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A Quasi-static Discomfort Measure in Whole-body Vibration

Salam RAHMATALLA[1, 2*], Rosalind SMITH[2], John MEUSCH[2], Ting XIA[3], Tim MARLER[2] and Michael CONTRATTO[4]

[1]Department of Civil and Environmental Engineering, 4121 Seamans Center for Engineering, The University of Iowa, Iowa City, Iowa 52242, USA
[2]Center for Computer-Aided Design, The University of Iowa, 330 S. Madison, Iowa City, Iowa 52242, USA
[3]Palmer College of Chiropratic, 741 Brady St, Davenport, Iowa 52803, USA
[4]Caterpillar Inc., Peoria, Illinois 61629, USA

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Abstract: A new methodology for objective evaluation of discomfort in whole-body vibration (WBV) is introduced in this work. The proposed objective discomfort characterizes discomfort based on the relative motion between adjacent segments of the human body from neutral positions. It peaks when the joints reach their limits. The objective discomfort has been tested on five subjects in the fore-aft direction using discrete sinusoidal frequencies of 0.5, 1, 1.5, 2, 2.5, 3, 3.5, 4, 4.5, 5, 5.5, 6, 7, 8, 9, 10, 12, 14, and 16 Hz. Each frequency file runs for 15 s with a 3 s resting period as a reference for discomfort comparison. All files run at a constant acceleration of 0.7 m/s^2. The subjects were tested with back support and without back support, and their subjective discomfort was reported based on the Borg CR-10 scale. The proposed objective discomfort has shown significant correlation with the subjective discomfort. The objective discomfort has also been tested on five subjects under multiple-axis random WBV with three common industrial seating configurations (seat-mounted control, floor-mounted control, and steering wheel), and has shown promising results.

Key words: Objective discomfort, Subjective discomfort, Whole-body vibration, Sitting posture, Seating configuration

Introduction

It is well known that sitting posture is associated with musculoskeletal discomfort and a number of disorders such as low back pain[1–3]. In whole-body vibration (WBV) encountered in aircrafts, ships, automobiles, farming machinery, construction equipment, army vehicles, and other moving environments, the problem becomes more acute as operators are also subjected to a multiple-axis form of vibration[4]. In such environments, the operators normally engage with their surrounding equipment, seat, pedal, and controls, with energy entering their bodies from multiple locations and directions, resulting in a very complicated coupled motion. As a result, the relative motion between body segments sometimes reaches its critical discomfort region, causing, in some instances, a considerable degree of discomfort. However, due to a large number of parameters affecting the way energy is transferred into the human body, the process of characterizing discomfort using dynamic biomechanical measures such as different types of global transfer functions becomes complicated.

People, especially those who deal with the car industry, have been viewing discomfort as a static feature[5, 6]. In this regard, the discomfort can be correlated objectively with seat pressure, contact area, thermal effect, and electromyography activity (EMG). With WBV, the scenario is different because many parameters, such as frequency and human-machine interface, can affect discomfort. Also, there is a significant contribution of posture to discomfort measures in WBV[7–11]. However, all studies in this regard consider posture as a static form and instruct the subjects to maintain the same posture during the testing. Hinz *et al*.[9] conducted experiments on 39 male subjects sitting on suspension seats with and without backrests during vertical WBV and used a finite-element-based human model to calculate the internal spinal loads. The authors concluded that backrest and posture conditions play an important role and should be included in risk assessment during WBV. Further, Wang *et al*.[7] found a significant effect of sitting posture on the biodynamic response under vertical vibration after considering 36 different sitting postures and seat configurations. The results showed important combined effects of inclined backrest and hand position on the absorbed power characteristics. Thus, although they are somewhat varied, most prior studies have demonstrated the importance of considering seated postures when investigating WBV.

Traditionally, in single-axis WBV, discomfort is measured subjectively[12] using verbal or paper-based techniques such as the judgment method[13]. Many researchers, however, are more interested in finding objective measures that can be later embedded inside a computer human model to assess

*To whom correspondence should be addressed.
E-mail: salam-rahmatalla@uiowa.edu

discomfort and modify designs. Although extensive work has been done to develop objective discomfort measures, these measures are limited to a single-axis direction and still have difficulty dealing with dual- and multiple-axis cases due to the cross-axis coupling effect. Among the objective measures, the power absorbed[7, 14], the apparent mass[15], and the transmissibility[16] have shown encouraging and consistent correlations with the subjective discomfort measures. However, recent studies[17, 18] have shown the incapability of the apparent mass to capture the peak in the discomfort when the subjects move, such as twisting their torsos or lifting their arms, at the same time. Maeda *et al.*[18] showed that the peak in the apparent mass was reduced and even diminished in some cases. Subashi *et al.*[19] hypothesized that nonlinearity in the subjective responses would be related to the nonlinearity in the dynamic response represented here by the apparent mass. They studied such nonlinearity in the fore-aft and lateral directions and showed good correlation with the subjective measures at relatively lower frequencies.

