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TITLE: Transmissibility and waveform purity of whole-body vibrations in older adults

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ABSTRACT

Background: This study examined the transmission power and waveform purity of vertical (synchronous) whole-body vibrations upon its propagation in the human body among older adults.

Methods: Forty community-dwelling older adults participated in the study (33 women; mean age: 60.3 (SD 5.7) years). Four vibration frequencies (25, 30, 35, 40 Hz), two amplitudes (0.6 and 0.9mm), and six different postures were tested. Skin-mounted tri-axial accelerometers were placed at the medial malleolus, tibial tuberosity, greater trochanter, third lumbar vertebra, and forehead. The transmissibility of vibration was computed as the ratio of the root-mean-square-acceleration at different body sites to that of the platform. Signal purity was expressed by the percentage of total transmitted power within 1 Hz of the nominal frequency delivered by the platform.

Findings: Vibration frequency and amplitude were inversely associated with transmissibility in all anatomical landmarks except the medial malleolus. Amplification of signals was noted at the medial malleolus in most testing conditions. The effect of posture on whole-body vibration transmission depends on its frequency and amplitude. In general, toe-standing led to the lowest transmissibility. Single-leg standing had the highest vibration transmission to the hip, while erect standing had the highest transmissibility to the head. The purity of waveform of the vibration signals was well conserved as the vibrations were transmitted from the feet to the upper body.

Interpretation: Whole-body vibration transmissibility was highly influenced by signal frequency, amplitude and posture. These parameters should be carefully considered when prescribing whole-body vibration to older adults.

Keywords: Aging; therapeutic; osteoporosis; exercise; skeletal muscle

1. INTRODUCTION

Whole body vibration (WBV) is gaining increasing interest as a treatment modality in geriatric rehabilitation. WBV is usually delivered to the human body while the individual is standing on the vibration platform. Several studies have reported an increase in lower limb muscle activity during exposure to WBV, likely due to the activation of the tonic vibration reflex (Burke and Schiller, 1976; Lam et al., 2016; Machado et al., 2010). WBV was also found to improve proprioception (Fontana et al., 2005), modulate spinal reflex excitability (Armstrong et al., 2008), and modify motor cortex excitability (Mileva et al., 2009). WBV is also a form of dynamic mechanical loading that is a potent stimulation for osteogenesis (Turner et al., 2011). These proposed mechanisms may explain the improvement in lower limb muscle strength, balance, and bone health in the elderly after WBV intervention (Furness et al., 2010; Lam et al., 2012; Lau et al., 2011; Merriman and Jackson, 2009). However, two important issues have been raised regarding the application of WBV. The first is the lack of consensus on which WBV protocols are optimal for modifying different treatment outcomes (Lam et al., 2012; Lau et al., 2011; Marín et al., 2011; Marín and Rhea, 2010; Merriman and Jackson, 2009). The second issue is the safety of WBV applications (Bochnia et al., 2005; Ishitake et al., 1998; Lam et al., 2012). Established standards for WBV exposure limits, such as the British Standard (BS 6841) and the International Organization for Standardization (ISO 2631), focus on occupational exposure and cannot be fully applied as strict guidelines for WBV training. Investigating WBV transmissibility is thus important in identifying effective treatment protocols (effective transmission to lower limbs and spine), while ensuring safety (minimizing resonance and transmission to head).

WBV transmission is complex, as vibration signal propagation is greatly influenced by nonlinearities in body biomechanics (Kiiski et al., 2008), WBV frequency and amplitude, and postures assumed while on the platform (Rubin et al., 2003). Previous studies have identified peak resonance frequencies <20 Hz (Kiiski et al., 2008; Pollock et al., 2010; Rubin et al., 2003), and decrease in vibration signal transmissibility with increasing frequency (Cook et al., 2011; Kiiski et al., 2008; Pollock et al., 2010; Rubin et al., 2003). However, effects of body posture and WBV amplitude are relatively understudied. Only effects induced by changes in knee angles during squat positions have been examined previously (Abercromby et al., 2007; Avelar et al., 2013; Cook et al., 2011; Muir et al., 2013; Rubin et al., 2003; Tankisheva et al., 2013), and 5 of these 6 studies have small sample sizes ($n \leq 16$) (Abercromby et al., 2007; Cook et al., 2011; Muir et al., 2013; Rubin et al., 2003; Tankisheva et al., 2013). Regarding WBV amplitude, only one study has compared the transmissibility of vertical vibration signals of multiple amplitudes in the legs (Cook et al., 2011). In addition, only one has examined signal purity as vibrations are transmitted up the body during erect standing (Kiiski et al., 2008). This is an important issue, as the degree of signal distortion may directly affect therapeutic efficacy.

Despite the rising interest in the use of WBV in older adults as reflected by the increase in number of WBV clinical trials conducted in this population across different countries (Lam et al., 2012; Orr, 2015), most human studies on WBV transmission reported in the literature were conducted in young adults. During aging, musculoskeletal system changes occur, which may result in undesirable conditions, such as sarcopenia and osteoporosis (Keller and Engelhardt, 2013; McGregor et al., 2014; Zebaze et al., 2010). Since muscle and bone are the major pathways through which the WBV is transmitted, WBV transmissibility could be different between young and old adults.

To address these knowledge gaps, this study investigated the main effects and interactions of WBV frequency, amplitude, and body posture on WBV transmissibility, as well as signal purity during transmission, among older adults. We hypothesized that the transmissibility of the vibration signals would be affected by 1) WBV frequency; 2) WBV amplitude; 3) posture assumed on the vibration platform. We also hypothesized that 4) there would be significant interaction among WBV frequency, amplitude, and postures on WBV transmissibility.

2. METHODS

2.1 Experimental Approach to the Problem

A one-group experimental study with cross-over design was adopted. The transmission of WBV to the medial malleolus, tibial tuberosity, and greater trochanter on the right leg, third lumbar vertebra (L3), and forehead of older adults were measured when they were exposed to WBV of different frequencies and amplitudes while assuming different body postures. Therefore, the dependent variables were the transmissibility and waveform purity of WBV signals at various body parts, while the independent variables were WBV frequency, amplitude, and body posture.

