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Apparent mass of the standing human body when using a whole-body vibration training machine: Effect of knee angle and input frequency

Naser Nawayseh [a,*], Sadeque Hamdan [b]

[a] Mechanical Engineering Department, College of Engineering, University of Sharjah, PO Box 27272, Sharjah, United Arab Emirates
[b] Sustainable Engineering Asset Management Research Group (SEAM), University of Sharjah, PO Box 27272, Sharjah, United Arab Emirates

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ABSTRACT

Several studies have investigated the transmission of vibration from the vibrating plate of a whole-body vibration training machine (WBVTM) to different locations on the human body. No known work has investigated the interface force between the vibrating plate of the machine and the human body. This paper investigates the effect of bending the knees and the vibration frequency on the interface force (presented as apparent mass (AM)) between the vibrating plate and the body. Twelve male subjects stood with four different knee angles (180, 165, 150 and 135°) and were exposed to sinusoidal vertical vibration at eight frequencies in the range of 17–42 Hz. The vertical acceleration and the interface force between the body and the vibrating plate were measured and used to calculate the AM. The acceleration and force depended on the frequency and were found to vary with both the adopted posture and subject. The AM generally decreased with increasing the frequency but showed a peak at 24 Hz which was clearer when the knees were bent. Bending the knees showed an effect similar to increasing the damping of a system with base excitation; increasing the damping reduced the AM in the resonance region but increased the AM at higher frequencies. Users of WBVTMs have to be careful when choosing the training posture: although, as shown in previous studies, bending the knees reduces the transmission of vibration to the spine, it increases the interface forces which might indicate increased stresses on the lower legs and joints.

1. Introduction

In addition to their use under the supervision of health professionals and sport trainers, whole-body vibration training machines (WBVTMs) have been commercialised for personal use. The wide spread use of WBVTMs has motivated researchers from different disciplines to investigate their effectiveness and safe use (e.g. Muir et al., 2013). Two main research tracks can be identified in this regard. One track is concerned with investigating the benefits of using WBVTMs from medical rehabilitation and sports medicine/training point of view. This research direction focuses on studying the effect of vibration on muscles strength and/or activities (Delecluse et al., 2003; Lienhard et al., 2017; Rees et al., 2008; Verschueren et al., 2004; Yang et al., 2015), bone density (Humphries et al., 2009; Verschueren et al., 2004; Yang et al., 2015) and postural stability (de Ruiter et al., 2003; Niehoff et al., 2012; Pollock et al., 2011; Rees et al., 2008; Schlee et al., 2012; Sonza et al., 2015a; Torvinen et al., 2003; Verschueren et al., 2004; Yang et al., 2015).

The other research direction focuses on the biodynamic responses and threshold/sensitivity to vibration under vibration training conditions (Nawayseh, 2018; Pollock et al., 2011; Schlee et al., 2012; Sonza et al., 2015a, 2013). Biodynamic responses to vibration can be divided into two main responses, namely transmissibility and mechanical impedance/apparent mass (AM). The transmissibility is defined as the ratio between two same-type measures (e.g. acceleration) taken at different locations while the mechanical impedance (or AM) is defined as the ratio between force and velocity (or acceleration) measured at the same point (Griffin, 1990). Both responses are important when building models representing the response of humans to vibration (Matsumoto and Griffin, 2003, 2001; Subashi et al., 2008). The current study falls under the biodynamic responses research direction.

Previous studies focused on the transmission of vibration through the body when using WBVTMs. Vibration transmitted to the upper-body of a standing person has been reported to be less than that transmitted to the lower-body (Cook et al., 2011; Friesenbichler et al., 2014; Kiiski et al., 2008; Pel et al., 2009;

* Corresponding author.
E-mail address: nnawayseh@sharjah.ac.ae (N. Nawayseh).

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Sonza et al., 2015b; Tankisheva et al., 2013; Vasconcellos et al., 2014). Transmission of vibration through the body has also been reported to depend on the posture adopted during the training (Harazin and Grzesik, 1998; Munera et al., 2016; Nawayseh, 2018). For example, standing with bent knee during whole-body vibration (WBV) was found to decrease the vibration at the spine compared to standing with straight legs or locked knees (Matsumoto and Griffin, 1998; Munera et al., 2016; Tankisheva et al., 2013). Unlike the transmissibility, the AM reported in the literature (e.g. Matsumoto and Griffin, 1998; Subashi et al., 2006; Tarabini et al., 2013) is for conditions different from those used with WBVTMs: while previous studies have measured the AM at low-frequency low-magnitude vibrations, whole-body vibration training (WBVT) requires the exposure to combinations of high frequencies and high magnitudes of vibrations. To the best of the authors' knowledge, no previous work has measured the AM under such conditions.