The frequency weighting and the RMS averaging as specified in the ISO 2631-1[20] can be used to evaluate discomfort. However, there is ongoing debate on the validity of the ISO 2631-1 for several applications. Kaneko *et al.*[13] showed that when the random signals are applied as vibration stimuli, even if the frequency-weighted RMS acceleration by the ISO-2631-1 is the same, signals made up of different frequency spectra will elicit differing evaluations of the degree of comfort. Maeda *et al.*[18] used Kaneko's results and came up with an alternative approach that showed superior results to ISO 2631-1, but that still depends on the subjective measures. Maeda *et al.*[18] and Subashi *et al.*[19] showed the ineffectiveness of ISO standard 2631-1, which uses the frequency weighting of evaluating human exposure to vibration. The latter does not consider the effect of the frequency content in the signal but only uses the RMS weighted component of the signal, meaning that the vibration perception would be the same for two signals of the same magnitude but different frequency content.

The objective of this work was to introduce a new methodology for a theoretical objective evaluation of discomfort in WBV and use it to assess the discomfort in the fore-aft and multiple-axis WBV of dozer drivers. The objective discomfort model was validated in single axis (fore-aft) discrete frequency WBV. The objective discomfort has been also applied to more complex 3D WBV motion with three arm-support configurations to demonstrate its potential.

Methods

Single-axis (Fore-aft) vibration

Five healthy male subjects with a mean age of 24 yr (ranging from 19–29 yr), a mean stature of 188 cm (ranging from 180–196 cm), and a mean body mass of 84.5 kg (ranging from 71–98 kg) were recruited. Written informed consent, as approved by the University of Iowa Institutional Review Board, was obtained prior to testing. Subjects were seated in an uncushioned, rigid seat mounted to a vibration platform. Two sitting postures were considered, one with the subject sitting in a standard posture supported by the seat back, and the second in a forward unsupported upright posture. Vibration was generated using a six-degree-of-freedom man-rated vibration platform (Moog-FCS, Ann Arbor, MI, USA). Signals with a constant unweighted RMS acceleration magnitude of 0.7 m/s^2 were tested. Discrete frequencies from 0.5, 1, 1.5, 2, 2.5, 3, 3.5 4, 4.5, 5, 5.5, 6, 7, 8, 9, 10, 12, 14, and 16 Hz were chosen and randomized. In each frequency test, the subjects reported their perception of discomfort using the Borg CR-10 scale. Particularly, the subjects were exposed to 15 s of vibration at each frequency and a 3 s resting period in between. They rated their perceived discomfort during vibration by comparing it with their perception during the resting period. Twelve 0.3 megapixel Vicon SV cameras with a sampling rate of 200 frames per second were used in tracking the motion, while surface EMG of the cervical and lumbar erector spinae were collected using a Delsys system.

Multiple-axis vibration

Five healthy subjects with a mean age of 33.5 yr (ranging from 22 to 45 yr), a mean stature of 176.5 cm (ranging from 165–188 cm), and a mean body mass of 77.5 kg (ranging from 64–91 kg) were tested in a multiple-axis whole-body environment using a ride file (60 s) from a heavy construction machine, the Caterpillar D10 dozer. Written informed consent, as approved by the University of Iowa Institutional Review Board, was obtained prior to testing. A six-degree-of-freedom Servotest (Sears seating facility, Davenport, IA, USA) hydraulic motion platform was used in the testing. Eight 0.3 megapixel Vicon SV cameras with a sampling rate of 200 frames per second were used in tracking the motion; accelerometers were attached to the head and the torso areas. A seat with three arm-support configurations (Fig. 1) was used in this study: steering wheel (ST) with the steering wheel and the seat (with no arm support) attached separately to the motion platform, seat-mounted (SM) with the arm support attached to the seat, and floor-mounted (FM) with the arm support attached to the floor. An objective discomfort measure for the neck was calculated and used to compare the discomfort level between the three configurations. The subjects were asked about their perceptions of each configuration.

The input motion was measured by attaching reflective markers to the rigid part of the seat. The norm of the displacement vector from a marker was used to represent the magnitude of the input motion to the system. In WBV, the interest would be more in the frequency components of the signal. Therefore, the input motion file with 60 s containing 12,000 frames was segmented into windows, each of which contains 256 frames. With a sampling rate of 200 frames per second, each window represents 1.28 s of the ride time. The windows then mapped to the frequency domain using Matlab's fft function. The relationship between the frequency components of the input signal and the time of the signal was plotted using a contour plot in Matlab as shown in Fig. 2. In order to correlate the input and the output motions, the time domain of the output motion was also segmented with the same number of windows. The magnitude of the output motion was considered as the RMS value of the motion at each window.