2.2 Subjects

2.2.1 Sampling

Community-dwelling older adults were recruited via advertising in Hong Kong from September 2013 to April 2014. Inclusion criteria were 1) aged ≥50 years, 2) medically stable, 3) able to stand for at least 1 minute with minimal hand support, and 4) able to understand simple verbal commands. Exclusion criteria were: 1) any neurological conditions (e.g., stroke), 2) significant musculoskeletal conditions (e.g., amputation), 3) metal implants in the leg, 4) previous leg fracture, 5) osteoporosis, 6) vestibular disorders, 7) peripheral vascular disease, and 8) other serious illnesses or contraindications to exercise.

2.2.2 Sample size estimation

Studies that compared transmissibility among different WBV frequencies in younger adults yielded large effect sizes (Cohen's d = 1.6–2.4) (Rubin et al., 2003). An analysis of variance (ANOVA) indicated that 34 participants were needed to detect differences with an effect size f = 0.35, α of 0.05, and power of 0.8.

2.2.3 Ethical approval

This study conforms to the ethical principles of the World Medical Association Declaration of Helsinki — Ethical Principles for Medical Research Involving Human Subjects. Ethical approval of the study was granted by the Human Research Ethics Subcommittee of the university. Written informed consent was obtained from each participant.

2.2.4 Demographic characteristics

Forty community-dwelling older adults were enrolled (33 women; mean age: 60.3 (SD=5.7) years). Demographic data are shown in Table 1. The age and body mass index (BMI) of men and women showed no significant difference (*P*≥0.557).

Table 1. Demographic data of participants

	Mean (SD)			
	Male (n=7)	Female (n=33)	All (n=40)	p-value
Age (years)	59.9 (8.4)	60.1 (5.6)	60.3 (5.7)	0.728
Body mass index (kg/m^2)	22.9 (3.8)	23.7 (3.2)	23.6 (3.3)	0.577

2.3 Procedures

2.3.1 Testing conditions

All participants attended a single session of experiment. A vibration platform (Fitvibe Medical, GymnaUniphy NV, Bilzen, Belgium) that generated vibration frequencies of 25Hz, 30Hz, 35Hz, and 40 Hz and amplitudes of approximately 0.6mm and 0.9mm was used for testing. As vibration frequency increased, the protocol yielded platform peak acceleration of 1.70 units of gravitational constant (G=9.81ms^{-2}), 2.25G, 2.90G, and 3.65G if an amplitude of 0.6mm was used, and 2.50G, 3.40G, 4.35G, and 5.50G if an amplitude of 0.9mm was used.

During WBV exposure, participants assumed six different postures: (1) erect standing (knee flexion, 20°), (2) semi-squat (knee flexion, 45°), deep squat (knee flexion, 70°), (4) toe-standing, (5) forward lunge, (6) single-leg standing (right leg knee flexion, 20°) (Fig. 1). An electronic goniometer monitored participants' knee angle (Twin Axis Goniometer SG150; Biometrics Ltd, Newport, UK). The selected postures were commonly used in previous studies (Lam et al., 2012; Lau et al., 2011). During testing, participants held onto the rail lightly for safety and were asked not to put weight on it unless they lost balance. A new trial would be done if the subject lost balance during testing. In postures with bilateral stance, feet were placed shoulder-width apart. Participants stood barefoot on the platform to avoid external damping. The body posture, frequency, and amplitude combinations yielded 48 conditions. For each condition, WBV was sustained for 10s. To minimize potential order-effect bias, the testing conditions were randomly sequenced. Participants rested intermittently to minimize fatigue.

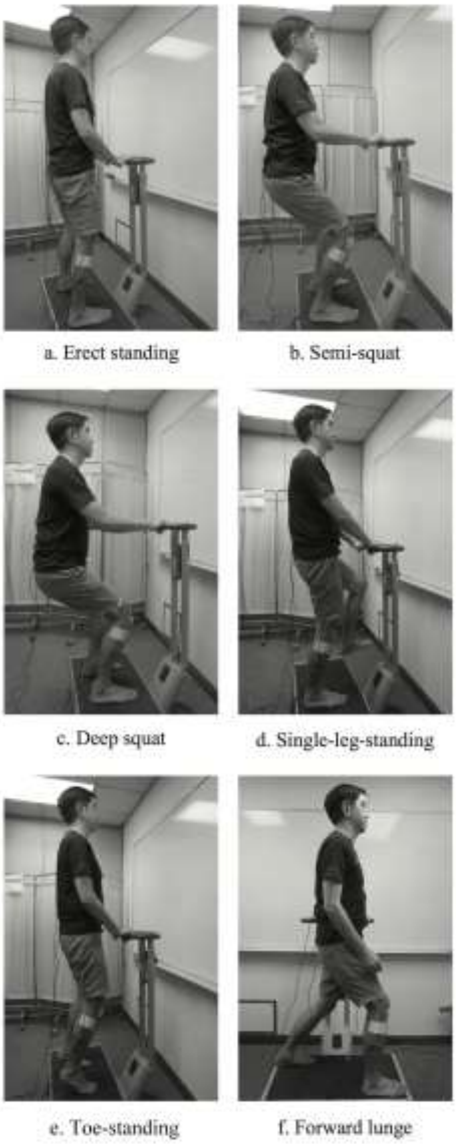

Fig. 1 Demonstration of the six postures

2.3.2 Measurement of acceleration

After calibration, five tri-axial accelerometers (Dytran 7523A5; Dytran Instruments, Inc., CA) were attached to the vibration platform with double-sided tape to measure accelerations at the platform level, for 10 seconds for each of the 8 frequency and amplitude combinations. A total of five trials were performed and the average was used as the platform acceleration. The platform acceleration was measured in both unloaded and loaded conditions. In the former condition, nobody was standing on the platform. In the latter condition, a 50kg person was standing on the platform in a single leg standing posture (the most unstable posture in the testing protocol). The platform acceleration ratios in the unloaded and loaded conditions were similar (ranging from 1.01 to 1.04) across the eight frequency × amplitude combinations. We had also checked the accelerations generated by the platform regularly during the course of the study, and found that the output was consistent over time.

