral to the midline within the lumbar triangle at a 45° orientation over Internal Oblique muscles [101] [Figure 3].

The EMG data were collected at 1500 Hz. The EMG signals are shown within the frequency range of 0- 500 Hz, where the usable signals are within 50- 150 Hz range [102]. Raw EMG data were band pass filtered between 20 and 500 Hz, with several notches filters (60, 120, 180 and 240 Hz) to remove electrical and electromagnetic noise. The

EMG data were rectified and integrated using a 100 point Hanning window. The average of the integrated EMG (iEMG) for these maximal exertions was collected and used to normalize all subsequent iEMG signals.

A 3-D electromagnetic motion analysis system, Motion Star (Ascension Tech., VT) was used to collect kinematic data. Five electromagnetic sensors were used to monitor torso flexion as well as to assess lumbar angle based on surface curvature. They were placed on the skin over the manubrium, cervical (C-7), thoracic (T-10, T-12) and sacrum (S1) position of the spine respectively using double-sided tape. The positions of the sensors were kept consistent with previous literature on lumbar position sense and lumbar pelvic coordination[103].These sensors gave both orientation and position. Lumbar angle was defined as the difference in angle of the T10 and S1 sensors [Figure 4a]. Torso flexion angle was defined as the angle between the vertical line and a line connecting the T10, S1 sensors [Figure 4b]

Subject's target lumbar angle was obtained from their maximum and minimum range of motion. A real time feedback display was provided on a computer screen and it was placed straight infront at a distance of approximately 0.46 m from the subject. The distance of the computer screen from the participant was maintained for easy viewing. The subject was instructed to give his or her maximum and minimum range of motion (three times each) while maintaining zero degrees of torso flexion. The midpoint between the average maximum and average minimum of the range of motion was selected as the target lumbar angle. The target flexion angle was kept at zero throughout the experiment as the subject did the experiment standing upright.

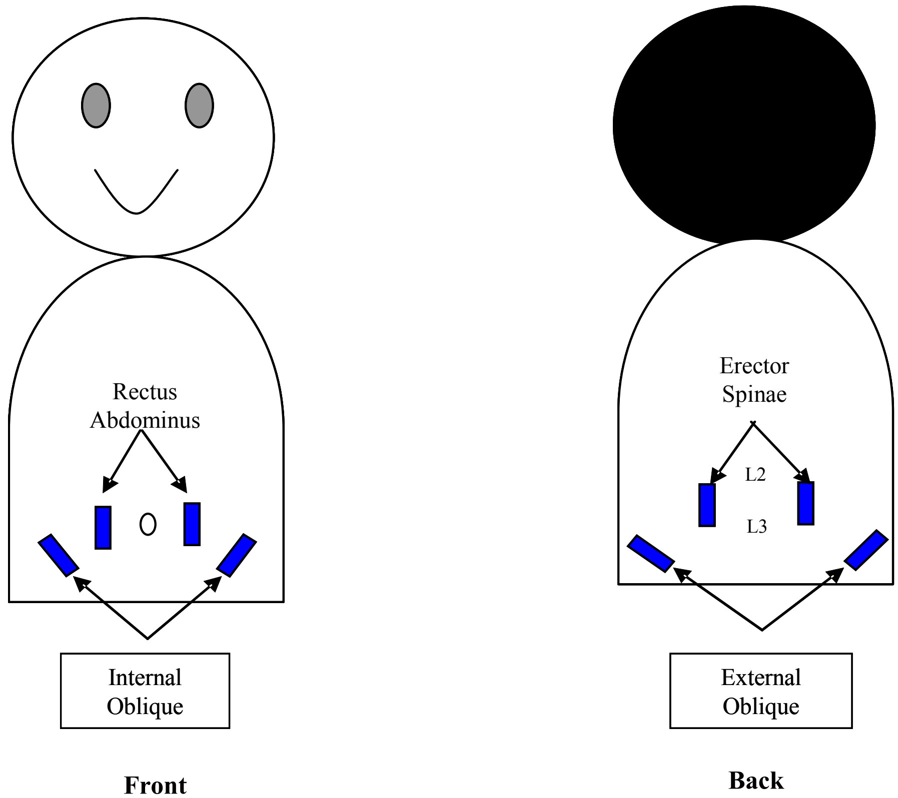


**Figure 3. Placement of eight electromyigraphic (EMG) sensors in Resctus abdominus, Internal obliques, Erector spinae and external obliques.**

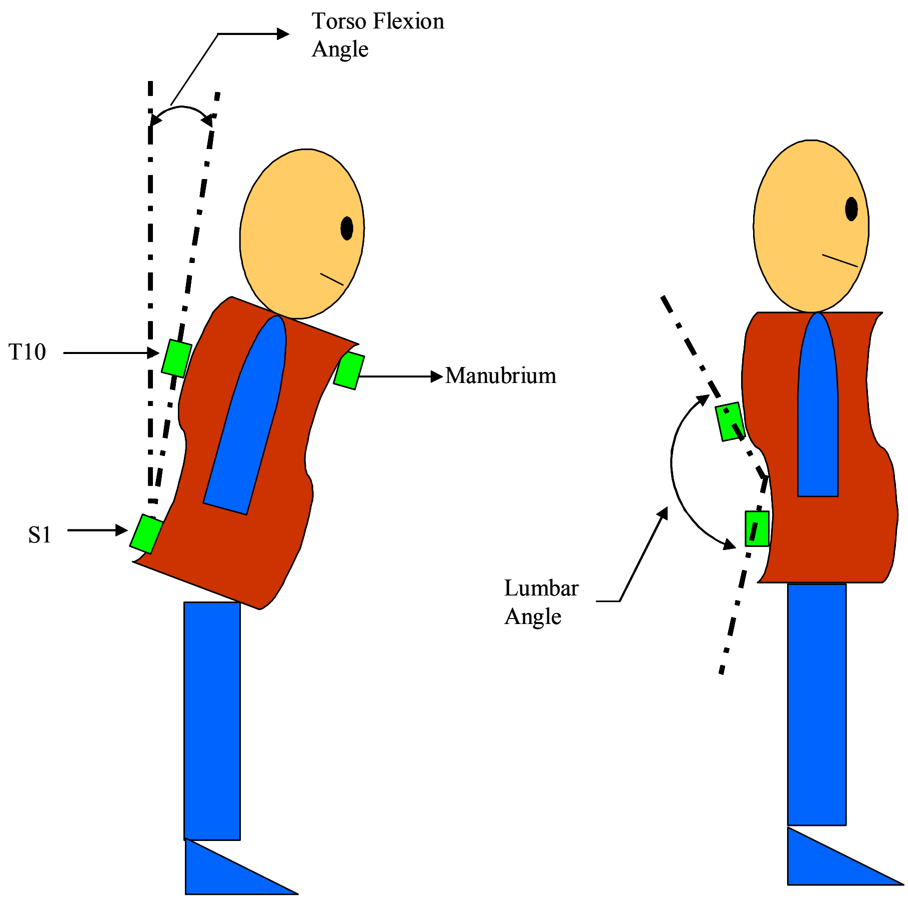


**Figure 4 (A). Torso Flexion Angle was defined as the angle between the vertical and a line connecting the T10 and S1 markers. (B) Lumbar Curvature Angle was defined as the difference in orientation of the T10 and S1 markers.**

## 2.3 Reposition Sense Protocol:

Reposition sense test was designed to determine the subject's ability to sense lumbar angle. During the experiment, each participant stood on a wooden platform and a real time feedback display was provided on a computer screen [Figure 5]. The reposition sense of the lumbar spine was assessed using a 3-D electromagnetic motion analysis system. This system provided both position and orientation of the electromagnetic sensors. The position and orientation data of the electromagnetic sensors were collected for 5 seconds at a frequency of 40 Hz. The reposition sense protocol includes five training trials followed by three assessment trials.

During the training trials, visual feedback of both lumbar angle and flexion angle was on and the subjects were asked to match their target lumbar angle, keeping their flexion angle at zero. The subjects were instructed to remember their target posture. After the subjects reached their target posture, they were instructed to hold that posture for five seconds while kinematic data was recorded. In the assessment trials, visual feedback for lumbar curvature was turned off and subjects were asked to reproduce the target posture from memory. Using the position of T10 and S1 sensors, torso flexion was determined as the angle between a line connecting these sensors and the vertical. The difference in angular orientation between the T10 and S1 sensors in the anterior posterior plane was defined as the lumbar angle [79]. The manubrium marker allowed detection of trunk rotation and asymmetry of motion. After every trial (training and assessment), the subjects were asked to flex approximately $30^0$ to make sure that they were not holding the same posture.

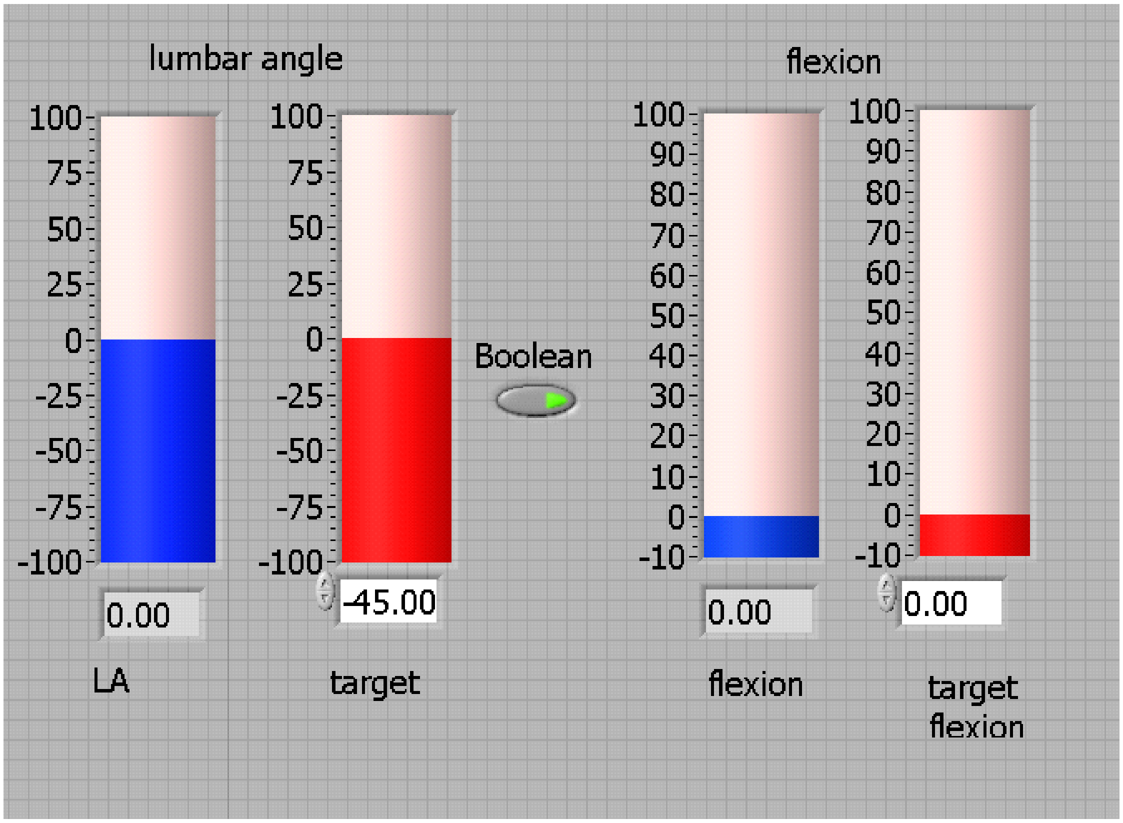


**Figure 5. Real time feed back display. The left pair of bars represents the lumbar angle and the right pair of bars represents flexion angles. The left bar of each pair represents actual lumbar and flexion angle measured from the electromagnetic marker. The right bar of each pair represents targeted lumbar and flexion angle. The button allowed the operator to turn off the lumbar curvature display during assessment trials.**

## 2.4 Sudden Loading Protocol

A sudden loading test protocol was used to determine dynamic spinal and neuromotor response. During this protocol, the subject stood upright on a wooden platform with extended knees and their hands were straight on the side. The subject's pelvis was fixed in place by using a belt attached to the wooden frame to avoid the movement of the trunk. The subject wore a chest harness which was attached to a load of 4.5 kg through a pulley and a Kevlar cable to the load cell. Visual and auditory cues were blocked by having the subject to stand behind a black curtain and wear a head phones playing white noise. To create a sudden impulse, the 4.5 kg load was dropped a height of 10 cm. A spring was attached to the load holder which allows the load to bounce against the spring to create an impulse load. In addition a contact switch was also attached to the load to record the instant of applied sudden load [Figure 6]. Before the load was dropped, training trials were performed and the subject was instructed to match his/ her target lumbar curvature and flexion angle with biofeedback to ensure same posture through out the experiment. The sudden loading protocol was repeated 5 times.