Quasi-static discomfort

The concept of quasi-static discomfort in WBV is intro-

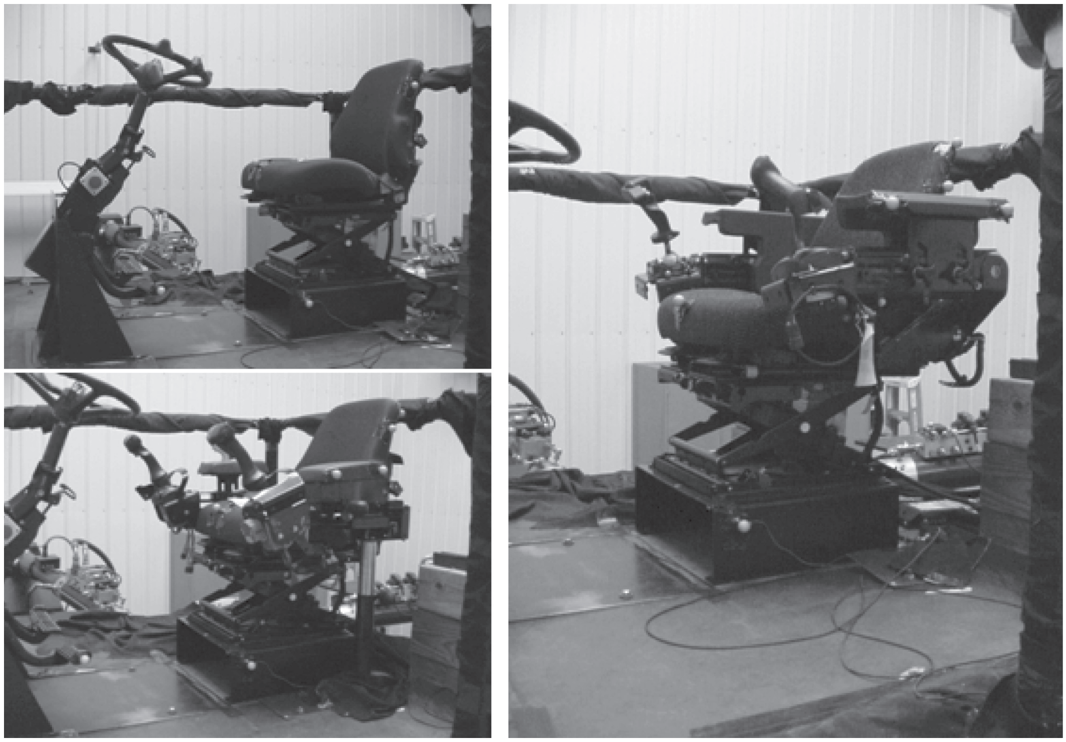

Fig. 1. Steering wheel (ST) control (upper left), floor-mounted control (FM) (lower left), and seat-mounted control (SM) (right).

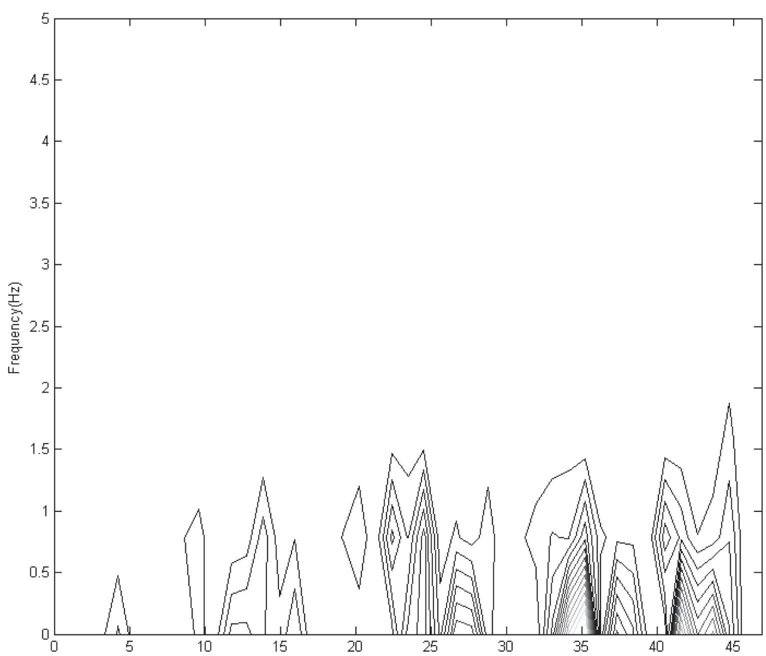

Fig. 2. A contour plot of the input motion with x-axis (time-window in 1.28 s), and y-axis (frequency in Hz).

duced in this work. The concept was modeled using posture-based discomfort as its foundation[21–23]. While quasi-static discomfort is not a new concept, it has generally been used to evaluate discomfort in static configurations. The discomfort performance measure incorporates three key factors associated with musculoskeletal discomfort: (1) the tendency to move different segments of the body sequentially, (2) the tendency to gravitate to a comfortable neutral position, and (3) the discomfort associated with moving while joints are near their respective limits. The neutral position is a relaxed comfortable configuration; when the body moves, it tries to move relatively close to this neutral position to minimize discomfort. This performance measure operates in real time and provides realistic postures. Discomfort has been categorized in the literature as static and dynamics discomfort and a combination of the two2[4, 25]. Both may find difficulties in capturing the discomfort associated with intentional or unintentional body movements during WBV. Essentially, the proposed objective discomfort is a quasi-static discomfort measure that could effectively capture the effect of the body postures/movements on discomfort during WBV. The objective discomfort has been proven to be very efficient for evaluating discomfort in seated reaching[21, 22]. This research lays groundwork for studying how and why humans move as they do. The fundamental discomfort model is described as follows:

Given the context of optimization-based posture prediction, three key factors are incorporated in the discomfort model, by drawing on multi-objective optimization (MOO). In order to incorporate the first factor, the tendency to move different segments of the body sequentially, the objective discomfort is based on the *lexicographic method* for MOO, which is discussed in detail by Marler and Arora[26]. With the lexicographic method, one simply prioritizes the objectives rather than articulating preferences with weights that indicate the relative importance of individual objective functions shown as follows in a traditional joint-displacement function:

$$f_{Joint\ displacement}(\mathbf{q}) = \sum_{i=1}^{DOF} w_i f_i(q_i) \tag{1}$$

$$f_i(q_i) = \left(q_i - q_i^N\right)^2 \tag{2}$$

where q is a vector of joint angles, q_i^N is the *neutral position* of a single joint, and the neutral position of the complete system, $\mathbf{q}^N$, represents a relatively comfortable position. With this formulation, the joint's position gravitates towards the neutral position. w_i represents scalar weights and is used to stress the importance of particular joints. Then, one objective at a time is minimized in a sequence of separate optimization problems. After an objective has been minimized, it is incorporated as a constraint in the subsequent problems. By using the concept behind the lexicographic method, one is able to model the idea that groups of joints are utilized sequentially. That is, in an effort to reach a particular target point, one first uses one's arm. Then, only if necessary, does one bend the torso. Finally, if the target is still out of reach, one may exercise the clavicle joint. Essentially, different groups of joints are included in one of three objective functions (one each for the arm, torso, and clavicle), which are then optimized lexicographically. Miettinen[27] and Romero[28] suggested that the weighted sum method can be used to approximate results of the lexicographic method if the weights have infinitely different orders of magnitude. This is the approach taken with the proposed objective discomfort. The weights γ_i, which are used to approximate the lexicographic approach, are shown in Table 1.

Table 1. Joint Weights for Discomfort

Joint Variables	γ_i
$q_1, \ldots, q_{12}$	1×10^4
$q_{13}, \ldots, q_{14}$	1×10^8
$q_{15}, \ldots, q_{21}$	1

There is one γ_i associated with each joint variable, but only three different values. Although weights are used, they do not need to be determined as indicators of the relative significance of their respective joints; they are simply fixed mathematical parameters. The exact values of the weights are irrelevant; they simply have to have significantly different orders of magnitude. An additional benefit is that this approach avoids computational difficulties associated with discontinuous values for the weights typically used in weighted sums, which are common in the literature.

The second discomfort factor, the tendency to gravitate to a reasonably comfortable neutral position, is incorporated by using the weights in Table 1 with a function that is based loosely on a weighted sum with the neutral position representing a posture with the arms straight down, parallel to the torso. Note that for this model, the objective functions should be normalized when weights are incorporated a priori. Consequently, prior to applying the weights, each joint term is normalized as follows:

$$\Delta q^{norm} = \frac{q_i - q_i^N}{q_i^U - q_i^L} \tag{3}$$

With this normalization scheme, each term $(\Delta q_i^{norm})^2$ acts as an individual objective function and has values between zero and one. The final aggregated discomfort function is given as follows:

$$f_{Discomfort}(\mathbf{q}) = \sum_{i=1}^{DOF} \gamma_i \left(\Delta q_i^{norm}\right)^2 \tag{4}$$

where γ_i are the weights defined in Table 1.

Generally, Eq. (4) is effective in modeling the tendency to move body segments sequentially and the tendency to gravitate towards a neutral position. However, it often results in postures with joints extended to their limits, and such postures can be uncomfortable and unrealistic. Consequently, to rectify this problem and to incorporate the final factor (the discomfort associated with moving while joints are near their respective limits), specially designed *penalty terms* are added to the discomfort function. Consequently, the modeled discomfort increases significantly as joint values approach their limits. The final discomfort function is given as follows:

$$f_{Discomfort}(\mathbf{q}) = \frac{1}{G} \sum_{i=1}^{DOF} \left[\gamma_i \left(\Delta q_i^{norm}\right) + G \times QU_i + G \times QL_i \right] \tag{5}$$

$$QU_i = \left(0.5sin\left(\frac{5.0\left(q_i^U - q_i\right)}{q_i^U - q_i^L} + 1.571\right) + 1\right)^{100} \tag{6}$$

$$QL_i = \left(0.5sin\left(\frac{5.0\left(q_i - q_i^L\right)}{q_i^U - q_i^L} + 1.571\right) + 1\right)^{100} \tag{7}$$

where $G \times QU$ is a penalty term associated with joint values that approach their upper limits, and $G \times QL$ is a penalty term associated with joint values that approach their lower limits. Each penalty term varies between zero and G, as the following two terms vary between zero and one:

$$\left(q_i^U - q_i\right) / \left(q_i^U - q_i^L\right) \tag{8}$$

$$\left(q_i - q_i^L\right) / \left(q_i^U - q_i^L\right) \tag{9}$$

Figure 3 illustrates the curve for the following function, which represents the basic structure of the penalty terms:

$$Q = \left(0.5sin\left(5.0r + 1.571\right) + 1\right)^{100} \tag{10}$$

r represents the expression in either Eq. (8) or Eq. (9). The penalty term has a value of zero until the joint value reaches the upper or lower 10% of its range (Fig. 3). The curve for the penalty term is differentiable, and it reaches its maximum penalty value of G=10^6 when r=0.