The objective of this study is to investigate the effect of the knee angle and vibration frequency on the AM under WBVT conditions. It was hypothesised that bending the knees will increase the damping of the body and hence will change the AM. It was further hypothesised that the AM will depend on the vibration frequency.

2. Methods

2.1. Apparatus and stimuli

A WBVTM (Model TO-600, The One Fitness Co., Ltd) capable of producing vertical sinusoidal vibration was used in the experiment (Fig. 1). The stimuli consisted of eight vertical sinusoidal signals at frequencies of 17, 20, 24, 28, 31, 35, 38, and 42 Hz. Those are the actual frequencies measured on the vibrating plate when the machine was set at frequencies 20, 25, 30, 35, 40, 45, 50 and 55 Hz, respectively. Similar discrepancies between the actual and set frequencies have been reported previously (Alizadeh-Meghrazi et al., 2014; Bressel et al., 2010; Pel et al., 2009). The actual frequencies are within the range of interest for medical rehabilitation and sports training (Friesenbichler et al., 2014; Marín et al., 2009; Munera et al., 2016; Pollock et al., 2011; Schlee et al., 2012; Sonza et al., 2015a; Vasconcellos et al., 2014). The duration of each stimulus was 5 s followed by 20-second rest to minimise the effect of fatigue. The vibration magnitude increased with increasing the frequency and depended on both the posture and subject as will be shown in the Results section.

The subjects stood on a Vernier FP-BTA force platform mounted on the vibrating plate to measure the vertical force during the exposure to vibration. The force plate has a resonance frequency in the neighbourhood of 250 Hz which is way above the maximum frequency of interest. The vertical acceleration of the vibrating plate was measured using Vernier 3D-BTA accelerometer mounted on the centre line of the plate just in front of the force platform (Fig. 1). The acceleration uniformity over the vibrating plate was confirmed by measuring and comparing the acceleration at five locations on the plate. The acceleration and force signals were acquired at 256 Hz sampling rate via LabQuest 2 data-acquisition system and saved to a laptop.

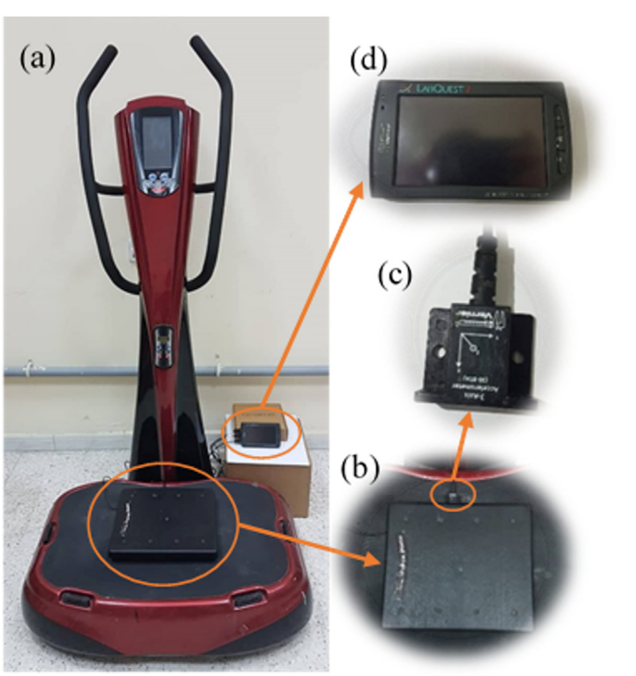

Fig. 1. Experimental setup. (a) Whole-body vibration training machine; (b) force platform; (c) accelerometer; (d) data acquisition system.

2.2. Postures

Four standing postures with different knee angles (135, 150, 165, and 180°) were used to study the effect of knee angle on the AM. The knee angle was fixed at the required angle using a simple goniometer made of two metal rulers and a protractor. With each posture, the subjects stood barefooted having their upper-body upright, folding their arms against their chests and looking straight ahead. The separation between the centre of the feet was fixed at 200 mm.

2.3. Subjects

Twelve healthy male subjects with average age 22.5 years (18–26 years), mass 74.7 kg (58–87 kg), and stature 1.77 m (1.60–1.86 m) participated in the experiment. None of the subjects had previously used a WBVTM.

The objectives, benefits, and procedure of the study were explained to the subjects before starting the experiment. The subjects were informed about their right to stop the experiment any time should they decide not to continue. All subjects signed an informed consent form before starting the experiment. The experiment was approved by the Research Ethics Committee at the University of Sharjah (Approval reference REC/16/05/02/S).