Impulse forces applied to the subject through sudden load were recorded from the contact switch. Electromyographic (EMG) data were collected from the EMG sensors placed in eight different muscle groups. The EMG and contact switch data were collected at 1500 Hz. The flexion movement caused by the sudden load was recorded from surface mounted electromagnetic markers (Ascension Tech., VT).

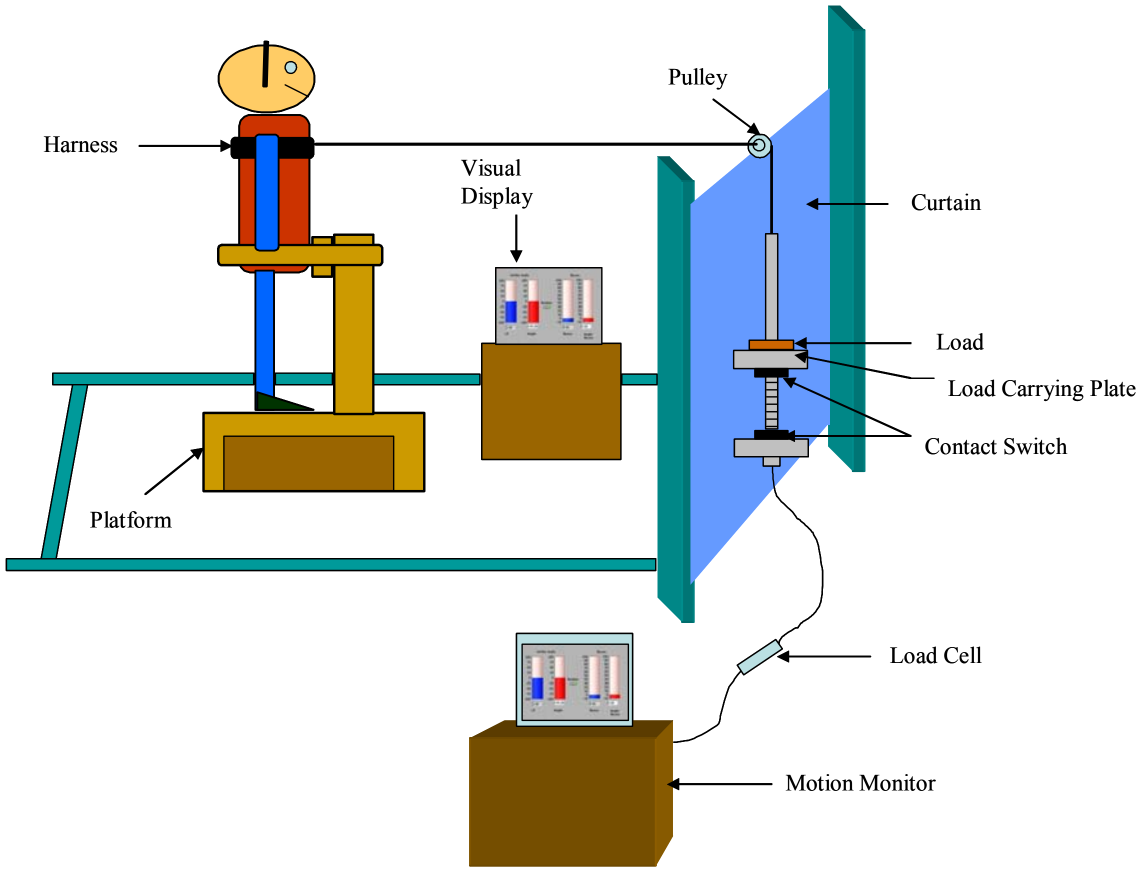


**Figure 6  Sudden loading test setup. Subject stood on a wooden platform with their pelvis fixed with a wooden bar and belt. The subject wore a harness which was attached to the load through a pulley. Both auditory and visual blocks were applied. Sudden loading was applied by dropping the 4.5 kg load from a height of 10 cm. Load cell and contact switch were attached to the load to measure the sudden onset of impact force.**

## 2.5 Whole Body Horizontal Vibration Protocol

A Ling 1512, electrodynamic, vibration shaker (Anaheim, CA) was used to introduce horizontal vibration to the seatpan. The shaker was powered by a DMA 2/X solid state amplifier (Anaheim, CA) and controlled by Daktron shaker control system (Fremont, CA). This controller allows the instructor to deliver sinusoidal vibration with a frequency range of 3 Hz to 14 Hz at magnitudes of 1 RMS and 2 RMS. The subject was instructed to seat on an unpadded seat without a backrest and with an adjustable, stable (non-vibrating) foot rest [Figure 7]. The subject was asked to sit in a comfortable, relaxed posture during the 30 minutes of exposure. This experiment was designed to analyze low back injuries and muscle response due to horizontal vibration, not the fatigue of muscle due to sitting posture. Therefore natural sitting posture was preferred to eliminate muscle fatigue.

A vibration of 5 Hz with a constant acceleration of 0.284 RMS ($m/s^{-2}$) was selected. This frequency is similar to that used in the previous whole body vertical vibration study (ref) In the previous study, the vibration frequency was chosen 5 Hz because transmissibility from the seat of the vibration has been shown to resonate at around 5 Hz and drop quickly at higher frequencies. The acceleration magnitude of the vibration was set at 0.284 $m/s^2$ RMS because, according to the ISO 2631-1:1997 standard, the subjects or operator should not feel discomfort when the overall R.M.S value of the frequency weighted acceleration is below 0.315 RMS ($m/s^{-2}$). After 30 minutes of exposure to vibration, the subjects were asked to perform reposition sense test and sudden loading test.

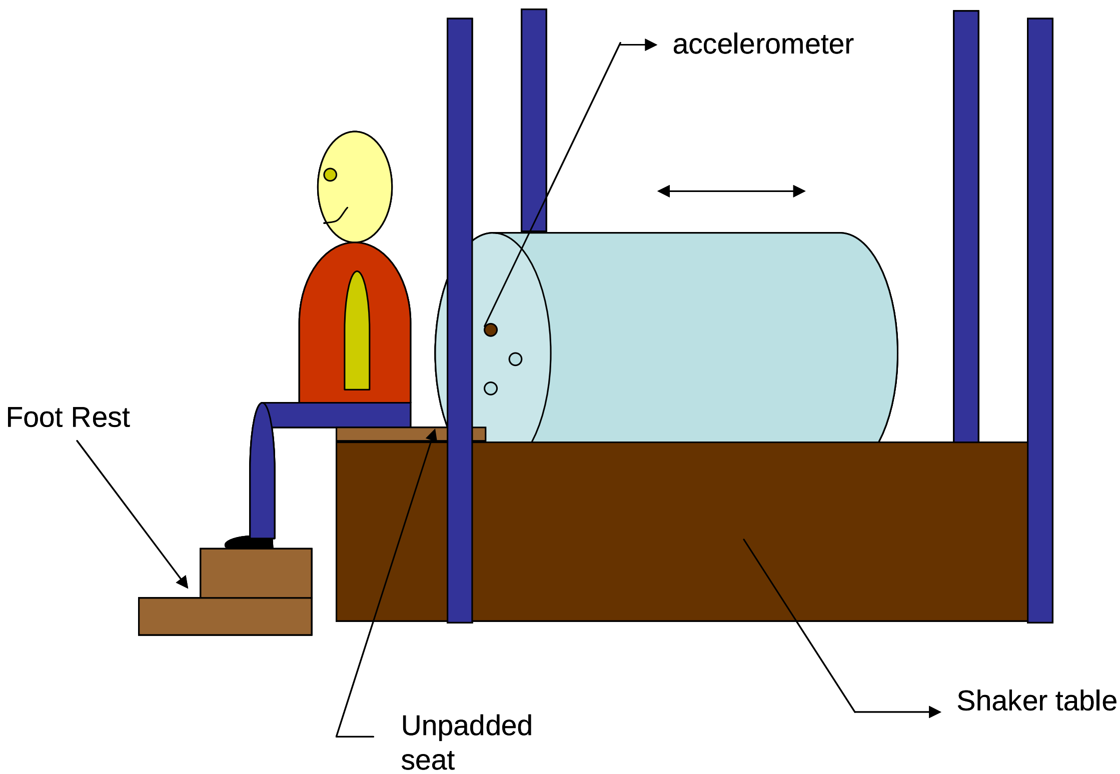


**Figure 7 Subject seated in a relax posture on an unpadded seat without backrest. The shaker table supplied the vibration in fore and aft direction. Subject's foot was rested over an adjustable footrest.**

## 2.6 Washout Protocol

In this study, 30 minutes washout period was observed during the experiment. At pre condition reposition sense test and sudden loading test were performed. After the pre condition the subject was exposed to 30 minutes of washout period or 30 minutes of whole body horizontal vibration exposure. The order of vibration exposure and washout exposure were randomized. During the washout period, the subject was asked to sit in a chair and relax for 30 minutes. This washout period was selected to give a control condition that could be compared to the vibration exposure condition. Any learning effects that might occur with multiple testing would be observed in the pre and post washout data. After washout period reposition sense test and sudden loading test were performed simultaneously.

## 2.7 Data Analysis

### 2.7.1 Reposition Sense Test

In reposition sense test, after the subject reached his/ her target flexion and lumbar angle in training and assessment trials, kinematic data were recorded at 40 Hz for 5 seconds for nineteen subjects. The training error for position sense was defined as the difference between target lumbar angle and lumbar angle attained lumbar angle (degree) with the visual feedback on. Absolute reposition sense error was defined as an absolute difference between the target lumbar angle ($\theta_t$) and the assumed lumbar angle ($\theta_{a,.}$) in degrees. These errors were averaged over 5 seconds.

$$\text{Absolute Reposition Sense Error (RSE)} = |(\theta_a) - (\theta_t)|$$

### 2.7.2 Sudden Loading Test

During the sudden loading protocol, the electromyographic data, the load cell data and the contact switch data were simultaneously collected at 1500 Hz. Trunk flexion was collected from the electromagnetic markers. Although EMG data was collected for all muscle groups, right and left side of Erector Spinae muscle and average of right and left side of Erector Spinae muscle data were analyzed to determine the response time between the onset of impact force and the onset of reflex response. Electromyographic data for the Erector Spinae muscle was analyzed for delay in the time to peak muscle activity. The onset of impact force was determined from the contact switch data and the onset of reflex response was determined from the signals of preparatory muscle activity. Preparatory muscle activity was derived from the EMG data collected 500 milliseconds prior to each impulse force.

The Erector Spinae response time to sudden perturbation was defined as the temporal difference between the start of the applied sudden load ($\tau_l$) and first peak muscle reaction ($\tau_m$). In addition to Erector Spinae Muscle Response Time, magnitude of peak muscle activity, and average Erector Spinae activity before the perturbation were also analyzed. Flexion motion data was recorded at 40 Hz and was analyzed for torso flexion magnitude, lumbar curvature magnitude.

$$\text{Erector Spinae Muscle Response Time (MRT)} = \tau_m - \tau_l$$

### 2.7.3 Statistical Analysis

A student T test was performed to assess whether the data of two groups of different conditions (pre, postw and postw) are statistically different from each other. With a sample size of nineteen, the student paired two tailed T test was performed to determine statistical significance among the data of pre exposure and post exposure condition.

### 2.7.4 Comparison of Results between Vertical and Horizontal Vibration

To compare the results between vertical and horizontal vibration, the data were normalized by dividing after exposure condition to their pre exposure condition for each subject. The ratio of absolute reposition sense error, time to peak response, torso flexion magnitude and lumbar curvature magnitude were compared between vertical and horizontal vibration.

# References

[1] Mirka, G.A., *The quantification of EMG normalization error.* Ergonomics, 1991. **34**: p. 343-52.

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# 3.0 Results

In this study of whole body horizontal seat pan vibration, the objective was to investigate how position sense and sudden loading dynamics altered with horizontal vibration exposure of 5 Hz and compare these results with the previously studied whole body vertical vibration exposure. 19 subjects were participated in this study including 10 male (age 25 ± SD 3.91 years, height 1.76 ± SD 0.1 meter and weight 83 ± SD 10.57 kg) and 9 female (age 21 ± SD 3.01 years, height 1.68 ± SD 0.1 meter, weight 61 ± SD 8.13 kg). The primary variables analyzed in this research are:

- Position sense tTestest: Absolute reposition sense error during assessment trials.
- Sudden loading test: Time to peak muscle response, torso flexion magnitude and lumbar angle.