While it is possible to use the discomfort function to predict human postures for quasi-static postures, this concept is tailored in this work to calculate discomfort from postures and is modified for WBV by (1) solving the inverse kinematics problem at each frame in the vibration time history and calculating discomfort using the resulting joint angles, and (2) streaming the discomfort for each posture with time and computing a discomfort measure as a function of time.

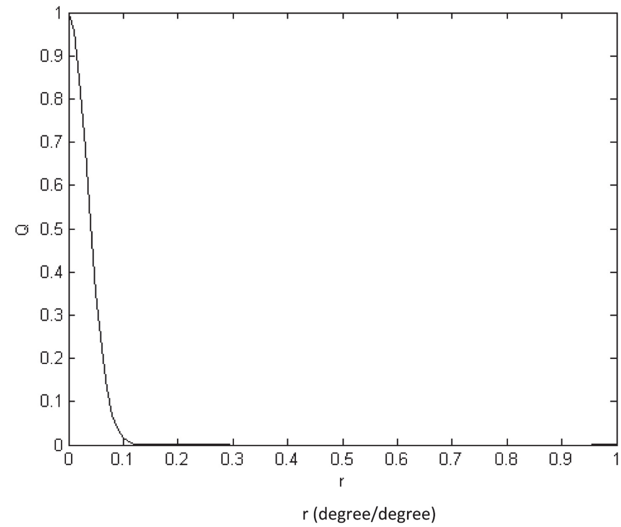

Fig. 3. Graph of discomfort joint-limit penalty dimensionless term (Q) in Equation 10 with the joint angle ratio (r) in Equations 8 and 9.

Results

Fore-aft WBV

The mean subjective discomfort curves for the frequency under consideration (Figs. 4 and 5) have shown trends that are consistent with the literature[18, 19], especially the location of the peak discomfort. Also, the subjective measures have captured the effect of the seatback on discomfort, where the peak discomfort was shifted from 4–6 Hz in the supported condition (Fig. 4) to 2–3 Hz for the unsupported condition (Fig. 5). The mean discomfort function curves have also shown characteristics similar to the subjective measures in terms of the trends and the location of the peak discomfort (Figs. 4 and 5). For the supported condition (Fig. 4), the proposed objective measure is showing a peak around 4 Hz. The objective discomfort has also captured the shift in the discomfort peak

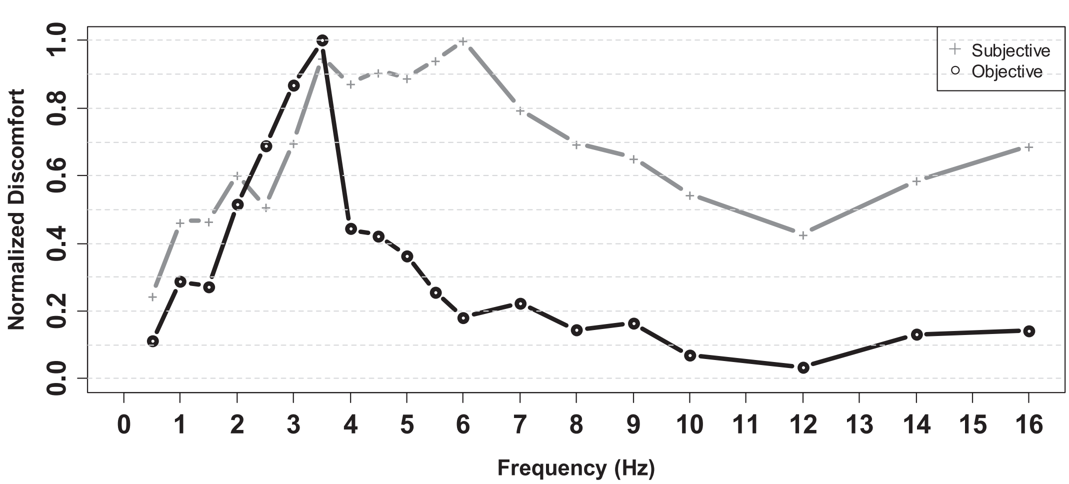

Fig. 4. Mean subjective and frequency-weighted objective discomfort measures (normalized with respect to their maximum) for five subjects in the fore-aft direction in the supported condition with sinusoidal discrete frequencies of 0.5–16 Hz.