2.4. Procedures

The experimenter helped the subject adopt the correct posture including bending the legs at the required angle. Before starting the vibration, the experimenter adjusted the goniometer to the required angle and placed it on the subject's legs. The subject was then asked to bend the knees until the angle between the upper and lower legs matched the goniometer angle.

Each subject was exposed to a total of 32 conditions (4 postures and 8 input frequencies) randomised among subjects to remove the effect of order. With a 5-second exposure duration, it was not difficult for the subjects to maintain their posture throughout the exposure. Nevertheless, the experimenter watched the subjects carefully throughout each exposure to ensure no change in posture during the exposure to vibration.

2.5. Analysis

Before calculating the AM, mass cancellation was performed to remove the effect of the mass of the top plate of the force platform on which the subject stood. Mass cancellation was performed in the time domain by multiplying the mass of the top plate (1.98 kg) by the measured acceleration and subtracting the result from the measured force to obtain the corrected force. The AM was then calculated as the ratio of the r.m.s. value of the corrected force to the r.m.s value of the acceleration (Eq. (1)).

$$AM(f) = \frac{r.m.s.(F_z)}{r.m.s.(a_z)} \quad (1)$$

where AM is the apparent mass, f is the frequency, F_z, is the corrected force, and a_z is the acceleration measured on the platform.

The phase angle between the corrected force and the acceleration was calculated at each frequency using the dot product between the force and acceleration signals (Eq. (2)).

$$\theta(f) = \cos^{-1}\frac{F_z \cdot a_z}{|F_z||a_z|} \quad (2)$$

where θ is the phase angle between the corrected force (F_z) and the acceleration (a_z).

The coherency between the corrected force and the acceleration was calculated using Eq. (3).

$$\gamma^2(f) = \frac{|SF_z a_z(f)|^2}{SF_z F_z(f) S a_z a_z(f)} \quad (3)$$

where γ^2 is the coherency, $SF_za_z(f)$ is the cross-spectrum between the force (F_z) and acceleration (a_z), and $SF_zF_z(f)$ and $Sa_za_z(f)$ are the auto-spectra of the corrected force (F_z) and acceleration (a_z), respectively. The length of each signal was 1280 samples. Hamming windowing and 50% overlap between the sections of the signals were used in the spectral calculation.

2.6. Statistical analysis

The normality of the data was checked using Shapiro-Wilk test in order to decide whether to use a parametric or non-parametric test to analyse the data. The results are presented as individual and mean data for the AM magnitude and phase. Statistical analysis was performed using IBM SPSS Statistics 24 software. The significance level (p) was taken as 0.05.

3. Results

The data were found normally distributed ($p > 0.05$). Hence, the effect of knee angle on the AM was investigated statistically using the parametric paired-samples t-test.

Table 1 shows the mean and standard deviation of the acceleration and force of the twelve subjects at each frequency and posture. As shown by the standard deviation, the force and, to a lesser extent, acceleration vary with the subject standing on the platform. This inter-subject variability could be attributed to the different physical characteristics of the subjects (mass, stature, build, etc.).

The acceleration and force were found to depend on the knee angle (Fig. 2, Table 1). For the acceleration, the effect was more pronounced between the 180° knee angle and the other three knee angles especially in the frequency range 20–31 Hz. For the force, bending the knees decreased the force at frequencies below 31 Hz and increased the force at frequencies above 31 Hz (Fig. 2, Tables 1 and 2). However, the change in force with changing the knee angle from 150° to 135° was not significant at high frequencies (lower part of Table 2). The effect of bending the knees on the acceleration and force will be reflected in the calculated AM.

Fig. 3 shows high inter-subject variability in the AM. Nevertheless, general trends in the AM can be observed. With all knee angles, the AM was generally greatest at 17 Hz compared to the other frequencies (Fig. 3 for individual data, Figs. 4 and 5 for mean data). With knee angles of 165, 150, and 135°, the AM magnitude tends to have a peak at 24 Hz the value of which decreased with more bending of the knees ($p < 0.05$, except between 150° and 135°, lower part of Table 3). Compared to a knee angle of 180°, bending the knees generally decreased the AM magnitude at frequencies below about 24 Hz and increased the AM magnitude at higher frequencies (Fig. 4). This observation was statistically significant in most comparisons between 180° knee angle and 165°, 150°, and 135° knee angles ($p < 0.05$, the upper part of Table 3). It has also been observed that the change in the AM magnitude diminishes with more bending of the knees (lower part of Table 3). For example, no statistically significant differences were found between the AM magnitudes measured with knee angles of 150° and 135° at any frequency ($p > 0.05$) except at 17 Hz (lower part of Table 3).