## 3.1 Reposition Sense Test

Reposition sense test was performed to measure the ability to sense joint position and detect lumbar posture. This test consisted of 5 training trials followed by the alteration of 3 assessment trials. Training error was defined as the difference in angle between the target lumbar angle (degree) and the attained lumbar angle (degree) when the visual feed back was on. The assessment error, when the visual feedback was turned off, is the difference between target lumbar angle and the assumed lumbar angle during assessment trials. Absolute reposition error was defined as the absolute difference between the target lumbar angle and the assumed lumbar angle.

Reposition accuracy was significantly reduced when visual feedback was removed. The absolute value of error in lumbar curvature during training trial was small [Figure 1]. This illustrates the subjects were able to effectively use the visual feed back to control lumbar angle. The absolute value of error was increased significantly ($p<0.0001$) in the reposition assessment trials when visual feedback was no longer available. The absolute value of the error indicates the magnitude of the error.

Absolute reposition sense error during assessment trials decreased 15% in post washout (postw) condition compared to the pre condition [Figure 8]. However, with exposure to whole body vibration (postv) for 30 minutes, reposition sense error was higher (5%) than the post washout condition [Table 1]. A two-tail, paired student T test was performed between pre, post washout and postv condition. No significant statistical difference ($p= 0.42$ for pre and postw, $p= 0.7$ for postw and postv and $p= 0.6$ for pre and postv ) was found between these conditions.

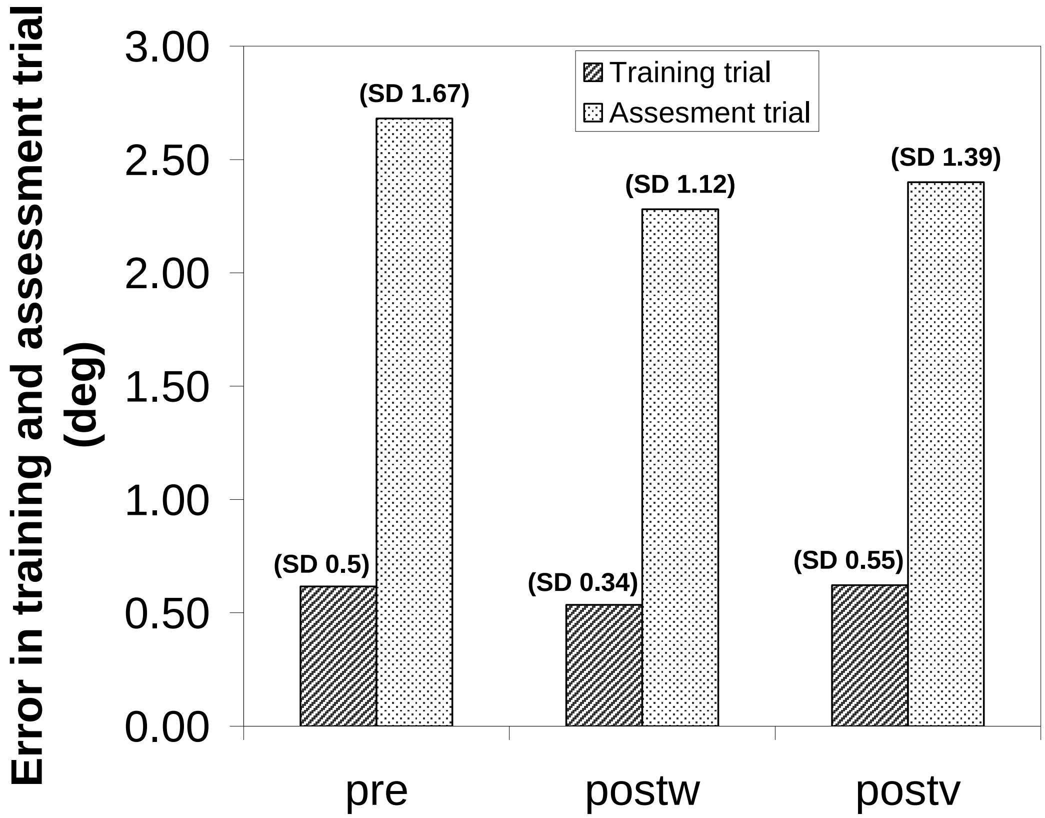



**Figure 8. The absolute value of the error in lumbar curvature was small during training trials. Whereas this error increased significantly in assessment trials.**

**Table 1. MEAN $\pm$ SD for Average reposition sense error (degree) for pre, postw and postv condition.**

| | Pre | StDev | Postw | StDev | Postv | StDev |
|---|---|---|---|---|---|---|
| **Average reposition sense error (deg)** | 0.62 | 0.50 | 0.54 | 0.34 | 0.62 | 0.55 |

**Table 2. MEAN $\pm$ SD for Average absolute reposition sense error (degree) for pre, postw and postv condition.**

| | Pre | StDev | Postw | StDev | Postv | StDev |
|---|---|---|---|---|---|---|
| **Average absolute reposition sense error (deg)** | 2.68 | 1.67 | 2.28 | 1.12 | 2.40 | 1.39 |

## 3.2 Sudden Loading Test

Sudden loading response dynamics were determined using a sudden drop protocol. Subjects stood upright on a wooden platform and visual and auditory cues were blocked. A sudden load was applied by dropping a 10 lbs weight attached to the cable at a height of 10 cm. A contact switch was used to indicate the instant the sudden load applied. Right Erector Spinae muscle data were analyzed to determine the response time between the onset of impact force and the onset of reflex response, collected from electomyographic (EMG) sensors at 1500 Hz. Torso flexion magnitude and Lumbar angle were also observed from the electromagnetic sensors.

### 3.2.1 Time to Peak Muscle Response

Time to sudden perturbation was defined as the temporal difference between the start of the applied sudden load and first peak muscle reaction. This response was decreased 5% in post washout condition compared to the pre condition. With vibration, time to peak response was 7% larger than the post washout condition [Figure 9]. A paired two tailed student T test was performed and did not show significant difference (p= 0.3) between postv and postw , pre and postw (p= 0.4) and pre and postv condition (p= 0.7).

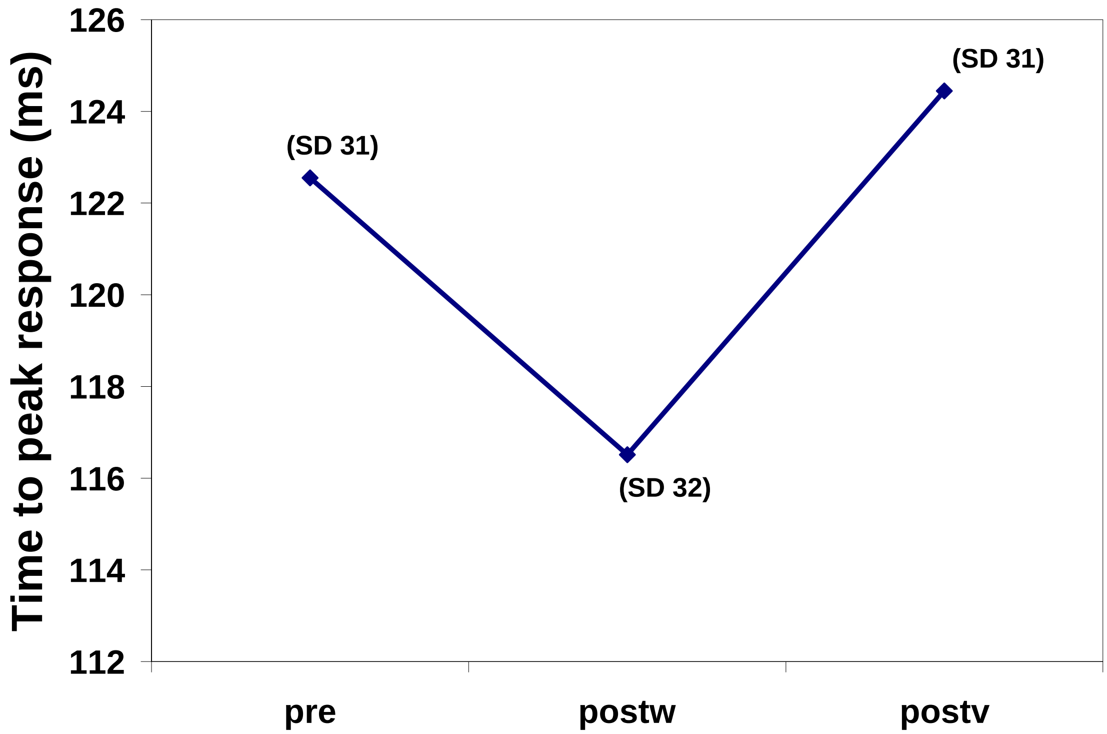



**Figure 9. The sudden loading response time was found to decrease 5% during pre to postw condition. This response was increased by 7% during postv condition compared to postw condition.**

**Table 3. MEAN ± SD for Time to peak muscle response time (ms) for pre, postw and postv condition.**

| | Pre | StDev | Postw | StDev | Postv | StDev |
|---|---|---|---|---|---|---|
| **Time to peak muscle response (ms)** | 122.55 | 31.00 | 116.51 | 32.00 | 124.44 | 31.00 |

### 3.2.2 Torso Flexion Magnitude

Torso flexion magnitude was observed and it increased 12% in post washout condition compared to pre condition.  Post vibration flexion was found to be slighter larger than post washout flexion (3%) [Figure 10]. A paired two tailed student T-test were performed between each condition. A trend was observed ($p = 0.07$) between pre and postw and between pre and postv conditions ($p = 0.09$). No statistical significance was observed between postw and postv condition ($p = 0.61$).

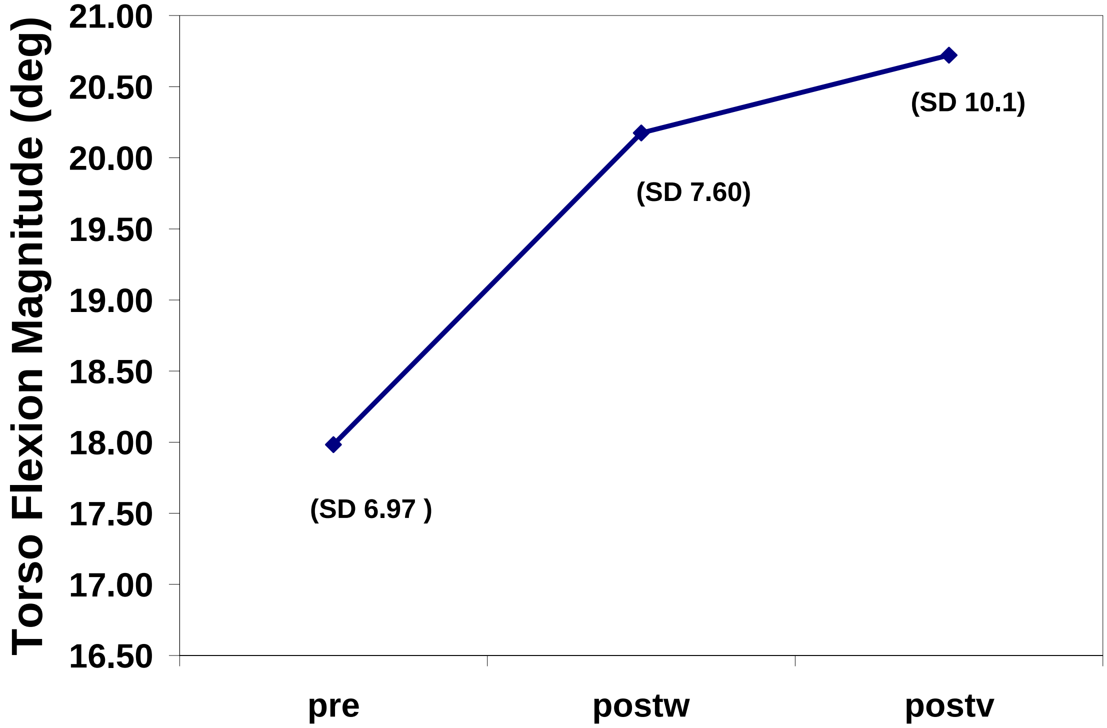



**Figure 10 The torso flexion magnitude increased by 12 % during pre to postw condition and further increased slightly (3%) in postv condition.**

**Table 4 MEAN $\pm$ SD for Torso flexion magnitude (degree) for pre, postw and postv condition.**

| | Pre | StDev | Postw | StDev | Postv | StDev |
|---|---|---|---|---|---|---|
| **Torso flexion magnitude (deg)** | 17.98 | 6.97 | 20.18 | 7.60 | 20.72 | 10.10 |

### 3.2.3 Lumbar Angle

The lumbar magnitude was only increased 6% in postw condition compared to the pre, while it was 7% smaller after the exposure to whole body vibration compared to postw condition. A paired, two-tailed, student T-test was performed. Within subjects contrasts the result between pre and postv condition (p=0.87) was not significant. A trend was observed between postw and postv condition (p=0.08) as shown in Figure 11.