During the experiment, tri-axial accelerometers were attached to the medial malleolus, tibial tuberosity, and greater trochanter on the right leg, third lumbar vertebra (L3), and forehead. The body sites were selected to allow the measurement of acceleration at common target areas in WBV therapy (lower limb and spine), and also in the head where safety consideration was particularly important. These body sites were also commonly used in previous studies conducted in younger individuals (Kiiski et al., 2008; Tankisheva et al., 2013). The accelerometers were fixed by double-sided tape. Omnifix dressing retention tape and Coban self-adherent wrap were used to minimize skin translation. Acceleration data were acquired through an AD-converter using a sampling frequency of 1000 Hz. The platform acceleration was not measured in conjunction with the acceleration at the body site because the interface used in this study contained only 15 channels and could only acquire data from 5 triaxial accelerometers for the 5 body sites simultaneously.

2.3.3 Data Processing

The data obtained in the middle 3s of the 10s recording period for each condition (i.e., 75-200 vibration cycles depending on WBV frequency) was used in analysis, to eliminate the data captured during the accelerating and decelerating phases of the platform. Resultant calibrated accelerations were obtained from the three measurement axes of accelerometers to represent vibration signals for each testing condition. For measurements of accelerations at the platform level, the data acquired from the five trials were averaged to provide a reference platform vibration signal for each vibration condition (Kiiski et al., 2008). Transmissibility at each body site was computed as the ratio of the root-mean-square resultant acceleration (rms-acceleration) of the vibration signals measured at the body site to the rms-acceleration of vibration signals measured at the platform with acceleration from gravity eliminated (Kiiski et al., 2008). To evaluate the signal purity, Fast-Fourier transform analysis with a Hamming window was used to analyze the power spectra of site-specific vibration signals separately for each axis (Kiiski et al., 2008). The proportion of signal power within 1 Hz of the nominal frequency (e.g. signal between 29Hz and 31Hz for a nominal frequency of 30Hz) was computed to assess the degree of sinusoidal waveform distortion as vibrations were transmitted to different body sites (Kiiski et al., 2008). This proportion was calculated as: 100% × [(power of signal within 1 Hz of *x*-axis nominal frequency + power of signal within 1 Hz of *y*-axis nominal frequency + power of signal within 1 Hz of *z*-axis nominal frequency) / (total *x*-axis power + total *y*-axis power + total *z*-axis power)]. The greater the nominal frequency percentage power, the lesser the signal distortion.

2.4 Statistical Analysis

Statistical analysis was performed with SPSS (version 20.0, Armonk, NY). The demographic variables (age, BMI) were compared between men and women using Mann-Whitney U Test because of the small number of men (n=7) in our sample. Histogram was used to examine the normality of the variables. Given that the histograms showed a largely normally distributed sample without outliers, and that skewed distributions have very little effect on the level of significance and power (Glass & Hopkins, 1996), a three-way, repeated-measures ANOVA was conducted to examine the influence of amplitude, frequency, and posture on transmissibility at each site. Contrast analysis using the Bonferroni paired *t*-test was performed for any overall significant results. Among the 9,000 mean transmissibility values obtained (2 amplitudes × 4 frequencies × 6 postures × 5 body sites × 40 subjects), 99 of them were missing due to technical error. The missing data were not specific to any subjects or testing conditions. Significance was set at $P<0.05$. To examine the association of age and body composition with transmissibility, Pearson's correlation coefficient was used. A more stringent significance level was set for the correlation analysis ($P<0.01$), because of the many correlation coefficients generated.

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3. RESULTS

The transmissibility across frequencies, amplitudes, and postures ranged between 0.93–2.55, 0.14–1.22, 0.05–0.77, 0.05–0.41, and 0.07–0.50 at the medial malleolus, tibial tuberosity, greater trochanter, L3, and forehead respectively (Table 2). No adverse effects were reported during the experiments.

3.1 Effect of vibration frequency

A significant main effect of WBV frequency at all anatomical sites (F=72.787–164.190, $P<0.001$, partial η^2=0.651–0.808). Increased WBV frequency significantly increased transmissibility at the medial malleolus, with the opposite trend at the tibial tuberosity, greater trochanter, L3, and forehead, where increased WBV frequency significantly decreased transmissibility (Fig. 2 and 3). Post-hoc analysis revealed significant differences in transmissibility for all comparisons across different frequencies at all sites ($P \leq 0.008$), except between 35 Hz and 40 Hz at the tibial tuberosity ($P=0.084$).