The AM phase angle was affected by both the frequency and knee angle. With knee angle of 180°, the phase angle was smallest at 17 Hz and increased with increasing the frequency. With knee angles of 165, 150, and 135°, the phase angle decreased with increasing the frequency from 17 Hz to 20 and 24 Hz but increased with further increase in frequency (Fig. 3 for individual data, Figs. 4 and 6 for mean data). Statistical analysis showed significant differences between the phase angle measured with knee angle of 180° and those measured with knee angles of 165, 150 and 135° at all frequencies ($p < 0.05$) except at 17 and 38 Hz and between 180° and 165° at both 35 and 42 Hz (upper part of Table 3). Between knee angles of 165°, 150°, and 135°, statistical analysis showed a more pronounced knee angle effect on the phase angle in the frequency range 20–35 Hz (lower part of Table 3). The coherency between the acceleration and the force was found close to 1.0 for most conditions indicating a causal relationship between the two signals (Figs. 3 and 4).

Table 1
Mean and standard deviation of the measured (a) acceleration (m/s^2 r.m.s.) and (b) force (N r.m.s.) at each frequency and posture.

Posture	Frequency (Hz)							
	17	20	24	28	31	35	38	42
(a) Acceleration: mean (standard deviation)								
180°	4.93 (0.6)	11.52 (0.79)	12.14 (1.39)	13.06 (0.79)	14.65 (0.54)	15.82 (0.78)	17.53 (0.78)	19.08 (0.76)
165°	6.09 (0.75)	9.84 (1.63)	8.33 (1)	10.53 (1.05)	13.53 (0.61)	16 (0.62)	18.17 (0.67)	19.83 (0.98)
150°	6.23 (0.88)	9.22 (1.64)	8.22 (0.69)	9.72 (0.91)	12.51 (1)	15.08 (0.88)	17.95 (0.84)	20.01 (0.79)
135°	5.58 (1.22)	8.56 (1.15)	7.71 (0.69)	9.51 (0.58)	11.89 (1.15)	14.6 (1.22)	17.46 (0.88)	20.15 (0.89)
(b) Force: mean (standard deviation)								
180°	77.24 (9.9)	120.39 (8.94)	123.12 (12.94)	119.21 (21.77)	110.48 (17.68)	105.03 (22.24)	93.51 (18)	81.19 (12.58)
165°	80.56 (12.47)	94.98 (9.53)	98.86 (14.55)	105.62 (29.2)	117.44 (31.57)	124.13 (27.54)	123.23 (26.25)	112.96 (28.33)
150°	74.17 (10.71)	83.57 (11.24)	90.58 (13.57)	99.47 (21.42)	117.23 (27.78)	140.63 (33.31)	136.79 (30.73)	138.55 (40.77)
135°	72.79 (10.82)	76.12 (11.77)	84.23 (8.78)	91.26 (16.19)	111.26 (26.8)	129.94 (27.74)	139.5 (32.4)	142.63 (27.44)

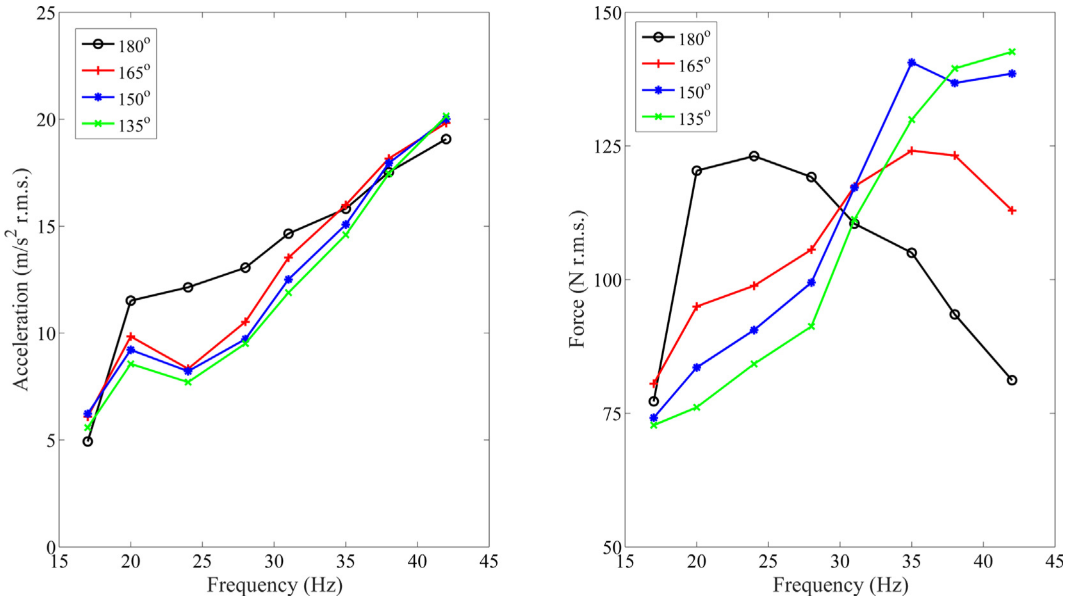

Fig. 2. Mean input acceleration and interface force between the vibrating plate and the feet of 12 subjects. (a) input acceleration; (b) interface force.