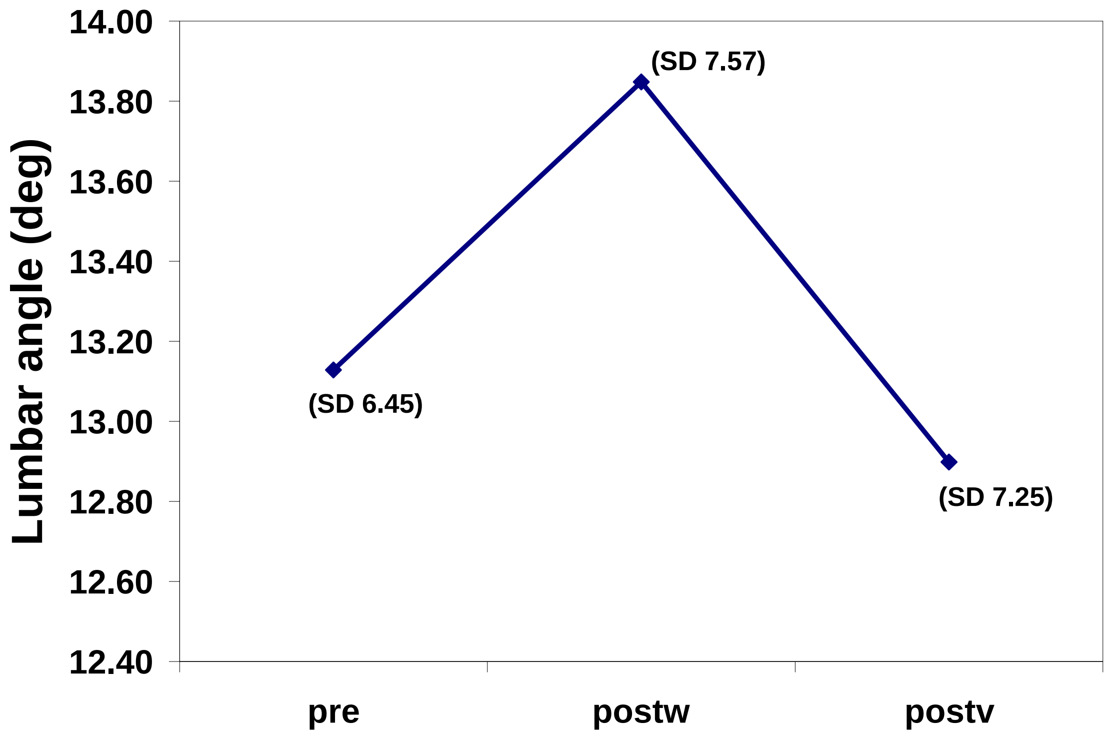



**Figure 11 Lumbar angle increased 6% in postw compared to the pre and decreased 7% postv condition compared to postw condition.**

**Table 5 MEAN ± SD for Lumbar angle (degree) pre, postw and postv condition.**

| | Pre | StDev | Postw | StDev | Postv | StDev |
|---|---|---|---|---|---|---|
| **Lumbar magnitude (deg)** | 13.13 | 6.45 | 13.85 | 7.57 | 12.90 | 7.25 |

### 3.2.4 Preperatory Muscle Stiffness and Muscle Response Magnitude

The preparatory muscle activity was calculated by averaging EMG data 500 ms before sudden loading. Preparatory stiffness didn't vary much throughout the different conditions [Figure 12]. A paired two tailed student T test was performed between within subject contrast, but no significant changes were observed within different conditions.

The muscle response magnitude, defined as the difference between the first EMG peak activity and the onset EMG activity, decreased 21% in po stw condition compared to the pre condition. After exposure to whole body vibration this magnitude was almost same compared to the postw condition. A two tailed paired student T test was performed and found a trend within subject contrast ($p< 0.01$) between pre and postv condition but there was no significant change between pre and postw condition ($p= 0.17$) and postv and postw condition ($p= 0.97$) [Figure 6].

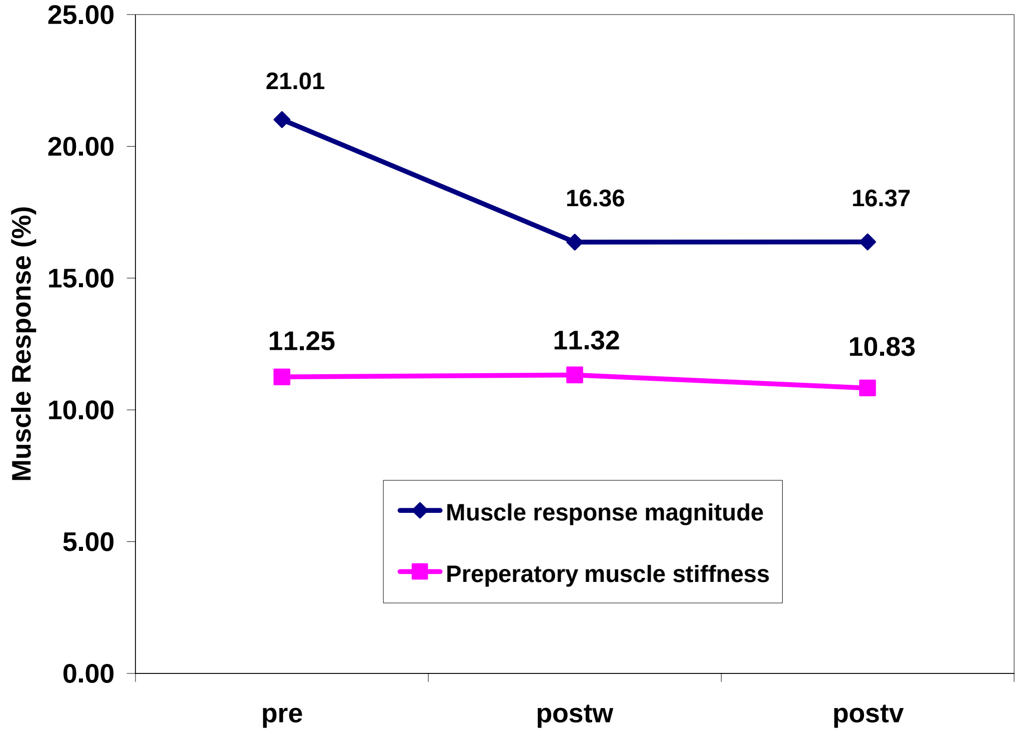



**Figure 12. Response Magnitude was found to decrease 21% during postw condition compared to pre condition. This value was almost same between postw and postv condition. Preparatory stiffness was found to vary very little during different conditions.**

# 4.0 Discussion

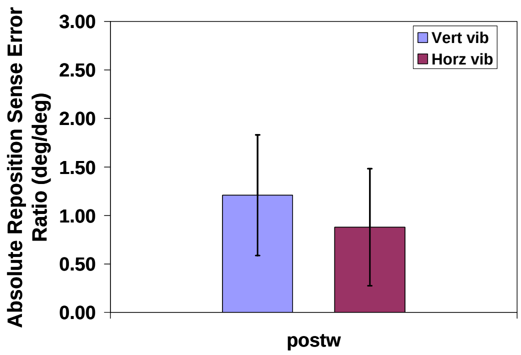
Whole body vibration has long to be known to be a risk factor of low back injury. A previous study found that whole body vertical vibration can decrease the ability of the propriceptive system to perceive position (increase reposition error) and this loss in turn resulted in slower response times in the reflex response behavior to sudden external impact (increased sudden loading muscle response time). The delayed response time can in turn threaten the stability and decrease dynamic control of the lumbar spine and result in unexpected low back injuries [104]. In the previous study, subjects were exposed to 20 minutes of 5 Hz vertical seat pan vibration. While vertical vibration exposure is a common occupational exposure, in some cases, there is some element of horizontal movement and in some circumstances the horizontal movement can dominant. Vibration in working environments is almost always multi-axis, but most measurements of the biomechanical response of the seated person have only considered single axis vibration. [92]. For off-road tractors when ploughing, harrowing or drilling, the magnitudes of frequency-weighted horizontal vibration have been found to be either comparable to or exceeding that of the vertical vibration. Vehicles such as articulated trucks, earth moving machinery, and some industrial vehicles (including excavators, off-road, forklift trucks, and port cranes) exhibit large amounts of fore-and-aft seat vibration [105]. Tractors and tanks have been shown to have more weighted acceleration in the horizontal directions than in the vertical direction such as off road vehicles and construction vehicles horizontal (fore-and-aft) vibration may dominate [96].

In this study, the objective was to investigate how the whole body, horizontal, seatpan vibration affects muscle response and to compare these results with the previously studied whole body vertical vibration. While position sense and dynamic response were found in the previous study to be altered with vertical 5 Hz vibration, this was not found to be the case with horizontal 5 Hz vibration.

Some differences existed between the current study of horizontal vibration and the previous study of vertical vibration. In horizontal whole body vibration research, the order of vibration exposure (postv) and washout (postw) condition was randomized [Figure 13]. However, in the previous study (vertical vibration), the protocol followed a specific order of before vibration, immediately after vibration, 15 minutes after vibration and 30 minutes after vibration for an exposure group with a separate control group that was exposed to quiet sitting only. The current study was considered an improved design with greater statistical power due to the ability to use paired statistical testing [106]. The sample size between these two conditions was only slightly different (vertical vibration 17 and horizontal vibration 19). To compare the results between vertical and horizontal vibration, the data were normalized by dividing after exposure condition to their pre exposure condition for each subject. The ratio of absolute reposition sense error, time to peak response, torso flexion magnitude and lumbar curvature magnitude were compared between vertical and horizontal vibration.

In the previous study of 5 Hz, vertical, seatpan vibration, it was observed that the reposition sense error was increased significantly (31%) after vibration relative to a control condition as shown in Table 6 [Figure 13] [106]. It was suggested in the previous study that whole body vibration induced proprioceptive changes (reposition sense) lead to temporarily instability of the lumbar spine. The increased proprioceptive error in previous study reflects loss in the ability to sense lumbar posture and suggests neuromotor habituation or adaptation with exposure to whole body vibration [106]. In the current study of horizontal seat pan vibration, however, this error ratio increased slightly (2%) [Table 7] with vibration exposure relative to post washout and the increase was not significant.

It was observed that during assessment trials, the absolute value of the resposition sense error in the post washout period has a lower magnitude than the pre exposure condition ( pre 2.68 deg ± SD 1.67 and postw 2.28 deg ± SD 1.12). Although the results were not statistically significant ($p = 0.4$), one reason behind this pattern is the possible effect of learning. Subjects might be able to experience themselves of different testing conditions in pre trials, which results in lower values of post washout period.