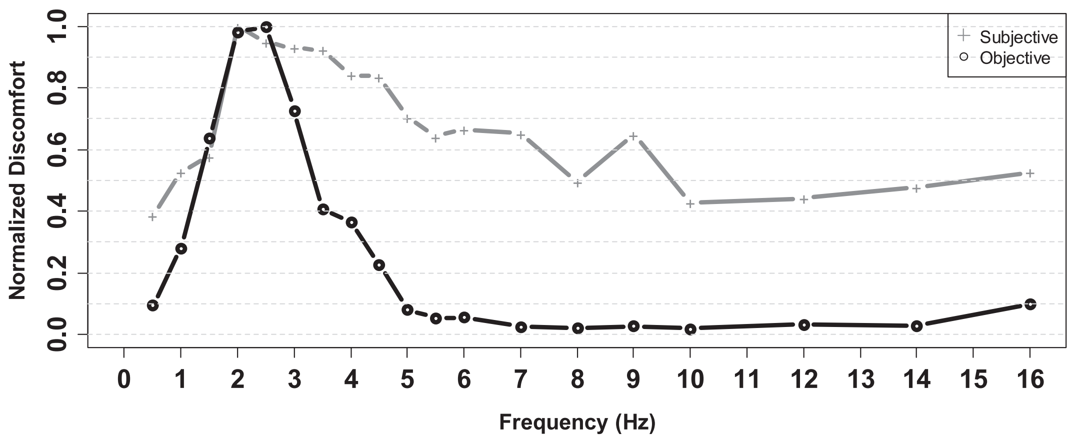

Fig. 5. Mean subjective and frequency-weighted objective discomfort measures (normalized with respect to their maximum) for five subjects in the fore-aft direction in the unsupported condition with sinusoidal discrete frequencies of 0.5–16 Hz.

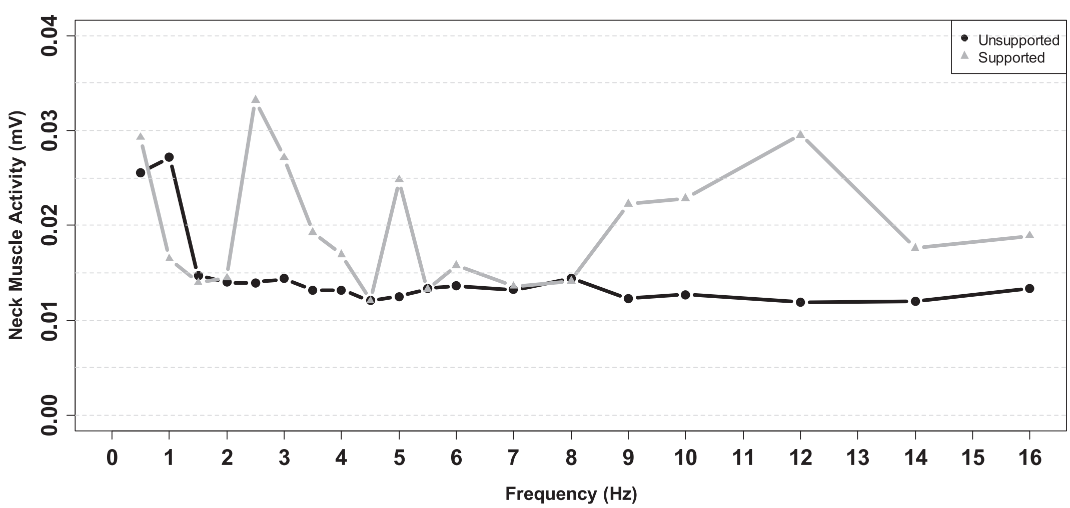

Fig. 6. Mean muscle activity at the neck region (average of left and right) for five subjects in the fore-aft direction in the supported and unsupported back conditions with sinusoidal discrete frequencies of 0.5–16 Hz.

due to the unsupported condition, where the peak happens at approximately 2 Hz (Fig. 5). For the unsupported back condition, the neck and lower back muscles (Figs. 6 and 7) have shown some activities at all frequencies, but they were generally flat. For the supported back condition, similar EMG characteristics have been noticed, but with higher activities around 2–5 Hz and around 12 Hz as shown in Figs. 6 and 7.

Multiple-axis WBV

Figure 8 shows the mean objective discomfort measures derived from the flexion-extension motion of the head-neck during WBV for the three seat configurations (ST, FM, and SM as shown in Fig. 1). The objective discomfort in the flexion-extension direction has shown noticeable peaks around windows 19–22, 34–36, and 42–44 of the ride time, which are fairly consistent with the severity of the input motion shown in Fig. 2. The subjective perceptions were conflicting. In particular, when the subjects were asked about their discomfort and which seat was better, they gave contradicting answers. Muscle activities at the neck and back regions[29] showed that the SM configuration has the least muscle activity, followed by FM and ST, although the SM did show the greatest motion in the head-neck area.

Discussion

An objective discomfort measure is presented in this work. It is based on solving an inverse kinematics problem and uses the calculated joint angles to quantify discomfort in a mathematical form. The objective discomfort characterizes discomfort as the postural deviation from a specified neutral posture and peaks when a joint reaches its limit. The hypothesis is that the peak discomfort occurs at the greatest motion. It has been shown to be realistic and consistent with the concept of dynamic discomfort in the fore-aft direction as demonstrated by Subashi and Griffin[19]. While the motion in multiple-axis WBV is more complicated, the same hypothesis has shown a potential to be extended to multiple-axis WBV.