Table 2. Transmissibility in all testing conditions*

	Erect standing		Semi-squat		Deep squat		Forward lunge		Toe-standing		Single-leg Standing	
Amplitude	Low	High	Low	High	Low	High	Low	High	Low	High	Low	High
Medial malleolus												
25Hz	1.78 (0.34)	1.88 (0.47)	1.68 (0.29)	1.77 (0.34)	1.58 (0.32)	1.71 (0.43)	1.28 (0.36)	1.66 (0.32)	1.06 (0.23)	0.96 (0.28)	1.47 (0.26)	1.62 (0.31)
30Hz	2.06 (0.45)	2.20 (0.57)	1.86 (0.35)	2.01 (0.42)	1.74 (0.37)	1.86 (0.41)	1.65 (0.29)	1.90 (0.32)	1.22 (0.31)	1.15 (0.36)	1.77 (0.32)	1.91 (0.39)
35Hz	2.39 (0.63)	2.40 (0.55)	2.19 (0.56)	2.26 (0.51)	1.94 (0.59)	2.12 (0.68)	2.01 (0.41)	2.06 (0.41)	1.25 (0.33)	1.12 (0.38)	2.12 (0.50)	2.19 (0.46)
40Hz	2.55 (0.58)	2.34 (0.74)	2.42 (0.55)	2.50 (0.77)	2.11 (0.62)	2.32 (0.71)	2.10 (0.47)	2.04 (0.55)	1.09 (0.32)	0.93 (0.27)	2.33 (0.47)	2.47 (0.50)
Tibial tuberosity												
25Hz	0.94 (0.30)	0.96 (0.36)	0.87 (0.29)	0.85 (0.33)	0.85 (0.25)	0.86 (0.33)	0.86 (0.33)	0.98 (0.36)	0.25 (0.08)	0.24 (0.07)	1.22 (0.51)	1.19 (0.34)
30Hz	0.77 (0.30)	0.80 (0.33)	0.64 (0.21)	0.62 (0.24)	0.63 (0.23)	0.62 (0.23)	0.77 (0.32)	0.79 (0.32)	0.25 (0.13)	0.23 (0.08)	0.85 (0.29)	0.82 (0.30)
35Hz	0.77 (0.31)	0.77 (0.26)	0.54 (0.20)	0.56 (0.26)	0.51 (0.15)	0.49 (0.17)	0.72 (0.31)	0.69 (0.34)	0.21 (0.08)	0.19 (0.07)	0.70 (0.23)	0.74 (0.25)
40Hz	0.80 (0.30)	0.74 (0.30)	0.53 (0.21)	0.54 (0.25)	0.45 (0.18)	0.48 (0.19)	0.68 (0.31)	0.67 (0.28)	0.17 (0.06)	0.14 (0.06)	0.69 (0.22)	0.72 (0.22)
Greater trochanter												
25Hz	0.40 (0.17)	0.37 (0.21)	0.26 (0.19)	0.22 (0.16)	0.27 (0.14)	0.23 (0.12)	0.26 (0.11)	0.23 (0.12)	0.18 (0.07)	0.15 (0.07)	0.77 (0.33)	0.67 (0.29)
30Hz	0.28 (0.14)	0.26 (0.15)	0.17 (0.12)	0.15 (0.08)	0.22 (0.23)	0.16 (0.09)	0.21 (0.15)	0.19 (0.15)	0.12 (0.06)	0.11 (0.04)	0.54 (0.28)	0.45 (0.21)
35Hz	0.21 (0.17)	0.20 (0.13)	0.12 (0.09)	0.11 (0.07)	0.16 (0.10)	0.13 (0.08)	0.16 (0.11)	0.14 (0.11)	0.08 (0.04)	0.08 (0.04)	0.36 (0.19)	0.32 (0.17)
40Hz	0.19 (0.15)	0.16 (0.12)	0.11 (0.10)	0.10 (0.10)	0.11 (0.06)	0.10 (0.06)	0.13 (0.10)	0.12 (0.09)	0.06 (0.03)	0.05 (0.02)	0.31 (0.17)	0.24 (0.11)
L3												
25Hz	0.38 (0.12)	0.41 (0.18)	0.28 (0.08)	0.24 (0.09)	0.23 (0.08)	0.19 (0.06)	0.30 (0.09)	0.29 (0.12)	0.13 (0.05)	0.11 (0.04)	0.33 (0.11)	0.31 (0.10)
30Hz	0.31 (0.13)	0.32 (0.14)	0.21 (0.08)	0.20 (0.08)	0.18 (0.06)	0.15 (0.04)	0.25 (0.10)	0.23 (0.11)	0.10 (0.04)	0.09 (0.03)	0.26 (0.10)	0.25 (0.12)
35Hz	0.27 (0.13)	0.29 (0.13)	0.18 (0.07)	0.17 (0.08)	0.14 (0.05)	0.12 (0.04)	0.22 (0.10)	0.20 (0.11)	0.08 (0.03)	0.07 (0.02)	0.22 (0.10)	0.23 (0.12)
40Hz	0.25 (0.13)	0.25 (0.13)	0.17 (0.07)	0.15 (0.07)	0.12 (0.04)	0.10 (0.03)	0.19 (0.09)	0.16 (0.08)	0.06 (0.02)	0.05 (0.02)	0.18 (0.08)	0.18 (0.10)
Forehead												
25Hz	0.50 (0.25)	0.45 (0.02)	0.24 (0.12)	0.19 (0.10)	0.22 (0.08)	0.18 (0.08)	0.27 (0.13)	0.24 (0.10)	0.20 (0.08)	0.18 (0.08)	0.30 (0.12)	0.27 (0.11)
30Hz	0.36 (0.14)	0.34 (0.15)	0.18 (0.10)	0.12 (0.06)	0.16 (0.07)	0.14 (0.06)	0.22 (0.11)	0.19 (0.08)	0.16 (0.07)	0.13 (0.04)	0.23 (0.09)	0.20 (0.10)
35Hz	0.28 (0.10)	0.27 (0.11)	0.13 (0.05)	0.10 (0.05)	0.13 (0.04)	0.11 (0.04)	0.16 (0.08)	0.13 (0.06)	0.11 (0.04)	0.10 (0.04)	0.17 (0.06)	0.16 (0.06)
40Hz	0.24 (0.10)	0.19 (0.09)	0.11 (0.05)	0.10 (0.04)	0.11 (0.04)	0.10 (0.04)	0.12 (0.04)	0.11 (0.06)	0.08 (0.03)	0.07 (0.03)	0.14 (0.05)	0.11 (0.05)

* Values are presented as mean (SD).