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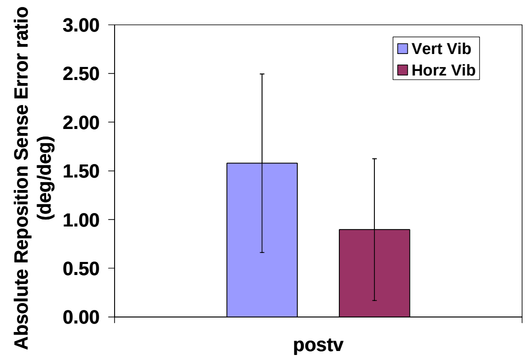



B

**Figure 13 A. Absolute reposition sense error ratio for vertical (1.21 ± SD 0.62) and horizontal (0.88 ± SD 0.6) vibration in postw condition. The error in postw period during horizontal vibration was less as a result of learning effect compared to vertical vibration.**

**B. Absolute reposition sense error ratio for vertical (1.58 ± SD 0.92) and horizontal ( 0.9± SD 0.73) vibration in postv Absolute reposition sense error ratio for increased 31% from postw to postv condition in vertical vibration, while in case of horizontal vibration, this error increased slightly 2%.**

**Table 6 Average Absolute Reposition Sense Error for Vertical Vibration**

| | Avg absolute error in assessment trial | Avg Normalized data ratio Rpostx= $postx_i$/ $pre_i$ | Standard Deviation rpostx | % increase ratio (postw-postv)/ postw |
|---|---|---|---|---|
| **pre** | 1.57 | | | |
| **postw** | 1.72 | 1.21 | 0.62 | |
| **postv** | 2.23 | 1.58 | 0.92 | 30.63 |

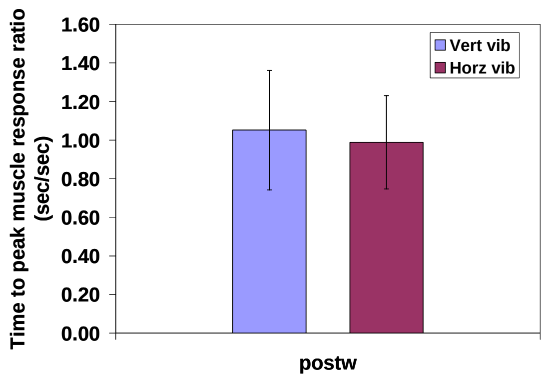
**Table 7 Average Absolute Reposition Sense Error for Horizontal Vibration**

| | Avg absolute error in assessment trial | Avg Normalized data ratio rpostx= $postx_i$/ $pre_i$ | Standard Deviation rpostx | % increase ratio (postw-postv)/ postw |
|---|---|---|---|---|
| **pre** | 2.68 | | | |
| **postw** | 2.28 | 0.88 | 0.60 | |
| **postv** | 2.40 | 0.90 | 0.73 | 2.13 |

In the whole body vertical vibration study, it was observed that the increase in reposition sense error was associated with a significantly increased delay in muscle response time after exposure to vibration. When the proprioception is altered due to the whole body vibration as shown in this case, it is plausible that the interrupted proprioception would affect the neuromotor control and hence result in different reflex pattern responding to external perturbation. The neural controller might need longer time to detect postural changes and process whole body vibration interrupted propriceptive information. As a result the muscle response to the external perturbation would be delayed. In the previous vertical seat pan vibration study, time to peak muscle response ratio was increased 11% after 30 minutes exposure to vibration. Although the results were not statistically significant in horizontal vibration (p= 0.7), the response time increased 5% in the current research which supports increased delay time for muscle response [Figure 14 A, B].

The delayed muscle response suggests that the lumbar spine may be less stable. When the intrinsically unstable spinal column does not get effective muscular support, the overall stabilization of the lumbar spine may be reduced and injury risk may increase. Under this condition, based on the previous studies [106, 107], even a small external perturbation could amplify even a small neuromuscular error and result in sudden undesired vertebral motion such as the spinal column buckling.

It was observed that time to peak muscle response in post washout period has a lower magnitude (116.51ms ± SD 32) than the pre exposure (122.55 ms ± SD 31) condition in horizontal vibration. One reason behind this pattern is the possible effect of learning. Subjects were able to experience themselves of different testing conditions in pre trials, which results in lower values of post washout period, although the results were not significant (p= 0.4).

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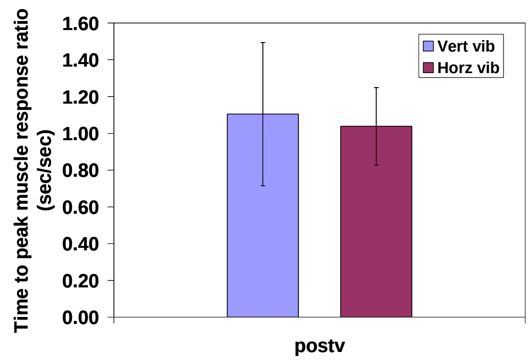



B

**Figure 14 A. Time to peak muscle response ratio for vertical (1.07± SD 0.31) and horizontal (SD 0.99 ± 0.24) vibration in postw condition. For horizontal vibration, muscle response required less time due to the result of the learning effect during pre condition.**

**B. Time to response ratio for vertical ( 1.19± SD 0.39) and horizontal (1.04± SD 0.21) vibration in postv condition. Response time ratio increased 5% in both vertical and horizontal vibration compared to postw condition.**

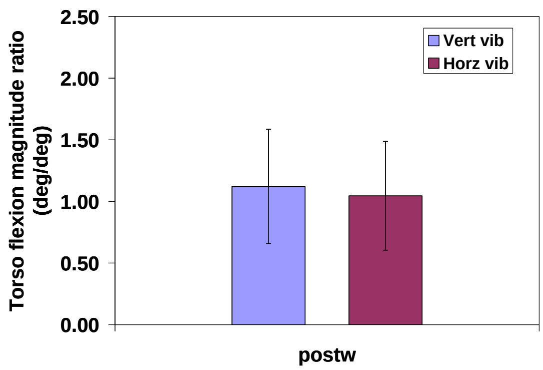
**Table 8 Time to peak muscle response ratio for vertical vibration.**

| | Avg peak time | Normalized data rpostx = $postx_i/ pre_i$ | Standard Deviation rpostx | % increase ratio (postw-postv)/ postw |
|---|---|---|---|---|
| **pre** | 207.11 | | | |
| **postw** | 217.81 | 1.07 | 0.31 | |
| **postv** | 228.78 | 1.19 | 0.39 | 11.28 |

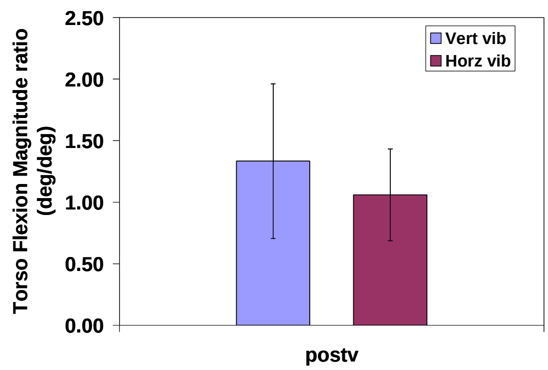
**Table 9 Time to peak muscle response ratio for horizontal vibration.**

| | Avg peak time | Normalized data rpostx = $postx_i/ pre_i$ | Standard Deviation rpostx | % increase ratio (postw-postv)/ postw |
|---|---|---|---|---|
| **pre** | 122.55 | | | |
| **postw** | 116.51 | 0.99 | 0.24 | |
| **postv** | 124.44 | 1.04 | 0.21 | 4.99 |

In addition to the reposition error and delayed muscle response to external perturbation, the torso flexion and lumbar curvature deflections were also analyzed. The flexion magnitude ratio increased 19% in vertical seat pan vibration and increased 3% in the current research of horizontal vibration compared to the control (pre) condition after the cessation of whole body vibration.[Figure 15 A, B]. A slight trend ($p = 0.07$) was observed in torso flexion magnitude during horizontal vibration. The lumbar curvature magnitude ratio was also observed to increase both in vertical (25%) and horizontal (6%) condition immediately after the termination of whole body vibration. This increased torso flexion deflection and lumbar curvature deflection in horizontal vibration further reflected the decreased stability of the lumbar spine due to the whole body vibration exposure.




A




B

**Figure 15 A Torso flexion magnitude ratio for vertical (1.12± SD 0.46) and horizontal (1.05± SD 0.44) vibration. In horizontal condition, flexion magnitude ratio was increased 6% in postw condition compared to that of vertical vibration.**

**B Torso flexion magnitude ratio for vertical (1.33± SD 0.63) and horizontal (1.06± SD 0.37) vibration in postv condition. The flexion magnitude ratio increased significantly (19%) in vertical vibration but increased slightly (3%) in horizontal vibration from postw period.**

**Table 10 Average Torso Flexion Magnitude for Vertical Vibration**

| | Avg flexion magnitude | Normalized data ratio $postx_i / pre_i$ | Standard Deviation | % increase ratio (postw-postv)/ postw |
|---|---|---|---|---|
| pre | 10.8 | | | |
| postw | 11.15 | 1.12 | 0.46 | |
| postv | 12.63 | 1.33 | 0.63 | 18.83 |

**Table 11 Average Torso Flexion Magnitude for Horizontal Vibration**

| | Avg flexion magnitude | Normalized data ratio $postx_i / pre_i$ | Standard Deviation | % increase ratio (postw-postv)/ postw |
|---|---|---|---|---|
| pre | 17.98 | | | |
| postw | 20.18 | 1.12 | 0.44 | |
| postv | 20.72 | 1.15 | 0.37 | 2.71 |

Comparing results between 5 Hz vertical and horizontal seat pan vibration it was observed that the results don't showed similar effects. Vibration attenuation of the different vibration inputs may be a possible reason for this difference. Griffin et al. measured the amount of fore-aft vibration transmission to the head and observed the measured transmissibility curves exhibited a gradual decline with increasing frequency [108]. The test was conducted between 0.2 and 16 Hz of fore-aft vibration and a transmissibility peak at 1.5 Hz (transmissibility ratio 1.3) was observed. At 5 Hz of horizontal vibration, Griffin et al. observed a transmissibility ratio less than 0.2. Fairley et al. assessed the response characteristics of seated occupants using apparent mass frequency functions in the fore- aft direction. The study was conducted from 0.25 to 20 Hz with random vibration and at magnitudes of 0.5-2.0 RMS ($ms^{-2}$). The apparent mass measure is a ratio of the horizontal force transmitted at the occupant and seat interface to the acceleration measured between the occupant and the seat interface. The results exhibit two resonance modes in the fore-aft direction, with a primary resonance peak at 0.7 Hz and secondary resonance peak at 2.5 Hz [109].

An ongoing study of vibration transmissibility found that transmissibility of horizontal vibration from the seat to the neuromotor system peaks (Transmissibility ratio 0.49) at a much lower frequency with horizontal (<3Hz) vibration. [110]. A gradual decline of transmissibility with increasing frequency with a little bump (transmissibility ratio 0.35) on 5 Hz was observed in the study [Figure 16]. While in another literature study, a similar research with vertical seat pan vibration was analyzed and primary resonance peak (transmissibility ratio 1.55) was observed at 4 Hz [111]. For the current study, the 5 Hz horizontal vibration may be attenuated within the soft tissues of the pelvis, reducing the potential effects of the vibration on lumbar proprioception and dynamics response. A lower horizontal vibration, however, may have a stronger effect.

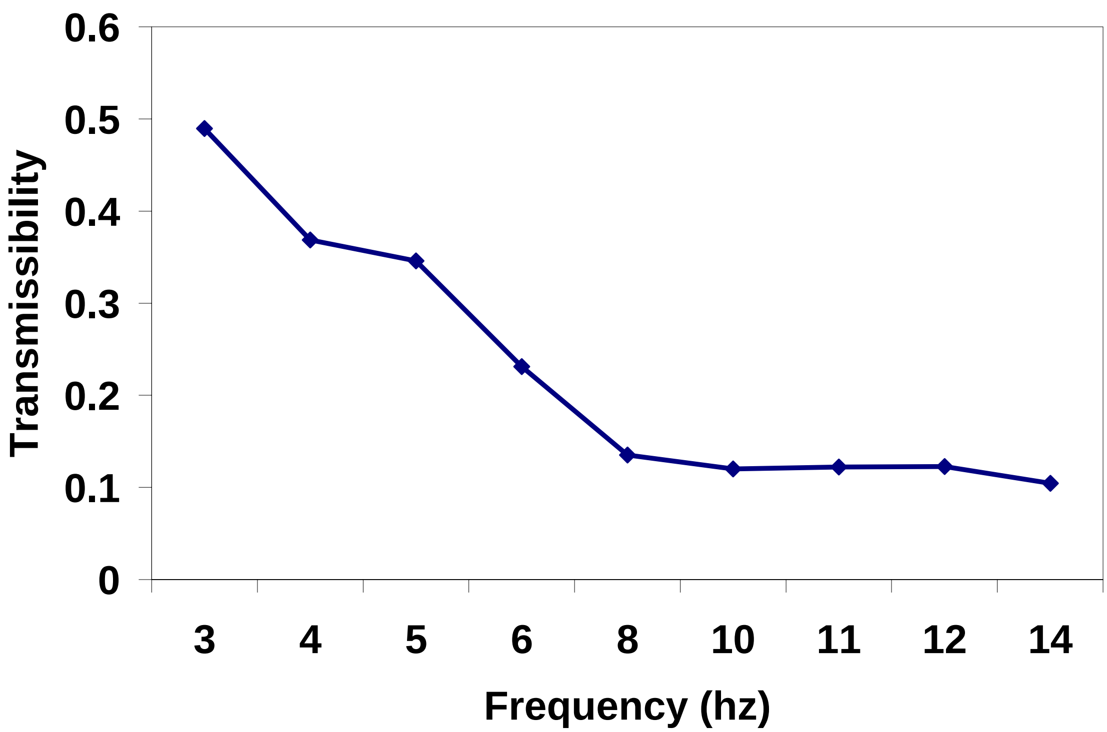



**Figure 16 Transmissibility of acceleration (2 RMS (ms-2)) at different frequency to the spine. Transmissibility decreased gradually with increasing frequency. Peak of transmissibility was observed below 3 Hz of frequency.**

## 4.1 Limitations

A number of limitations exist in this study. This experiment was conducted with an exposure to vibration for about 30 minutes. However it is possible that longer vibration exposure such as the 8 hour exposure of a normal working day may have different effects.