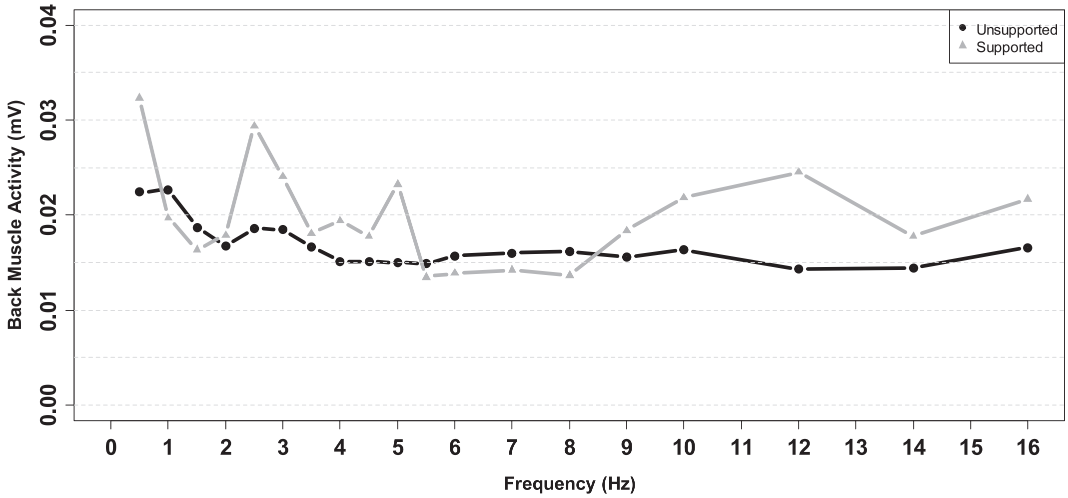

Fig. 7. Mean muscle activity at the lumbar region (average of left and right) for five subjects in the fore-aft direction in the supported and unsupported back conditions with sinusoidal discrete frequencies of 0.5–16 Hz.

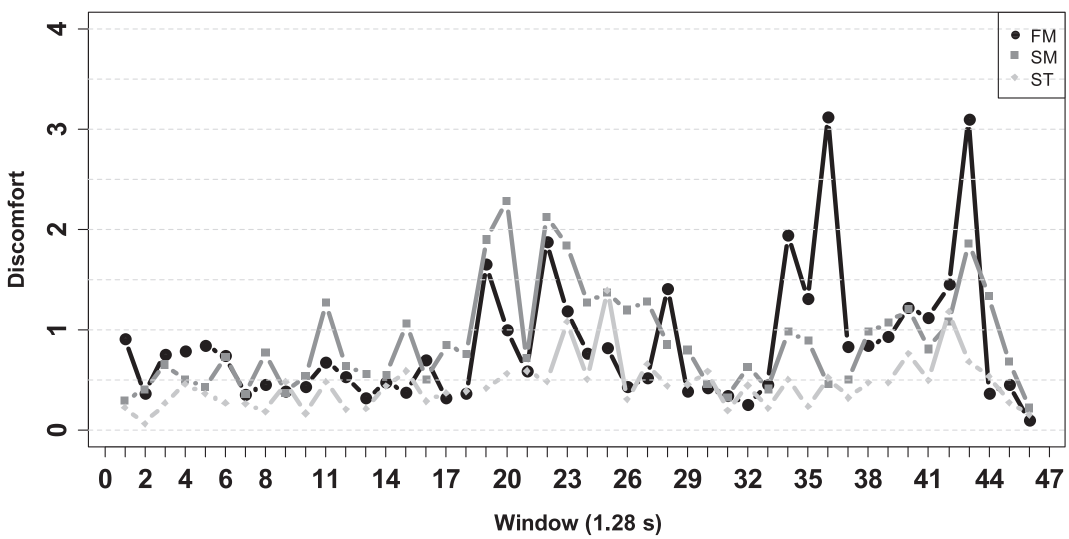

Fig. 8. Mean peak discomfort function curves derived from the flexion-extension motion of the neck joint during the WBV while using the floor-mounted control (FM), seat-mounted control (SM), and steering wheel (ST) seats. Each point in the graph represents the peak objective discomfort in a window of 256 frames (1.28 s).

While the proposed discomfort measure is currently based on the neck joint due to its significant relevancy to discomfort during vibration, especially for the supported back conditions, the behavior of this function could be improved when other joints are considered. The proposed methodology has been tested and has shown significant correlations with the subjective discomfort for the fore-aft direction in terms of the location and the shift of the peak discomfort for the supported and unsupported postures as shown in Figs. 4 and 5. The methodology has been also applied to multiple-axis WBV scenarios, where the initial results have shown reasonable correlations between the discomfort peaks and the WBV input in the time domain, and its ability to capture the effect of the seating conditions.