Table 2
Paired samples *t*-test for the effect of knee angle on the force at each frequency. Total refers to the total number of significant differences in the force between two knee angles over the whole frequency range.

Comparison pair	Frequency (Hz)								Total (out of 8)
	17	20	24	28	31	35	38	42	
180–165°	0.089	*	*	0.053	0.202	*	*	*	5
180–150°	0.081	*	*	*	0.174	*	*	*	6
180–135°	*	*	*	*	0.462	*	*	*	7
165–150°	*	*	*	*	0.471	*	*	*	7
165–135°	*	*	*	*	0.162	0.220	*	*	6
150–135°	0.077	*	*	*	0.183	0.050	0.390	0.361	3

* $p < 0.05$.

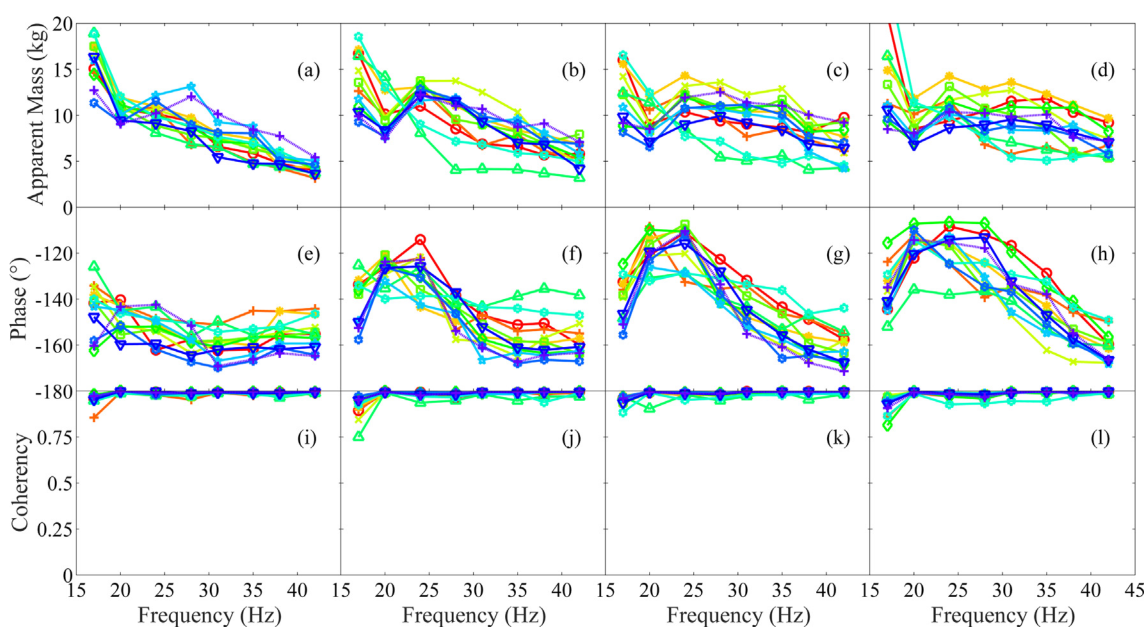

Fig. 3. Individual apparent mass magnitude, phase angle and coherency of 12 subjects measured with four different knee angles at eight frequencies and four postures. (a), (e), and (i): knee angle of 180°; (b), (f), and (j): knee angle of 165°; (c), (g), and (k): knee angle of 150°; (d), (h), and (l): knee angle of 135°. Each line code represents a subject.