The effect of sitting posture needs to be considered. In this study, no posture control was applied during the vibration exposure. Like the previous vertical vibration study, the subjects were asked to sit in their own comfortable and relaxed posture, with their feet placed on an adjustable footrest. This kept the subjects relatively relaxed preventing muscle fatigue. Natural sitting posture was preferred as this research was designed to focus the effect of whole body horizontal vibration, not the effect of posture. In future studies, controlled sitting posture should be considered to better understand the effects of muscle fatigue.

The sudden loading device used in this study was designed for manual handling. Human variability in the dropping of the weight may contribute to variability in the sudden loading response data. This device needs to improved and automated to reduce human error. Servomotor can be used to apply perturbation; encoder and force transducer can be used to measure postural displacement and applied force to reduce error.

## 4.2 Future Study

In the future studies of horizontal seat pan vibration, an exposure with lower frequencies should be used because in an ongoing study of vibration transmissibility, it was found that transmissibility reached its peak at a frequency lower than 3 Hz [ref]. In thate current research of horizontal seat pan vibration transmission, no significantlittle muscle response was observed at 5 HZHz. The shaker table used for this research has the specification for the vibrating frequencies greater than or equal to 3 Hz, restricting to our studiesy to the frequencies below above 3 Hz. Therefore for future study, a different shaker should be used.

In addition, other transmission mode like transmission through backrest should be considered. In an ongoing study of horizontal vibration, the a transmissibility ratio of 1.4 was observed with backrest at 3- 6 Hz frequenciesy. Therefore, in horizontal vibration with backrest, a different muscle response might be observed.

Also only right side of the erector spinae muscle EMG data was analyzed in this research. Other muscle groups like rectus aAbdominus, internal/ external obliques could give a more complete understanding. Longer vibration exposure and combination of frequencies should be studied to further justify relationship between whole body vibration and low back injuries.

In sudden load experiment, the load and dropping height was kept constant for all nineteen subjects. Therefore, the flexion motion was the result of sudden impulse load.

More set of weights and dropping heights for different subjects based on their own body weight percentage could be used to better understand the muscle response as a function of dropping energy. The torso flexion response might be more controllable and easy to evaluate.

In the sudden loading test, 10% body weight and a fixed height were used to study the muscle reactions. The muscle may react differently depending on different magnitude and type of load. Other types of load like continuous loading or random loading and their corresponding muscle reaction may give additional information. Because of the time constraints, all these could not be addressed in one study.

## 4.3 Conclusion

The effect of whole body of horizontal seat pan vibration on proprioception and muscular response due to sudden load which results in low back injury were observed in this research and compared the results with previously studied whole body vertical seat pan vibration study. Increases in aAbsolute position sense error and time to sudden loading muscular response, although not statistically significant, may lead to inappropriate stabilization of the spine which was observed by the significant increase both in muscle delay and flexion, thus making the spine unstable after horizontal vibration exposure. However, the small increases observed in this study, relative to those previously observed with a similar vertical vibration exposure, suggest that vertical vibration, at least at 5 Hz, may result in greater risk to workers exposed to whole body vibration.

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# Appendix I

## Matlab Program to calculate Reposition Sense Error

```
% Specify the subject, condition and target lumbar curvature

conditionvib(1).lab = 'pre';
conditionvib(2).lab = 'postv';
conditionvib(3).lab = 'postw';
subjectnum = '#';
target = -#;

%for each condition
clear dataout
for connum = 1:3
    %for each training trial
    for traintr = 1:5
        %open file
        filename = ['s' subjectnum 'v_' conditionvib(connum).lab '_ps_t' num2str(traintr)
'.exp'];
        f=fopen(filename);
        HZ = 1500;
        fclose(f);
        data = dlmread(filename,'\t',9,0); % skip the 9 description rows
        %data= data(length(data)- 5*1500: length(data), :);
        [N, w] = size(data);
        time = (0:N-1)'/HZ;

        % Get bird data and calculate lumbar angle
        count = data(:,1);
        sens(1).x = data(:,2);
        sens(1).y = data(:,3);
        sens(1).z = data(:,4);
        sens(2).x = data(:,5);
        sens(2).y = data(:,6);
        sens(2).z = data(:,7);
        sens(3).x = data(:,8);
        sens(3).y = data(:,9);
        sens(3).z = data(:,10);

        sens(1).q0 = data(:,11);
        sens(1).q1 = data(:,12);
        sens(1).q2 = data(:,13);
        sens(1).q3 = data(:,14);
        sens(2).q0 = data(:,15);
        sens(2).q1 = data(:,16);
        sens(2).q2 = data(:,17);
```

```
   sens(2).q3 = data(:,18);
   sens(3).q0 = data(:,19);
   sens(3).q1 = data(:,20);
   sens(3).q2 = data(:,21);
   sens(3).q3=  data(:,22);

   lc = data(:,23);
   cs = data(:,24);
   rawRES = data(:,25);
   rawLES = data(:,26);

   getflex
   figure(1)
   subplot(5,1,traintr)
   plot(Lordosis)

   % Taking the mean of  flexion and lumbar angle

   Torsoflex = mean(Torsoang);
   Lumbarang = mean(Lordosis);

   dataout(traintr) = Torsoflex;
   dataout(traintr+8) = Lumbarang;
   dataout(traintr+16) = Lumbarang - target;
end


% For each assessment trial
for assesstr = 1:3
  filename = ['s' subjectnum 'v_' conditionvib(connum).lab '_ps_a' num2str(assesstr)
    '.exp'];

   f=fopen(filename);
   HZ = 1500;
   fclose(f);

   data = dlmread(filename,'\t',9,0); % skip the 9 description rows

   %data= data(length(data)- 5*1500: length(data), :);

   [N, w] = size(data);
   time = (0:N-1)'/HZ;

   count = data(:,1);
   sens(1).x = data(:,2);
   sens(1).y = data(:,3);
```

```
        sens(1).z = data(:,4);
        sens(2).x = data(:,5);
        sens(2).y = data(:,6);
        sens(2).z = data(:,7);
        sens(3).x = data(:,8);
        sens(3).y = data(:,9);
        sens(3).z = data(:,10);

        sens(1).q0 = data(:,11);
        sens(1).q1 = data(:,12);
        sens(1).q2 = data(:,13);
        sens(1).q3 = data(:,14);
        sens(2).q0 = data(:,15);
        sens(2).q1 = data(:,16);
        sens(2).q2 = data(:,17);
        sens(2).q3 = data(:,18);
        sens(3).q0 = data(:,19);
        sens(3).q1 = data(:,20);
        sens(3).q2 = data(:,21);
        sens(3).q3=  data(:,22);

        lc = data(:,23);
        cs = data(:,24);
        rawRES = data(:,25);
        rawLES = data(:,26);

        getflex
        figure(2)
        subplot(5,1,assesstr)
        plot(Lordosis)

        % Taking the mean of  flexion and lumbar angle

        Torsoflex = mean(Torsoang);
        Lumbarang = mean(Lordosis);

        dataout(assesstr+5) = Torsoflex;
        dataout(assesstr+13) = Lumbarang;
        dataout(assesstr+21) = Lumbarang - target;
    end
    % Saving data
    eval(['save positionsense_' conditionvib(connum).lab '.txt dataout -ascii'])

end
```

## Matlab program to calculate time to peak muscle response from sudden loading experiment.

```
% Specify the subject and condition
subjectnum = '#';
conditionvib(1).lab = 'pre';
conditionvib(2).lab = 'postv';
conditionvib(3).lab = 'postw';

for connum = 1:3
clear dataoutsl
conditionhere = conditionvib(connum).lab;
run = '1';
printopt = 0;  %1 for print, 0 for no print

%Open and Read the file
filename = ['s' subjectnum 'v_' conditionhere '_sl_' run '.exp']
f=fopen(filename);
HZ = 1500;
fclose(f);
data = dlmread(filename,'\t',9,0); % skip the 9 description rows
[N, w] = size(data);
time = (0:N-1)'/HZ;


% Get bird data and calculate lumbar angle

count = data(:,1);
sens(1).x = data(:,2);
sens(1).y = data(:,3);
sens(1).z = data(:,4);
sens(2).x = data(:,5);
sens(2).y = data(:,6);
sens(2).z = data(:,7);
sens(3).x = data(:,8);
sens(3).y = data(:,9);
sens(3).z = data(:,10);

sens(1).q0 = data(:,11);
sens(1).q1 = data(:,12);
sens(1).q2 = data(:,13);
sens(1).q3 = data(:,14);
sens(2).q0 = data(:,15);
sens(2).q1 = data(:,16);
sens(2).q2 = data(:,17);
```

```
sens(2).q3 = data(:,18);
sens(3).q0 = data(:,19);
sens(3).q1 = data(:,20);
sens(3).q2 = data(:,21);
sens(3).q3=  data(:,22);

lc = data(:,23);
cs = data(:,24);
rawRES = data(:,25);
rawLES = data(:,26);

%Process Bird and EMG data
getflex
[iRES,fRES] = EMGprocess1500(rawRES);
[iLES,fLES] = EMGprocess1500(rawLES);
load Maxesout.txt
iRES = iRES/Maxesout(1)*100;
iLES = iLES/Maxesout(2)*100;
imeanES = (iLES+iRES)./2;



subplot(4,1,1)
plot(time,lc,time,cs)
subplot(4,1,2)
plot(time,Lordosis)
subplot(4,1,3)
plot(time, fRES, 'y')
hold on
plot(time,iRES,'k')
subplot(4,1,4)
plot(time, fLES, 'y')
hold on
plot(time,iLES,'k')

% Get when contact switch jumps 1 V in 5 points.
startcs=[];
countstart = 0;
for i=500:length(cs)
   c=1;
   if ((cs(i)-cs(i-5)>1)& (cs(i-100) < 1)& (cs(i+30) > 1))
      if (countstart<1)
      countstart = countstart +1;
      startcs(countstart) = i;
      elseif (i-(startcs(countstart))>3000)  % 2500 is the space expected between contact
switches
```

```
        countstart = countstart +1;
        startcs(countstart) = i;
        end
    end
end

figure(100)
clf
plot(cs)
title(['Contact Switch and Selected Start Times ' conditionhere])
hold on
plot(lc*5)
for i = 1:countstart
plot([startcs(i) startcs(i)],[min(cs) max(cs)],'r','linewidth',5)
end
if (printopt==1)
    print
end
if ((startcs(countstart)+2000)>length(cs))
    countstart = countstart-1;
end

% Get peaks for EMG, flexion and lumbar curvature
for i = 1:countstart
    timestart = startcs(i);
    figure(i)
    clf
    [RES.maxPeak,RES.indexPeak] = max(iRES((timestart+100):(timestart+900)));
    RES.indexPeak=RES.indexPeak+100;
    %   find max between 100 pts after the contact switch and 900 points
    %   after the contact switch for the RES
    RES.timePeak = RES.indexPeak/1500;
    RES.preparatory = mean(iRES(timestart-500:timestart));
    RES.delta = RES.maxPeak-RES.preparatory;

    [LES.maxPeak,LES.indexPeak] = max(iLES((timestart+100):(timestart+900)));
    LES.indexPeak=LES.indexPeak+100;
    %   find max between 100 pts after the contact switch and 900 points
    %   after the contact switch for the LES
    LES.timePeak = LES.indexPeak/1500;
    LES.preparatory = mean(iLES(timestart-500:timestart));
    LES.delta = LES.maxPeak-LES.preparatory;

    [meanES.maxPeak,meanES.indexPeak] =
max(imeanES((timestart+100):(timestart+900)));
    meanES.indexPeak=meanES.indexPeak+100;
```

```
    % find max between 100 pts after the contact switch and 900 points
    % after the contact switch for the average ES
    meanES.timePeak = meanES.indexPeak/1500;
    meanES.preparatory = mean(imeanES(timestart-500:timestart));
    meanES.delta = meanES.maxPeak-meanES.preparatory;

    subplot(4,1,1)
    plot(cs(timestart-200:(timestart+2000)))
    title(['Trial' num2str(i) ' ' conditionhere])
    ylabel('CS LC')
    hold on
    plot(lc(timestart-200:(timestart+2000))*5)
    plot([200 200],[min(cs) max(cs)],'r','linewidth',5)

    subplot(4,1,2)
    plot(iRES(timestart-200:(timestart+2000)),'b')
    ylabel('EMG')
    hold on
    plot([RES.indexPeak+200 RES.indexPeak+200],[RES.preparatory RES.maxPeak],'b')
    plot(iLES(timestart-200:(timestart+2000)),'m')
    plot([LES.indexPeak+200 LES.indexPeak+200],[LES.preparatory LES.maxPeak],'m')
    plot(imeanES(timestart-200:(timestart+2000)),'k')
    plot([meanES.indexPeak+200 meanES.indexPeak+200],[meanES.preparatory
meanES.maxPeak],'k')

    [flexion.maxPeak,flexion.indexPeak] = max(Torsoang(timestart:(timestart+2000)));
    flexion.timePeak = flexion.indexPeak/1500;
    flexion.preparatory = mean(Torsoang(timestart-500:timestart));
    flexion.delta = flexion.maxPeak-flexion.preparatory;

    subplot(4,1,3)
    plot(Torsoang(timestart-200:(timestart+2000)))
    ylabel('Flexion')
    hold on
    plot([flexion.indexPeak+200 flexion.indexPeak+200],[flexion.preparatory
flexion.maxPeak],'b')


    [lumbarang.maxPeak,lumbarang.indexPeak] =
   max(Lordosis(timestart:(timestart+2000)));
    lumbarang.timePeak = lumbarang.indexPeak/1500;
    lumbarang.preparatory = mean(Lordosis(timestart-500:timestart));
    lumbarang.delta = lumbarang.maxPeak-lumbarang.preparatory;

    subplot(4,1,4)
    plot(Lordosis(timestart-200:(timestart+2000)))
```

```
   ylabel('Lumbar')
   xlabel('Index Count')
   hold on
   plot([lumbarang.indexPeak+200 lumbarang.indexPeak+200],[lumbarang.preparatory
lumbarang.maxPeak],'b')
   if (printopt==1)
   print
   end
   dataoutsl(i,:) = [RES.indexPeak RES.timePeak RES.preparatory RES.delta ....
      LES.indexPeak LES.timePeak LES.preparatory LES.delta ...
      meanES.indexPeak meanES.timePeak meanES.preparatory meanES.delta ...
      flexion.indexPeak flexion.timePeak flexion.preparatory flexion.delta ...
      lumbarang.indexPeak lumbarang.timePeak lumbarang.preparatory lumbarang.delta];

end

%Save the data to a file
eval(['save SLdata_' conditionhere '.txt dataoutsl -ascii'])

end
```