The contour plot in Fig. 2 can be used to correlate the timing of the peaks in the objective discomfort in Fig. 8 with the magnitude and frequency content of the input signal (Fig. 2) at each relevant time window. It can be seen from Fig. 2 that the frequency components that cause severe motion are a combination of low frequencies (<2 Hz) and are happening at windows 19–24, 32–36, and 41–45 of the motion. The effects of the severe input motion at the aforementioned windows are reflected in Fig. 8, where the objective discomfort measure showed peaks at windows 19–22, 34–36, and 42–44.

While the subjective rating of discomfort for the fore-aft ride files was conducted at a reasonable time-frame of 15 s, the length of the ride file for the multiple-axis WBV (60 s) was too long for the subjects to give accurate assessments of their perceptions. Therefore, when asked about their discomfort levels in multiple-axis WBV, the subjects gave contradicting responses. Shorter ride files will be used in future studies to capture the characteristics of the subjective discomfort.

Nevertheless, the objective discomfort measure (Fig. 8) peaked in a manner consistent with the severity of the input motion in terms of magnitude and frequency contents (Fig. 2).

In general, Fig. 8 showed a higher discomfort with the SM and FM controls in comparison to the ST control. However, some subjects characterized the ST control as the least comfortable. One main reason behind that is related to the incompleteness of the current objective discomfort in capturing the discomfort at other joints. For example, the back motion for the ST was the highest among the seats but was not included in the current objective discomfort.

The frequency range used for the fore-aft study was selected to cover the critical zones, which were based on observation in the literature where the peak in other dynamics measures such as the apparent mass and transmissibility can occur in this range of frequencies. For example, the peak in the apparent mass is around 4 Hz for supported back and around 0.7 Hz if there is no backrest[30, 31].

As shown in Figs. 4 and 5, the peak in the objective discomfort shifted from 4 Hz for the supported posture to 2 Hz for the unsupported posture. These results are consistent with other objective measures presented in the literature, such as the transmissibility and the apparent mass. The shift in the objective-measure peak is well known in the literature and is attributed to the nonlinearity of the human body where the natural frequency of the body is shifted to lower values when more energy enters the body. For the supported posture with armrest, a decent amount of energy will be dissipated through the interaction between the subject and the back and arm supports. Particularly for the latter, the arms may interact with the armrests and work as active absorbers. As a result, less energy enters the body and, therefore, the resonance frequency was shifted to the right to 4 Hz.

It is interesting to notice the characteristics and the peaks in the neck and back muscle activities for the supported condition and how this related to the subjective and objective discomfort. The peak muscle activities were in the range 2–5 Hz and around 12 Hz as depicted in Figs. 6 and 7. These activities coincide with the peaks in the objective and subjective discomfort as shown in Figs. 4 and 5, and it reflects large muscle activities to overcome the uncomfortable motion at these frequency ranges. With the supported back, the subjects mostly lean on the back support and interact with it to minimize their motion; therefore, as Figs. 6 and 7 show, the muscle activations were low at most times[32]. However, peaks in the muscle activation appeared when the motion became large, as subjects felt more secure using their muscles to lessen the motion, and that's what might have happened around 2–5 Hz and around 12 Hz. For the unsupported back, the subjects activated their muscles as the only means to minimize the motion and showed flat continuous activation during the entire time of the ride; therefore, they didn't show more activity around 2–5 Hz and around 12 Hz.

While any biomechanical objective measure for discomfort should contain the motion and the muscle activity, for situations where the muscle activity is low and the relative motion is large, the acute discomfort measure may be based on the motion only. It has been shown that muscle activity at the neck and shoulder for the steering wheel controls (ST) (unsupported posture) are higher than for the seat-mounted controls (SM) (supported); however, the head-neck motions for the SM control were higher than ST[9, 10]. Therefore, the larger head-neck motion and smaller muscle activation in the case of the SM control may cause more discomfort in the short term, and the smaller head-neck motion with a plateau type higher muscle activity for the ST control may cause more discomfort for a longer time due to muscle fatigue.

At higher frequencies, the magnitude of the proposed objective discomfort measure becomes small, which makes it disproportional to the subjective discomfort. This problem was solved in this work by weighing the objective discomfort at each frequency, by multiplying it by the third power of the input frequency. With this weighting process, the objective discomfort was able to capture the characteristics of the subjective discomfort at higher frequencies around 12 Hz where both functions started to go up again as shown in Figs. 4 and 5. Because of the difference in the units and the magnitude of the subjective and objective discomfort and for the sake of comparison, each function was normalized by dividing it by its maximum and was represented in the range 0–1 as shown in Figs. 4 and 5. For multiple-axis WBV with random signals, the proposed frequency-weighted approach could be applied but has to be proven.

The objective of this article was to introduce a methodology to objectively calculate the discomfort for people inside a WBV environment. The methodology was tested in the fore-aft direction and showed significant correlation with the subjective discomfort. The methodology was also applied to assess the discomfort in multiple-axis WBV with three seat-control configurations and showed good correlation with the severity of the input signals. More work is warranted to demonstrate the validity and effectiveness of the proposed method for general multiple-axis WBV applications.

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