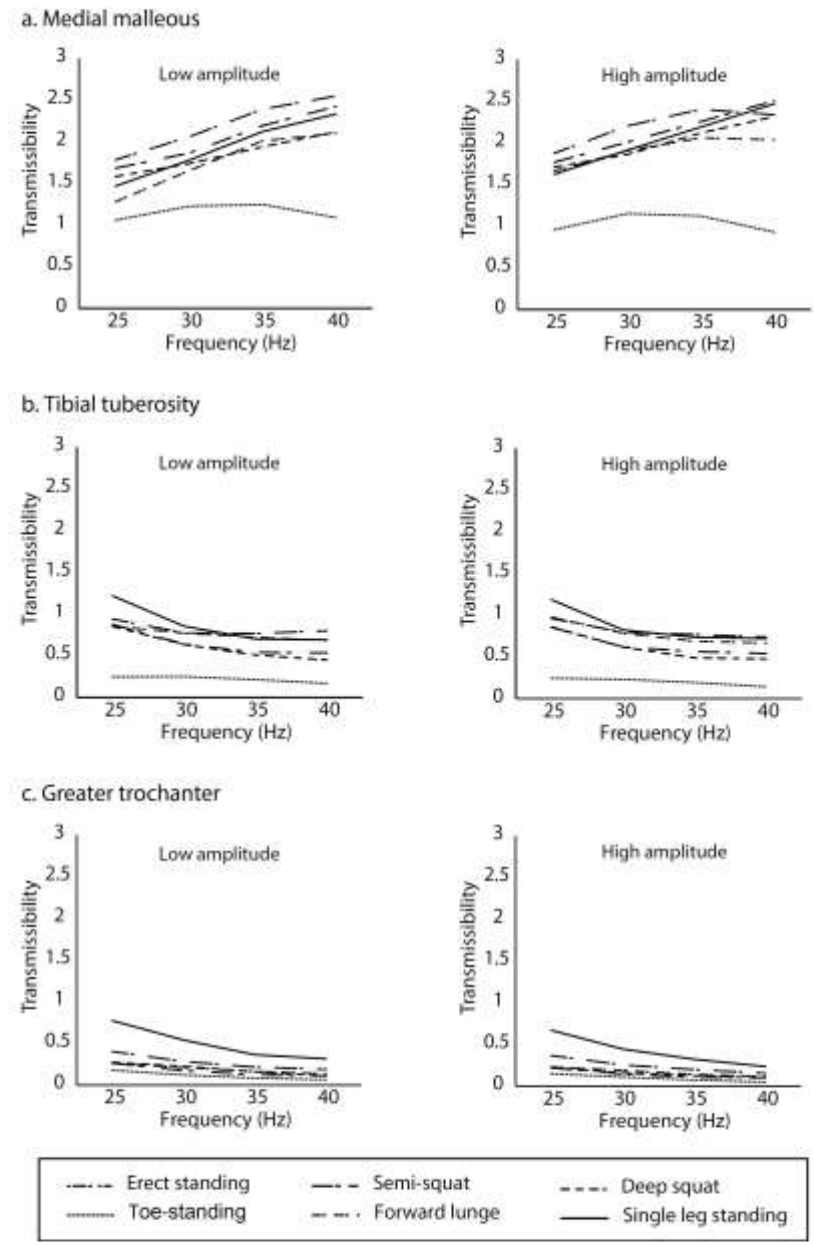

Fig. 2 Transmissibility of whole-body vibration signals at the medial malleolus, tibial tuberosity, and greater trochanter

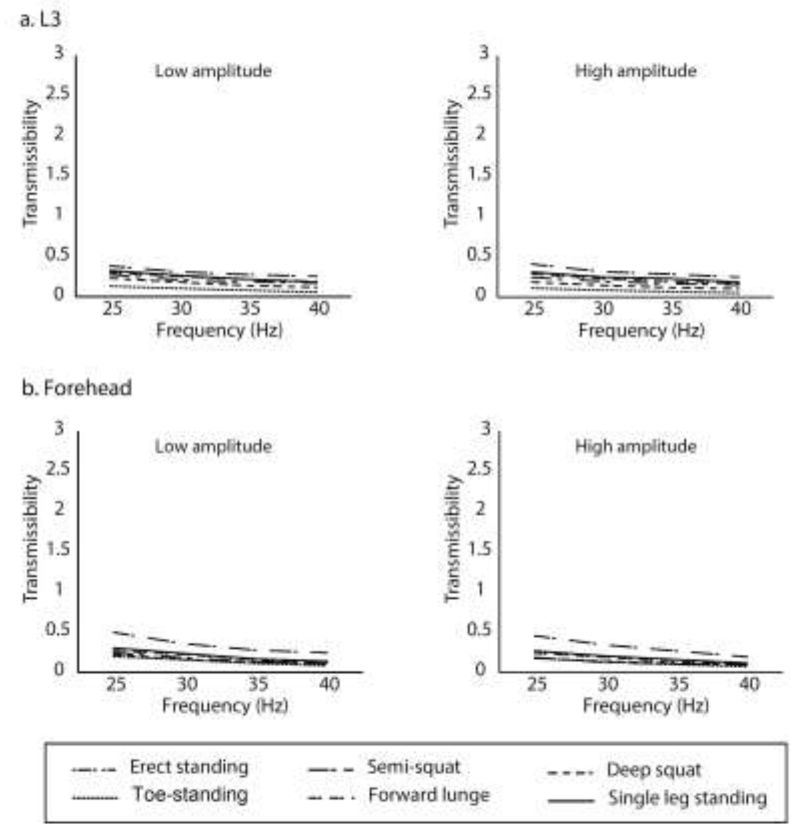

Fig. 3 Transmissibility of whole-body vibration signals at the third lumbar vertebra and the forehead

3.2 Effect of vibration amplitude

The 0.9mm-amplitude yielded significantly greater transmissibility at the medial malleolus ($F=14.561$, $P<0.001$, partial $\eta^2=0.272$) than did the 0.6mm-amplitude (Fig. 2). The effect of amplitude was not significant at the tibial tuberosity ($F<0.001$, $p=0.991$, partial $\eta^2<0.001$). In the greater trochanter, L3, and forehead, the 0.6mm-amplitude resulted in significantly greater transmissibility ($F=41.600$–54.283, $P\leq0.011$, partial $\eta^2=0.516$–0.582) (Fig. 2 and 3).

3.3 Effect of body posture

A significant effect of posture was observed at all sites ($F=66.805$–109.146, $P<0.001$, partial $\eta^2=0.631$–0.737). Toe-standing yielded the lowest transmissibility at all anatomical sites ($P\leq0.02$) (Fig. 2 and 3). The other postures resulted in similar transmissibility at the medial malleolus and tibial tuberosity. At the greater trochanter, single-leg standing yielded the highest transmissibility, followed by erect standing. Both postures resulted in significantly greater transmissibility than did other postures ($P<0.001$) (Fig. 2). The trend was reversed at the L3 and forehead, where erect standing had greater transmissibility than all other postures ($P<0.001$). Single-leg standing was second, with greater transmissibility than all remaining postures ($P<0.001$) except forward lunge ($P\geq0.131$) (Fig. 3).

3.4 Interaction effects

The effect of frequency on transmissibility was found to be dependent on posture in all anatomical positions ($P<0.001$). The effect of amplitude was also dependent on posture at the medial malleolus ($P<0.001$), greater trochanter ($P<0.001$), and L3 ($P=0.005$), but not tibial tuberosity ($P=0.746$) and forehead ($P=0.404$). The amplitude × frequency interaction ($P=0.001$) and amplitude × frequency × posture interaction ($P<0.001$) was only significant at the medial malleolus.