## Matlab program to calculate average max to process EMG data

```
clear

%specify subject
subjectnum = '#';

% Open and read max file
filename = ['s' subjectnum 'v_maxes_ES.exp']
f=fopen(filename);
HZ = 1500;
fclose(f);
data = dlmread(filename,'\t',9,0); % skip the 9 description rows
[N, w] = size(data);
time = (0:N-1)'/HZ;


% Process EMG data
count = data(:,1);
rawRES = data(:,25);
rawLES = data(:,26);

[iRES,fRES] = EMGprocess1500(rawRES);
[iLES,fLES] = EMGprocess1500(rawLES);


plot(iRES,'k')

%Have user pick beginning and eng of maxes from plot (right)
for i=1:3
  [x,y] = ginput(2);
  x = round(x);
  maxemgmaxR(i) = mean(iRES(x(1):x(2)));
end

avgmaxR= mean(maxemgmaxR)


clf
plot(iLES,'k')
%Have user pick beginning and eng of maxes from plot (left)
for i=1:3
  [x,y] = ginput(2);
  x = round(x);
  maxemgmaxL(i) = mean(iLES(x(1):x(2)));
```

```
end

avgmaxL= mean(maxemgmaxL)

dataout = [avgmaxR avgmaxL];
save Maxesout.txt dataout –ascii
```

## Matlab subprogram ‘getflex’ to calculate Reposition Sense Error

```
clear latang flexang twistang

for j= 1:length(data(:,1))
    thor = [sens(1).x(j) sens(1).y(j) sens(1).z(j)];
    sacr = [sens(2).x(j) sens(2).y(j) sens(2).z(j)];
    manu = [sens(3).x(j) sens(3).y(j) sens(3).z(j)];

    ST = sacr-thor;
    ST = ST/sqrt(dot(ST,ST));
    ainv(3,1:3) = ST;

    MT = manu-thor;
    MT = MT/sqrt(dot(MT,MT));

    N1 = cross(ST,MT);
    N1 = N1/sqrt(dot(N1,N1));
    ainv(2,1:3) = N1;

    N2 = cross(N1,ST);
    N2 = N2/sqrt(dot(N2,N2));
    ainv(1,1:3) = N2;

    A = inv(ainv);
            a11 = A(1,1);
        a12 = A(1,2);
        a13 = A(1,3);
        a21 = A(2,1);
        a22 = A(2,2);
        a23 = A(2,3);
        a31 = A(3,1);
        a32 = A(3,2);
        a33 = A(3,3);
            lat = asin(a32);
    flex = atan((-1)*a31./a33);
    twist = atan((-1)*a12./a22);

        if flex>1
            flex = flex-3.14159;
        end

    latang(j,4) = lat;
    flexang(j,4) = flex;
    twistang(j,4) = twist;
```

```
end
% defining matrix
for i= 1:3
    quaterns = sens(i);
    a11=2.*quaterns.q0.*quaterns.q0-1+2.*quaterns.q1.*quaterns.q1;
    a12=2.*(quaterns.q1.*quaterns.q2-quaterns.q0.*quaterns.q3);
    a13=2.*(quaterns.q1.*quaterns.q3+quaterns.q0.*quaterns.q2);
    a21=2.*(quaterns.q1.*quaterns.q2+quaterns.q0.*quaterns.q3);
    a22=2.*quaterns.q0.*quaterns.q0-1+2.*quaterns.q2.*quaterns.q2;
    a23=2.*(quaterns.q2.*quaterns.q3-quaterns.q0.*quaterns.q1);
    a31=2.*(quaterns.q1.*quaterns.q3-quaterns.q0.*quaterns.q2);
    a32=2.*(quaterns.q2.*quaterns.q3+quaterns.q0.*quaterns.q1);
    a33=2.*quaterns.q0.*quaterns.q0-1+2.*quaterns.q3.*quaterns.q3;

    for j=1:length(a11)
        A = [a11(j) a12(j) a13(j); a21(j) a22(j) a23(j);a31(j) a32(j) a33(j)];
        if i == 1
            %A = A*[0 0 1;0 1 0; -1 0 0];  tail down
            A = A*[0 0 1;0 -1 0; 1 0 0];     %tail up
            a11(j) = A(1,1);
            a12(j) = A(1,2);
            a13(j) = A(1,3);
            a21(j) = A(2,1);
            a22(j) = A(2,2);
            a23(j) = A(2,3);
            a31(j) = A(3,1);
            a32(j) = A(3,2);
            a33(j) = A(3,3);
        elseif i ==2
            A = A*[0 0 1;0 -1 0; 1 0 0];
            a11(j) = A(1,1);
            a12(j) = A(1,2);
            a13(j) = A(1,3);
            a21(j) = A(2,1);
            a22(j) = A(2,2);
            a23(j) = A(2,3);
            a31(j) = A(3,1);
            a32(j) = A(3,2);
            a33(j) = A(3,3);
        end
    end
    lat = asin(a32);
    flex = atan((-1)*a31./a33);
    twist = atan((-1)*a12./a22);

    for j=1:length(flex)
```

```
        if flex(j)>1 % changed from .05 to 1
            flex(j) = flex(j)-3.14159;
        end


    end

    latang(:,i) = lat;
    flexang(:,i) = flex;
    twistang(:,i) = twist;

end

% defining Lordosis and Torso flexion angle
Lordosis = ((-flexang(:,1)+flexang(:,2))*180/3.1415);
Torsoang = ((-flexang(:,4))*180/3.1415);
```

## Matlab sub-program EMG Process 1500

```
function [dataout,femg0] = EMGprocess(datain)

file_length = length(datain(:,1));
femg = datain;
%EMG processing
  %filter data-- Highpass filter, 30 Hz cutoff
  Wp = 30/750;
  Ws = 20/750;
  [n,Wn] = buttord(Wp,Ws,.01,10);
  [b,a]  = butter(n,Wn,'high');
        femg = filter(b,a,datain(file_length:-1:1,:));
  femg = filter(b,a,femg(file_length:-1:1,:));


  %filter data-- Lowpass filter, 250 Hz cutoff
  Wp = 250/750;
  Ws = 270/750;
  [n,Wn] = buttord(Wp,Ws,.01,20);
  [b,a]  = butter(n,Wn,'low');
        femg = filter(b,a,femg(file_length:-1:1,:));
  femg = filter(b,a,femg(file_length:-1:1,:));

  %filter data-- Bandstop filter @ 60 Hz (Ascension signal)
  Wp = [58,62]/750;
  Ws = [54,66]/750;
  [n,Wn] = buttord(Wp,Ws,.01,10);
  [b,a]  = butter(n,Wn,'stop');
        femg = filter(b,a,femg(file_length:-1:1,:));
  femg = filter(b,a,femg(file_length:-1:1,:));

  %filter data-- Bandstop filter @ 40 Hz (Ascension signal)
  Wp = [38,42]/750;
  Ws = [34,46]/750;
  [n,Wn] = buttord(Wp,Ws,.01,10);
  [b,a]  = butter(n,Wn,'stop');
   femg = filter(b,a,femg(file_length:-1:1,:));
  femg = filter(b,a,femg(file_length:-1:1,:));

 %filter data-- Bandstop filter @ 80 Hz (Ascension signal)
  Wp = [78,82]/750;
  Ws = [74,86]/750;
  [n,Wn] = buttord(Wp,Ws,.01,10);
  [b,a]  = butter(n,Wn,'stop');
   femg = filter(b,a,femg(file_length:-1:1,:));
```

```
femg = filter(b,a,femg(file_length:-1:1,:));

%filter data-- Bandstop filter @ 120 Hz (Ascension signal)
Wp = [118,122]/750;
Ws = [114,126]/750;
[n,Wn] = buttord(Wp,Ws,.01,10);
[b,a]  = butter(n,Wn,'stop');
        femg = filter(b,a,femg(file_length:-1:1,:));
femg = filter(b,a,femg(file_length:-1:1,:));



%mean data
        femg0 = femg - repmat(mean(femg),[file_length,1]);
```

## Matlab sub-program hfilter

```
function Y=hfilter(X,BW)

% Hanning window routine

[len,col] = size(X);

wt = cos(pi/2*(-1+1/BW:2/BW:1));
norm(1) = wt(1);
norm(len+BW-1) = norm(1);

for i=2:BW
	norm(i) = norm(i-1) + wt(i);
	norm(len+BW-i) = norm(i);
	end;
norm(BW+1:len) = norm(BW)*ones(1,len-BW);

for i=2:col
	norm(i,:) = norm(1,:);
	end;

Xf = zeros(len+BW-1,col);
for i=1:BW
	Xf = Xf + wt(i)*[zeros(BW-i,col);X;zeros(i-1,col)];
	end;

Xf = Xf./norm';

Y = Xf((BW-1)/2+1:len+(BW-1)/2,:);
```