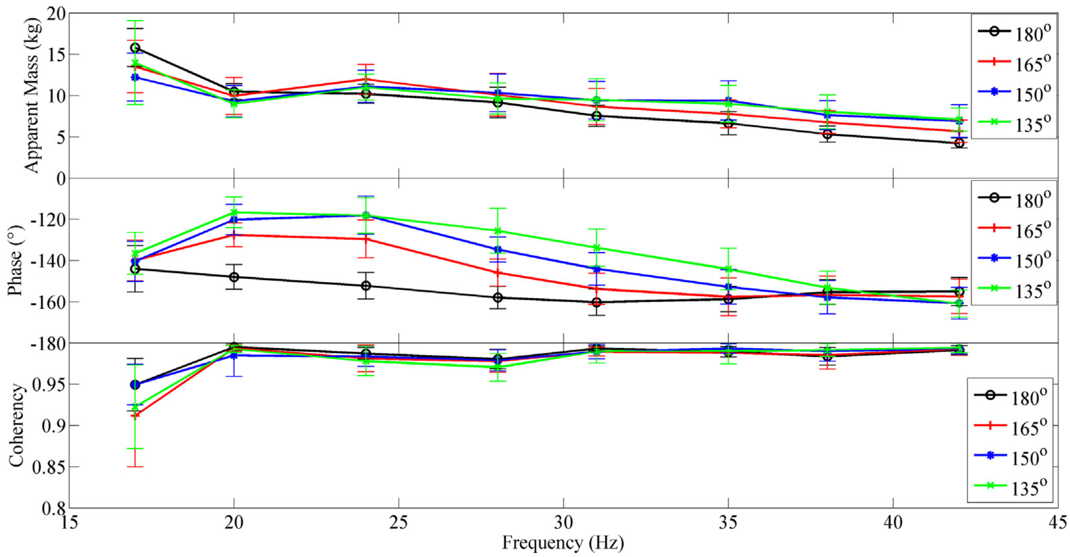

Fig. 4. Mean and standard deviation of the apparent mass magnitude, phase angle and coherency of 12 subjects.

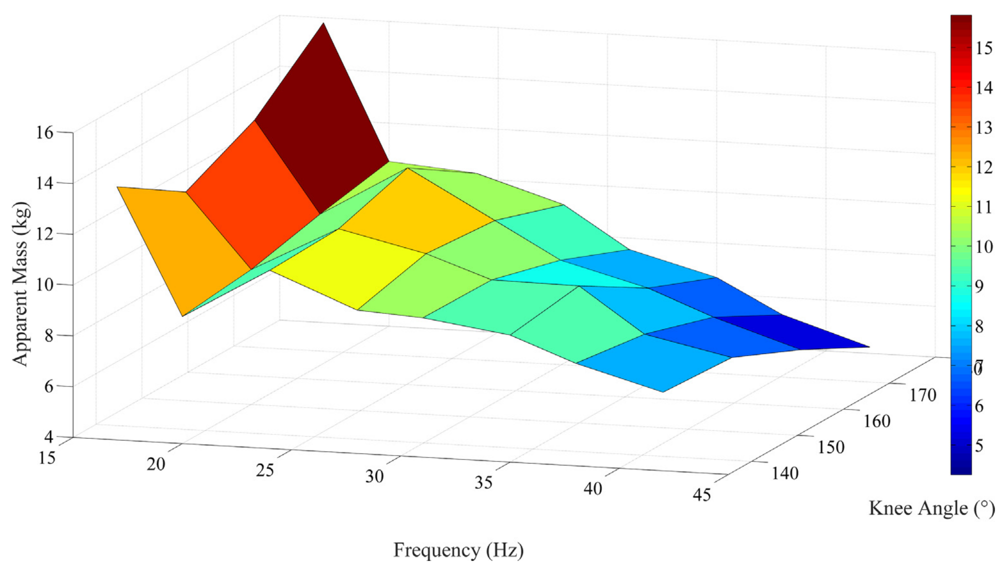

Fig. 5. 3D representation for the effect of knee angle and frequency on the mean apparent mass magnitude of 12 subjects.

Table 3
Paired samples *t*-test for the effect of knee angle on the apparent mass magnitude (AM) and phase (Ph) at each frequency. Total refers to the total number of significant differences between two knee angles for either the magnitude or the phase while overall refers to the overall effect on both the magnitude and phase together over the whole frequency range.

Comparison pair	Frequency (Hz)																Total (out of 8)		Overall (out of 16)
	17		20		24		28		31		35		38		42		AM	Ph	AM + Ph
	AM	Ph	AM	Ph	AM	Ph	AM	Ph	AM	Ph	AM	Ph	AM	Ph	AM	Ph			
180–165°	*	0.059	0.158	*	*	*	0.098	*	*	*	*	0.314	*	0.264	*	0.153	6	4	10
180–150°	*	0.196	*	*	0.056	*	*	*	*	*	*	*	*	0.085	*	*	7	6	13
180–135°	0.072	0.087	*	*	0.083	*	0.22	*	*	*	*	*	*	0.204	*	*	5	6	11
165–150°	*	0.428	*	*	*	*	0.248	*	*	*	*	*	*	0.229	*	*	7	6	13
165–135°	0.277	0.198	*	*	*	*	0.247	*	0.102	*	*	*	*	0.146	*	0.089	5	5	10
150–135°	*	0.066	0.067	*	0.401	0.480	0.075	*	0.462	*	0.236	*	0.259	*	0.377	0.453	1	5	6