3.5 Frequency domain analysis

The nominal percentage power ranged from 93.4% to 96.2% at the platform (Table 3). The average nominal percentage power across amplitudes, frequencies, and postures at the medial malleolus, tibial tuberosity, greater trochanter, L3, and forehead were 90.5%, 83.6%, 78.7%, 81.7%, and 77.8% respectively. Change in frequency did not affect signal purity across different anatomical sites. The 0.6mm-amplitude yielded slightly better signal purity than did the 0.9mm amplitude at the medial malleolus (difference: 2.9%) and tibial tuberosity (5.2%). This trend was reversed at the greater trochanter (-4.5%), L3 (-3.8%), and forehead (-6.8%). The maximal difference in nominal frequency percentage power among different postures was relatively small at the medial malleolus (5.5%) and tibial tuberosity (5.1%) and larger at greater trochanter (12.9%), L3 (24.5%) and forehead (20.6%). Erect and single-leg standing position best retained the original sinusoidal waveform of signals across different anatomical sites (≥83.0%). Deep squats (70.5–79.1%) and toe-standing (63.7–73.7%) resulted in the lowest signal purity at the greater trochanter, L3 and forehead.

3.6 Association of age and BMI with transmissibility

Of the possible 240 correlation coefficients generated (4 frequencies × 2 amplitudes × 6 postures × 5 body sites), only 3 and 7 were significant for age and BMI respectively (P<0.01). These significant correlations were not specific to any WBV parameters or body sites.

Table 3. Percentage of nominal signal power (signal purity) in all testing conditions*

	Erect standing		**Semi-squat**		**Deep squat**		**Forward lunge**		**Toe-standing**		**Single-leg Standing**	
Amplitude	Low	High	Low	High	Low	High	Low	High	Low	High	Low	High
Platform												
25Hz	95.5 (0.1)	95.8 (0.0)	95.5 (0.1)	95.8 (0.0)	95.5 (0.1)	95.8 (0.0)	95.5 (0.1)	95.8 (0.0)	95.5 (0.1)	95.8 (0.0)	95.5 (0.1)	95.8 (0.0)
30Hz	96.2 (0.0)	95.9 (0.0)	96.2 (0.0)	95.9 (0.0)	96.2 (0.0)	95.9 (0.0)	96.2 (0.0)	95.9 (0.0)	96.2 (0.0)	95.9 (0.0)	96.2 (0.0)	95.9 (0.0)
35Hz	93.4 (0.0)	95.7 (0.0)	93.4 (0.0)	95.7 (0.0)	93.4 (0.0)	95.7 (0.0)	93.4 (0.0)	95.7 (0.0)	93.4 (0.0)	95.7 (0.0)	93.4 (0.0)	95.7 (0.0)
40Hz	93.4 (0.0)	96.0 (0.0)	93.4 (0.0)	96.0 (0.0)	93.4 (0.0)	96.0 (0.0)	93.4 (0.0)	96.0 (0.0)	93.4 (0.0)	96.0 (0.0)	93.4 (0.0)	96.0 (0.0)
Medial malleolus												
25Hz	89.9 (5.6)	83.3 (12.3)	91.0 (4.9)	86.4 (8.1)	89.9 (5.2)	81.5 (13.3)	91.3 (5.4)	88.3 (6.2)	94.7 (2.3)	94.3 (5.0)	91.5 (4.0)	88.0 (6.2)
30Hz	93.1 (2.7)	89.1 (6.3)	92.8 (2.2)	90.5 (6.6)	91.9 (3.0)	89.3 (5.5)	92.5 (4.1)	89.1 (7.4)	95.3 (1.0)	94.3 (1.8)	93.1 (3.0)	92.2 (3.7)
35Hz	91.5 (4.2)	87.9 (5.7)	92.1 (4.0)	89.7 (5.7)	91.4 (2.8)	87.5 (7.4)	90.8 (4.8)	87.9 (7.6)	92.3 (1.2)	94.1 (1.7)	92.8 (3.5)	92.1 (3.8)
40Hz	90.4 (5.0)	85.7 (7.9)	91.2 (3.9)	87.4 (7.2)	90.8 (3.4)	85.9 (6.9)	90.8 (3.7)	86.6 (6.7)	92.4 (1.0)	94.7 (1.4)	92.2 (3.1)	90.5 (4.4)
Tibial tuberosity												
25Hz	84.4 (7.4)	75.4 (11.8)	84.8 (7.0)	75.2 (10.9)	86.7 (10.9)	78.1 (10.6)	86.3 (7.5)	75.6 (11.9)	82.5 (6.9)	87.4 (7.7)	84.7 (9.1)	76.2 (15.2)
30Hz	87.3 (5.6)	80.6 (10.4)	85.1 (6.3)	78.2 (12.7)	86.6 (7.6)	79.4 (9.8)	87.2 (4.4)	80.6 (8.9)	87.3 (6.2)	90.6 (3.8)	84.1 (7.4)	80.2 (10.5)
35Hz	88.9 (2.9)	80.6 (8.0)	86.6 (5.5)	80.6 (8.3)	86.2 (5.3)	80.1 (9.4)	85.7 (6.5)	79.9 (9.6)	86.0 (6.5)	90.1 (6.5)	86.7 (6.4)	81.5 (13.0)
40Hz	87.0 (5.1)	80.2 (6.8)	87.1 (4.5)	79.7 (8.1)	85.6 (10.4)	79.3 (7.7)	86.6 (5.6)	80.9 (6.6)	85.5 (8.3)	89.0 (8.7)	88.6 (5.0)	84.5 (7.1)
Greater trochanter												
25Hz	84.7 (4.6)	81.5 (8.0)	79.2 (4.5)	83.9 (9.4)	72.8 (4.8)	81.9 (10.4)	79.2 (4.8)	83.3 (8.8)	84.0 (1.2)	81.0 (10.0)	83.2 (3.6)	77.8 (11.5)
30Hz	84.9 (2.3)	81.9 (10.8)	77.4 (2.0)	81.2 (12.2)	71.3 (2.7)	81.7 (12.2)	77.1 (4.1)	83.2 (9.4)	67.2 (0.9)	77.9 (10.2)	84.5 (2.0)	82.6 (9.1)
35Hz	82.8 (3.6)	81.9 (11.0)	74.8 (3.2)	80.8 (11.7)	71.9 (2.7)	78.6 (15.8)	76.0 (4.2)	81.8 (10.0)	61.6 (0.4)	73.5 (14.7)	85.4 (1.2)	84.6 (6.6)
40Hz	83.5 (4.1)	82.6 (9.6)	75.6 (2.1)	79.0 (16.6)	71.5 (1.6)	83.0 (10.0)	76.6 (2.0)	82.7 (11.4)	54.9 (0.0)	69.2 (15.8)	87.3 (1.2)	86.6 (6.0)
L3												
25Hz	85.3 (10.6)	84.7 (6.7)	83.9 (9.4)	86.5 (5.9)	76.0 (12.1)	82.0 (10.8)	85.3 (8.5)	87.2 (6.6)	53.1 (18.9)	70.1 (13.5)	87.1 (6.7)	87.0 (10.7)
30Hz	87.6 (4.7)	85.1 (9.8)	84.6 (9.6)	86.2 (7.9)	76.5 (14.7)	82.0 (9.2)	87.3 (5.5)	88.1 (8.3)	57.1 (17.9)	71.5 (14.3)	87.5 (9.0)	88.1 (10.0)
35Hz	88.2 (5.5)	85.9 (8.7)	85.0 (12.4)	86.3 (11.6)	73.9 (14.6)	81.5 (10.8)	87.2 (7.2)	87.2 (10.7)	60.0 (16.2)	72.3 (14.9)	87.2 (8.0)	90.2 (4.5)
40Hz	87.4 (7.3)	86.1 (7.9)	87.2 (11.7)	86.0 (12.2)	77.6 (14.0)	83.5 (11.4)	86.8 (9.8)	87.7 (10.6)	55.3 (20.5)	69.8 (20.9)	87.3 (8.3)	90.4 (4.9)
Forehead												
25Hz	87.8 (9.8)	86.2 (9.9)	64.6 (23.3)	67.8 (24.6)	58.2 (25.5)	64.9 (27.4)	75.1 (16.4)	84.7 (8.1)	69.6 (16.8)	79.3 (13.4)	80.2 (12.9)	84.0 (12.4)
30Hz	88.4 (10.9)	89.6 (8.0)	63.2 (23.4)	69.3 (23.7)	60.4 (24.7)	76.2 (19.4)	77.0 (18.7)	85.1 (10.7)	70.7 (16.2)	80.7 (13.2)	83.6 (10.2)	87.5 (8.2)
35Hz	90.0 (5.2)	90.8 (3.2)	66.8 (21.1)	69.4 (24.7)	69.3 (19.3)	79.7 (15.6)	75.2 (15.5)	82.3 (14.2)	66.9 (17.5)	80.5 (12.3)	79.6 (15.1)	88.2 (6.2)
40Hz	91.0 (3.6)	91.4 (3.5)	71.5 (21.4)	78.2 (23.1)	70.7 (19.6)	84.1 (9.3)	74.8 (16.4)	84.5 (11.9)	66.4 (14.8)	75.1 (16.7)	83.1 (11.0)	88.2 (7.0)