# Appendix II

**Table 12 List of Absolute Reposition Sense Error in assessment trials for nineteen subjects in pre, postW and postV condition.**

| Subject | Pre | | | PostW | | | PostV | | |
|---|---|---|---|---|---|---|---|---|---|
| | a1 | a2 | a3 | a1 | a2 | a3 | a1 | a2 | a3 |
| 1 | 0.83 | 1.89 | 1.64 | 9.12 | 2.64 | 0.10 | 8.06 | 7.89 | 1.32 |
| 2 | 3.24 | 1.37 | 5.53 | 4.47 | 0.47 | 2.79 | 0.43 | 1.09 | 1.52 |
| 3 | 3.66 | 6.76 | 6.46 | 2.74 | 1.92 | 0.02 | 9.03 | 2.75 | 0.38 |
| 4 | 3.79 | 0.14 | 1.81 | 2.32 | 1.64 | 2.31 | 2.87 | 0.36 | 3.77 |
| 5 | 5.40 | 1.05 | 4.61 | 5.98 | 2.43 | 7.19 | 6.41 | 1.50 | 2.95 |
| 6 | 5.33 | 2.85 | 2.40 | 1.06 | 0.53 | 0.28 | 1.79 | 1.94 | 0.52 |
| 7 | 0.81 | 0.55 | 2.96 | 2.30 | 0.56 | 0.48 | 4.41 | 1.52 | 4.42 |
| 8 | 0.41 | 0.75 | 0.66 | 3.54 | 2.30 | 0.57 | 0.30 | 2.46 | 1.91 |
| 9 | 2.05 | 5.77 | 3.00 | 2.98 | 1.74 | 0.85 | 1.31 | 0.37 | 3.42 |
| 10 | 0.33 | 0.75 | 1.16 | 1.45 | 0.03 | 3.35 | 0.07 | 0.93 | 0.33 |
| 11 | 0.01 | 1.58 | 0.43 | 0.47 | 3.32 | 1.57 | 1.45 | 5.06 | 3.32 |
| 12 | 2.28 | 2.14 | 2.79 | 0.08 | 2.62 | 2.95 | 0.78 | 2.55 | 2.55 |
| 13 | 0.73 | 0.49 | 4.74 | 2.87 | 6.51 | 0.68 | 1.12 | 0.89 | 4.48 |
| 14 | 4.19 | 6.57 | 6.62 | 0.73 | 1.51 | 0.56 | 1.81 | 0.74 | 0.69 |
| 15 | 0.33 | 2.42 | 2.02 | 1.01 | 3.14 | 3.82 | 2.28 | 3.55 | 6.76 |
| 16 | 5.71 | 4.75 | 4.07 | 2.52 | 1.74 | 3.19 | 0.46 | 0.55 | 1.77 |
| 17 | 0.61 | 4.72 | 3.13 | 2.84 | 1.07 | 0.15 | 0.85 | 1.78 | 2.74 |
| 18 | 4.73 | 5.48 | 2.31 | 3.40 | 1.71 | 3.84 | 3.19 | 4.04 | 2.95 |
| 19 | 1.16 | 0.16 | 0.62 | 0.15 | 3.54 | 5.86 | 3.13 | 0.40 | 0.87 |

**Table 13 Average Absolute Reposition Sense Error in assessment trials for nineteen subjects in pre, postW and postV condition.**

| Sub | Pre | PostW | PostV |
|---|---|---|---|
| 1 | 1.46 | 3.95 | 5.76 |
| 2 | 3.38 | 2.58 | 1.01 |
| 3 | 5.63 | 1.56 | 4.05 |
| 4 | 1.91 | 2.09 | 2.33 |
| 5 | 3.69 | 5.20 | 3.62 |
| 6 | 3.53 | 0.62 | 1.42 |
| 7 | 1.44 | 1.11 | 3.45 |
| 8 | 0.61 | 2.14 | 1.56 |
| 9 | 3.61 | 1.86 | 1.70 |
| 10 | 0.74 | 1.61 | 0.45 |
| 11 | 0.68 | 1.79 | 3.28 |
| 12 | 2.40 | 1.88 | 1.96 |
| 13 | 1.99 | 3.35 | 2.16 |
| 14 | 5.79 | 0.93 | 1.08 |
| 15 | 1.59 | 2.66 | 4.19 |
| 16 | 4.85 | 2.48 | 0.93 |
| 17 | 2.82 | 1.35 | 1.79 |
| 18 | 4.17 | 2.98 | 3.39 |
| 19 | 0.65 | 3.19 | 1.47 |

**Table 14 Time to peak muscle response delay (ms) for Erector Spinae Muscle in pre, postW and postV condition.**

| Sub | Pre | PostW | PostV |
|---|---|---|---|
| 1 | 122.93 | 88.13 | 90.53 |
| 2 | 148.67 | 90.00 | 123.60 |
| 3 | 111.33 | 86.80 | 129.33 |
| 4 | 148.67 | 104.80 | 136.53 |
| 5 | 190.53 | 119.60 | 213.87 |
| 6 | 111.33 | 126.27 | 131.87 |
| 7 | 129.47 | 132.00 | 142.93 |
| 8 | 198.40 | 232.53 | 177.07 |
| 9 | 123.60 | 97.07 | 93.87 |
| 10 | 111.33 | 115.00 | 99.07 |
| 11 | 83.47 | 128.13 | 107.47 |
| 12 | 90.67 | 94.53 | 90.67 |
| 13 | 104.93 | 102.13 | 121.07 |
| 14 | 113.20 | 105.47 | 97.07 |
| 15 | 100.40 | 136.33 | 143.60 |
| 16 | 98.40 | 101.60 | 112.00 |
| 17 | 136.40 | 126.80 | 113.87 |
| 18 | 104.93 | 100.27 | 104.13 |
| 19 | 99.73 | 126.27 | 135.87 |

**Table 15 Torso flexion magnitude for nineteen subjects in pre, postW and postV condition.**

| Sub | Pre | PostW | PostV |
|---|---|---|---|
| 1 | 1.53 | 3.74 | 2.23 |
| 2 | 13.67 | 15.11 | 10.56 |
| 3 | 13.37 | 17.15 | 16.93 |
| 4 | 15.75 | 17.24 | 12.40 |
| 5 | 17.92 | 16.82 | 15.62 |
| 6 | 16.19 | 19.64 | 16.95 |
| 7 | 20.08 | 19.13 | 16.73 |
| 8 | 57.26 | 42.01 | 40.88 |
| 9 | 15.01 | 8.78 | 7.66 |
| 10 | 29.81 | 28.38 | 39.70 |
| 11 | 16.14 | 14.99 | 16.81 |
| 12 | 23.17 | 26.99 | 25.79 |
| 13 | 31.74 | 34.69 | 33.96 |
| 14 | 23.23 | 32.25 | 38.52 |
| 15 | 13.21 | 20.26 | 25.08 |
| 16 | 15.52 | 26.78 | 24.64 |
| 17 | 21.32 | 17.21 | 18.69 |
| 18 | 20.30 | 15.38 | 16.95 |
| 19 | 17.94 | 24.04 | 25.78 |

# Appendix IV

# Subject Consent Form

**INTRODUCTION**
The Department of Mechanical Engineering at the University of Kansas supports the practice of protection for human subjects participating in research. The following information is provided for you to decide whether you wish to participate in the present study. You may refuse to sign this form and not participate in this study. You should be aware that even if you agree to participate, you are free to withdraw at any time. If you do withdraw from this study, it will not affect your relationship with this unit, the services it may provide to you, or the University of Kansas.

**PURPOSE OF THE STUDY**
We are interested in evaluating how truck driver and other workers who are exposed to

vibration move and how their reflexes change.

**PROCEDURES**
If you choose to participate, we will first give you a health questionnaire to make sure you do not have any heart problems or back injuries that might make it difficult to do the experiment.
Magnetic markers will be taped or strapped to your back. The markers are used to sense how you move. In addition we will put electromyographic sensors on your back that will measure what you muscles are doing.
While wearing these markers, you will be asked to participate in two kinds of measurements. This first measurement measures your ability to sense position. You will be asked to match your posture to a display on a computer screen. You will then be asked to repeat that posture once the display has been turned off.
The second measurement measure your back reflexes. You will be asked to stand in an apparatus that will hold your hips still. You will be asked to wear a chest harness that is attached to a rope. After a random amount of time, the rope will be pulled by having a small weight drop. The pull from the rope should be just enough to pull you forward an inch or so. We will demonstrate this for you before the experiment so you can feel comfortable with the measurement.
Then you will be asked to seat in a vibrating chair for 30 minutes. The vibrating chair will vibrate less than 1/2 inch. The vibration will be like you might experience if you sat on your dryer. After vibration you will be asked to repeat the measurements and after that you will be seated in chair for another 30 mnts and repeat the measurements again.
You will be asked to repeat these measurements three times before and after sitting in a chair for 20 minutes. Your participation is strictly voluntary and you can stop at anytime. We assure that your name will not be associated in any way with the research findings. This protocol will take approximately two hours to complete.

**RISKS**
Some people have allergies to adhesives such as in band-aids or in the tape we are using to attach the markers.

**BENEFITS**
With this research we hope to be able to understand what happens to truck drivers and similar workers. We believe that understand how a person changes how they move after vibration will tell us something about why these workers get injured more often. There is, however, no direct benefit for the subject of this study.

**PAYMENT TO PARTICIPANTS**
Subjects will receive $20 for participation in the study.

**INFORMATION TO BE COLLECTED**
To perform this study, researchers will collect information about you. This information will be obtained from a questionnaire that will assess if you have heart or back problems that might make exercise inadvisable. Also, information will be collected from the study activities that are listed in the Procedures section of this consent form. This includes information about how you walk, your height and your weight.
Your name will not be associated in any way with the information collected about you or with the research findings from this study. The researcher(s) will use a study number instead of your name.
In addition, Dr. Wilson and her team may share the information gathered in this study, including your information, with the Whitaker Foundation that is funding the study. Again, your name would not be associated with the information disclosed to these individuals. Some persons or groups that receive your information may not be required to comply with the Health Insurance Portability and Accountability Act's privacy regulations, and your information may lose this federal protection if those persons or groups disclose it.

The researchers will not share information about you with anyone not specified above unless required by law or unless you give written permission.

Permission granted on this date to use and disclose your information remains in effect indefinitely. By signing this form you give permission for the use and disclosure of your information for purposes of this study at any time in the future.

**INSTITUTIONAL DISCLAIMER STATEMENT**
In the event of injury, the Kansas Tort Claims Act provides for compensation if it can be demonstrated that the injury was caused by the negligent or wrongful act or omission of a state employee acting within the scope of his/her employment.

**REFUSAL TO SIGN CONSENT AND AUTHORIZATION**
You are not required to sign this Consent and Authorization form and you may refuse to do so without affecting your right to any services you are receiving or may receive from

the University of Kansas or to participate in any programs or events of the University of Kansas. However, if you refuse to sign, you cannot participate in this study.

**CANCELLING THIS CONSENT AND AUTHORIZATION**

You may withdraw your consent to participate in this study at any time. You also have the right to cancel your permission to use and disclose information collected about you, in writing, at any time, by sending your written request to: Dr. Sara Wilson, Mechanical Engineering, University of Kansas, Lawrence, KS 66045. If you cancel permission to use your information, the researchers will stop collecting additional information about you. However, the research team may use and disclose information that was gathered before they received your cancellation, as described above.

**PARTICIPANT CERTIFICATION:**

I have read this Consent and Authorization form. I have had the opportunity to ask, and I have received answers to, any questions I had regarding the study and the use and disclosure of information about me for the study. I understand that if I have any additional questions about my rights as a research participant, I may call (785) 864-7429 or write the Human Subjects Committee Lawrence Campus (HSCL), University of Kansas, 2385 Irving Hill Road, Lawrence, Kansas 66045-7563, email dhann@ku.edu.

I agree to take part in this study as a research participant. I further agree to the uses and disclosures of my information as described above. By my signature I affirm that I am at least 18 years old and that I have received a copy of this Consent and Authorization form.

_______________________________ _______________________

Type/Print Participant's Name Date

_________________________________________

Participant's Signature

Researcher Contact Information

Sara E. Wilson
Principal Investigator
Mechanical Engineering
3013 Learned Hall
University of Kansas
Lawrence, KS 66045
785 864-2103

# Medical Questionnaires

**Medical History**

Subject Number: ______________________________________________
Age: ______________________________________________
Height: ______________________________________________
Weight: ______________________________________________

Do you have any history of cardiovascular (heart) disease?

Have you ever had any of the following (circle any that you have experienced)?

Prolapsed Heart Valve
Heart Murmur
Myocardial Infarction (heart attack)
Angiography
Chest Pain
Hypertension (high blood pressure)
Shortness of Breath on Exertion
Pulmonary (lung) Disease
Dizziness on Light Exertion
Claudication (pain in arms and legs during light exertion)
Diabetes
Fainting
Seizures

Are you on any current medications? If yes, what?

Have you ever had pain in your low back for more than one week? Have you had any instances of low back pain within the last year? If yes, describe.

Do you currently have any musculoskeletal injuries (sprains, broken bones, sore muscles…)? If yes, describe.

When did you eat your last meal?.