* $p < 0.05$

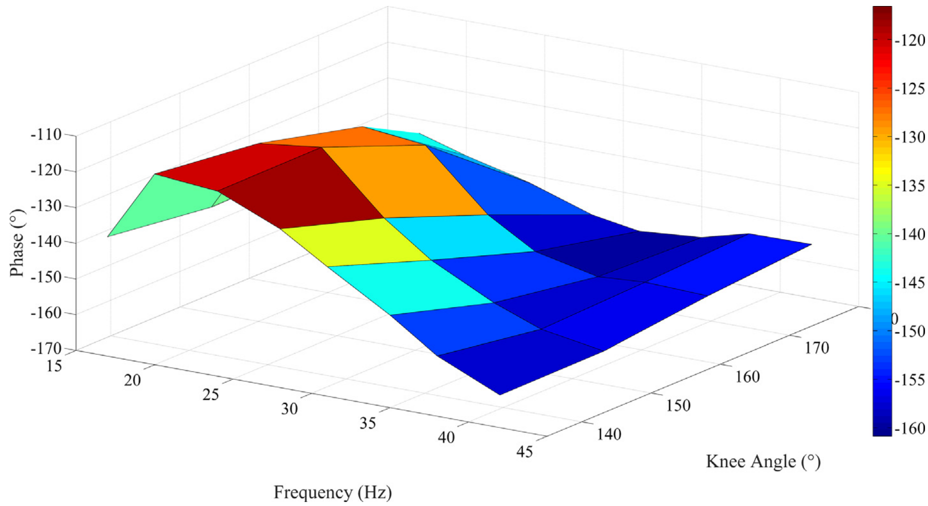

Fig. 6. 3D representation for the effect of knee angle and frequency on the mean apparent mass phase angle of 12 subjects.

4. Discussion

A few previous studies have reported the AM of the standing human body but with vibration conditions different from those used during WBVT (Matsumoto and Griffin, 1998). The main purpose of those studies was to identify and understand the response of the human body to low vibration magnitudes and frequencies. The current work is concerned with measuring the AM at much higher vibration magnitudes and frequencies than those used previously. Given the nonlinear response of the human body and its possible association with muscle tension (Matsumoto and Griffin, 2002), the response measured with high vibration levels is expected to be different from that measured with low vibration levels in the same frequency range. Further research is needed to compare the responses measured with high and low vibration levels in the same frequency range.

The AM magnitude was highest at 17 Hz and decreased with increasing the frequency. This is consistent with the reported decrease in the AM with increasing the frequency above the whole-body resonance frequency (Matsumoto and Griffin, 1998; Subashi et al., 2006; Tarabini et al., 2013). The AM decreases with increasing the frequency because at high frequencies, the upper-body parts are coupled to each other loosely and hence, the measured reaction force is dominated by the soft tissues directly in contact with the vibrating plate leading to a drop in the AM (Griffin, 1990).

The AM peak at around 24 Hz was more pronounced in the bent knee postures. This frequency is within the resonance frequency range (15–30 Hz) of the ankle and lower-extremity muscles (Goggins et al., 2018a; Wakeling et al., 2002). A previous study reported that all subjects felt discomfort when exposed to vertical vibration at frequencies between 20 and 25 Hz (Kiiski et al., 2008) likely due to resonance. The subjects in the current study did not report discomfort at any frequency possibly due to the lower exposure duration and vibration magnitude used here than used by (Kiiski et al., 2008). The accentuation of the 24 Hz peak when the legs were bent as opposed to straight may indicate that this peak is produced by pitching modes in the lower-extremities. The higher pitch motion with the bent knee postures than the straight knee posture could be due to the different static weight distribution over the feet between those postures: while the weight is mostly concentrated over the rearfoot with the straight knee posture, it is uniformly distributed over the forefoot, midfoot and rearfoot with the bent knee postures (Tarabini et al., 2013). The percentage of the static weight on the forefoot also increases with the vibration (Tarabini et al., 2013). Those observations indicate an increase in the fore-and-aft distance between the centre of mass of the body and the ankle joint when bending the knees compared to straight knees and hence more pitch motion. This is also supported by the high anterior-posterior shank acceleration (Cook et al., 2011) and fore-and-aft cross-axis forces at the feet (Subashi et al., 2006) during vertical vibration.