* Values are presented as mean (SD). The value of the platform was obtained by the average of 25 readings (5 accelerometers tested 5 times).

4. DISCUSSION

4.1 Signal amplification and attenuation

Amplification of WBV signals was noted at the medial malleolus for most testing conditions. This site-specific resonance effect was previously observed in the younger populations (Kiiski et al., 2008; Tankisheva et al., 2013). The average transmissibility of up to 2.55 reported here was close to the range of transmissibility values previously reported in the semi- and high-squat postures (range: 1.5-3) among young adults (Tankisheva et al., 2013).

Otherwise, attenuation of WBV occurred in other sites. Generally, transmissibility decreased as the WBV signals traveled from the ankles upward to the head. This phenomenon found in the elderly population was also highly consistent with that in young adults (Tankisheva et al., 2013). The damping occurred as the signals propagated through shock absorbers of the body. The foot arch and the flexibility at the knee and hip also helped cushion the impact generated from the vibration platform (Ker et al., 1987; Lafortune et al., 1996). The muscle activity needed to maintain the body posture, which may also be enhanced by WBV, could also effectively dampen the signal (Bressel et al., 2010). The change in these factors may explain the difference in transmissibility across the various testing conditions, which is discussed in more detail in the following sections.

4.2 Effects of vibration frequency on transmissibility

Higher frequencies led to greater transmissibility at the medial malleolus. This trend was reversed when the signals reached the tibial tuberosity and above, corresponding well to findings in previous studies in young adults (Cook et al., 2011; Harazin and Grzesik, 1998; Kiiski et al., 2008; Muir et al., 2013; Rubin et al., 2003). Our first hypothesis was thus supported. When signals were transmitted above the ankle joint and passed through major muscle groups that act as shock absorbers (e.g., the gastrocnemius), vibration signal propagation was complicated by muscle damping (Bressel et al., 2010; Cardinale and Wakeling, 2005; Pollock et al., 2010). As muscle activation increased with increasing vibration frequency (Liao et al., 2014; Pollock et al., 2010), active damping of the vibration signals could have increased with frequency (Wakeling et al., 2002; Wakeling and Nigg, 2001). Muscle-damping effects may have out-weighed the better transmissibility associated with higher-frequency signals, reversing trends for transmissibility at the tibial tuberosity and above.

4.3 Effects of vibration amplitude on transmissibility

The high amplitude (0.9mm) used in this study resulted in greater transmissibility than did the low amplitude (0.6mm) at the medial malleolus. This trend was reversed at the greater trochanter, L3, and forehead. The results thus supported our second hypothesis. Cook et al. (2011) also showed that transmissibility from platform to shank, as well as from platform to thigh, was higher for low-amplitude (1.5mm) than for high-amplitude vibration (3mm). No data from previous studies were available for comparing the effect of amplitude in upper body sites. Similar to the effect of frequency, muscle tuning effects may apply here, as higher vibration amplitude may give rise to greater muscle activation (Hazell et al., 2007; Lam et al., 2016; Marín et al., 2009).

4.4 Effect of posture on transmissibility

Different postures significantly impacted on WBV transmissibility, thus supporting our third hypothesis. By including postures with unequal weight-bearing between the two legs (i.e., single-leg standing, forward lunge), we are the first to investigate the effect of weight-bearing on transmissibility. An interesting observation was noted for single-leg standing. At the tibial tuberosity, transmissibility was similar between single-leg and erect standing. However, at the greater trochanter, transmissibility during single-leg standing was 1.5–1.9 times greater than that during erect standing at all vibration frequencies and amplitudes. The increase in weight-bearing, which may increase compression force at the knee, hence lower-limb rigidity (Marouane et al., 2015), appeared to be the primary cause of this observation. This explained the greater signal transmission to the greater trochanter as it was the first measured body site reached after signals have passed through the knee joint. Transmissibility during erect standing exceeded that during single-leg standing at the L3 vertebra and head because signals transmitted from both legs joined at the lumbar region.