The results showed that bending the knees increased the AM magnitude at high frequencies. A similar increase in the transmissibility to the shank and ankle with bending the knees was reported previously (Cook et al., 2011; Munera et al., 2016; Tankisheva et al., 2013; Yang et al., 2012). Previous studies have also reported a decrease in the transmissibility to locations above the knees when bending the knees (Munera et al., 2016; Tankisheva et al., 2013; Yang et al., 2012) which is opposite to the trend found with the AM. This indicates that measuring the transmissibility to the upper-body is probably not enough to evaluate the risk of the high vibration produced by WBVTMs: while bending the knees can be effective in damping the vibration at sites above the knees, it increases the stresses on the lower parts of the body such as the ankle and foot joints.

The majority of the transmissibility studies concluded that bending the knees increased the damping and reduced the transmission of vibration to locations above the knees. Several possible mechanisms may lead to increasing the damping with bending the knees. Increased muscles stiffness, neuromuscular activation, possible muscle contraction and the natural oscillation of the muscles surrounding the lower limb have been suggested (Heitmann et al., 2012; Munera et al., 2016). Damping performance of the muscles of the legs have been found to increase with increasing muscle activity (Tankisheva et al., 2013; Wakeling et al., 2002). Concentrating the centre of pressure on a certain location of the foot has been shown to increase the peak transmissibility frequency and decrease the peak transmissibility magnitude to that location (Goggins et al., 2018b). This implies an increase in tissue stiffness (e.g. contact stiffness below that location) and/or muscle activities accompanied with increased damping, which is consistent with the above suggestions**.** The dealignment of the body segments during bending the knees leads to activation of muscles to maintain the required posture (Harazin and Grzesik, 1998; Vasconcellos et al., 2014). This encourages changes in the elastic and damping properties of musculoskeletal tissues, leading to a decrease in the acceleration transmitted to the body (Matsumoto and Griffin, 1998;

Paddan and Griffin, 1993). However, with more bending of the knees, the ability to damp the vibration seems to decline due to the changes in joint compliance (Abercromby et al., 2007). This is consistent with the results obtained in the current study where no significant difference was found in the AM magnitude between knee angles 150° and 135°.

With the soft sole tissues in contact with the vibrating plate, the body can be thought of as a mass-spring-damper system with base excitation. The force at the interface between the feet and the vibrating plate represents the force transmitted to the plate due to the reactions from the springs and dampers (which represent the stiffness and damping properties of the soft tissues) (Rao, 2011). With base-excitation, more damping of the system reduces the transmitted force around the resonance region and increases it at higher frequencies (Rao, 2011). This might explain the decrease in the AM at around 24 Hz and the increase in the AM at high frequencies with increasing the damping (i.e. further bending of the knees).

A previous study attempted to estimate the relative AM at 30 Hz from the reciprocal of the acceleration of the platform without measuring the interface force (Abercromby et al., 2007). In this estimation, the authors assumed the interface force as an input force that had a constant peak. They further assumed that the relative AM magnitude varied in direct proportionality with the actual AM. The authors reported that the more the knees were bent, the lower the relative AM at 30 Hz was which is inconsistent with our results. The assumed constant input force by Abercromby et al. (2007) was found to be not constant in this study but to depend on the posture. The change in the interface force with posture should not be surprising: this force represents the reactive force due to the vertical as well as rotational dynamic motions of the body and not an input constant force. The AM obtained in this work from the measured acceleration and interface force indicates clearly that bending the knees increases the AM at high frequencies which contradicts the conclusion of Abercromby et al. (2007).

The results show an effect of bending the knees on the AM magnitude and phase. This adds to the existing knowledge on the effect of bending the knees on the transmissibility to locations below and above the knees. The effect of bending the knees on the AM and transmissibility to locations below the knee is opposite to its effect on the transmissibility to locations above the knees. Those effects indicate the need for a standardized protocol for the use of WBVTMs that should take into consideration not only the magnitude, frequency and duration of vibration but also the adopted posture for safe use of the machine. In the current BS 6841 (British Standards Institution, 1987), for example, the frequency weighting used for vibration evaluation is applicable to erect standing posture and may not be applicable to postures with bent knees (Patelli et al., 2015). Moreover, this frequency weighting filter seems to overestimate the discomfort caused by vibration at high frequency (Patelli et al., 2015). So, from discomfort point of view, this weighting filter is not applicable to WBVT conditions where high frequencies are essential for training.

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Conflict of interest

The authors have no conflicts of interest to disclose.

Research data for this article

The data of the paper can be made available upon request.

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