The effect of the knee angle on transmissibility could be deduced from comparing three postures (i.e., erect standing [20°], semi-squat [45°], and deep squat [70°]). We found that erect standing (20°) yielded significantly greater transmissibility than that for semi-squat and deep squat at all measured sites. These findings were consistent with previous studies in younger adults (Avelar et al., 2013; Muir et al., 2013; Tankisheva et al., 2013). The effect of knee angle was the most prominent at L3. Bending the knee while maintaining an erect upper body inevitably increased the hip flexion angle, thereby augmenting the damping effect as the signals were transmitted further up to the lumbar spine. The effect of knee angle on transmissibility may also be attributed to the greater degree of leg muscle activation associated with a deeper squat, as discussed previously (Avelar et al., 2013; Pollock et al., 2010; Wakeling et al., 2002).

Toe-standing yielded the lowest transmissibility. Direct body contact with the vibration platform is considerably smaller in the toe-standing posture (i.e., forefoot) than in other postures (i.e. whole plantar aspect of the foot). The arch of the human foot has spring-like properties (Ker et al., 1987). In the toe-standing position, a large part of the feet are lifted off the platform and the metatarsal joints are free to move. This allows the elastic soft tissues in the arch (e.g., plantar aponeurosis, ligaments) to better store and release energy to damp the vibration signals. Movements in the ankle joint also become possible and may help in further damping the signals. Furthermore, greater activation of the gastrocnemius muscle during toe-standing may also augment the damping effect (Wakeling et al., 2002; Wakeling and Nigg, 2001).

The above factors (e.g., weightbearing, joint angle, contact area with platform, muscle activity) may also explain why the change in posture would modify the effect of WBV frequency and amplitude on transmissibility (i.e., interaction effect), thus supporting our fourth hypothesis. Thus, apart from WBV frequency, amplitude, the posture assumed should also be carefully considered when prescribing WBV exercise to older adults.

4.5 Signal purity

Vibration signal purity was largely retained during its transmission from the platform to the head. Only Kiiski et al. (2008) have investigated WBV signal purity in a small group of young adults during erect standing. With a similar WBV amplitude (1mm) at 30 Hz and 40 Hz, the proportion of signal power within 1 Hz of the nominal frequency at the greater trochanter reported in their study (78–92%) was similar to our findings (82–85%). However, the signal purity reported at the medial malleolus (35–38%), tibial tuberosity (64–70%), and L3 (50–75%) in their study was generally lower than what we found (medial malleolus: 86–93%, tibial tuberosity: 80–89%, L3: 85–88%). Variability in signal purity data reported in their study (SD, 2–38%) was also considerably higher than in ours (SD, 1–12%). The smaller sample size used in their study and different characteristics of the participants (e.g., young vs.

old) may partially explain these differences. Details regarding the frequency domain analysis, signal purity at the platform level, and methods used to monitor the posture of the participants were not reported in Kiiski et al. (2008). Therefore, it was difficult to make a direct comparison between their study and ours.

4.6 Association of age and BMI with transmissibility

No previous research has examined the association of WBV transmissibility with age and BMI. The results of our correlation analysis were largely insignificant. Out of the 240 correlation coefficients generated for each variable, only 3 and 7 were found to be significant for age and BMI respectively. There was no reasonable physiological explanation as to why age and BMI were only correlated with only a few specific posture × frequency × amplitude × body site combinations but not the majority of others. Moreover, the significant correlations also followed no identifiable pattern. A more plausible explanation is that the few "significant" correlations found were likely related to the increased chance of committing a type I error as a large number of correlation coefficients were generated.

4.7 Study limitations

The results can be generalized only to older adults who shared similar characteristics of our subjects. Muscle activity was not examined in conjunction with transmissibility. The correlation between muscle activation and transmissibility could thus not be evaluated to verify muscle tuning effects. The platform accelerations were not measured synchronously with the accelerations at the body parts. We acknowledge this as a limitation but it should have little impact on our results, given the small difference in platform accelerations in loaded and unloaded condition as well as the consistent output over time.

Skin-mounted accelerometers were used in this study; thus, translations of the accelerometers on the skin could not be totally ruled out, despite efforts to ensure secure mounting. More invasive assessment (e.g., implantation of transcutaneous pins (Rubin et al., 2003)) was not implemented to ensure patient safety. Skin mounted accelerometers were also the most commonly used method to measure accelerations in similar studies, which also facilitate comparison of results across studies (Cook et al., 2011; Kiiski et al., 2008; Tankisheva et al., 2013). Our sample includes more women than men. Due to the small number of men, we did not perform subgroup analysis to compare the transmissibility between men and women. However, considering that BMI and age showed no significant difference between the two subgroups (Table 1) and that these two factors were not strongly associated with transmissibility, the impact of biological sex on transmissibility, if any, should be minimal. However, this postulation will need to be confirmed in future research.

4.8 Practical applications

WBV transmissibility to the head was greatly dampened at the head, which largely eased the safety concern of WBV treatment. Lower vibration frequencies and amplitudes are better at retaining signal power during propagation in the human body and may be more appropriate if the goal was to provide mechanical strains to bone structures in the lower extremities/lumbar spine to maintain or improve bone health among older adults. However, the decreased transmissibility to the upper body associated with higher vibration frequencies and amplitude was possibly due to the muscle tuning effects (Wakeling et al., 2002; Wakeling and Nigg, 2001), which may imply a better lower-limb muscle training effect. Static erect standing posture should be avoided, as its transmissibility to the head is much greater than that for other postures. In view of its good signal transmission to the greater trochanter and lumbar spine and its ability to retain signal purity, single-leg standing strikes the best balance between achieving therapeutic effect on bone health (i.e., good transmissibility to the lower limbs and spine) and ensuring safety (i.e., low transmissibility to the head). Finally, considering that amplification of signals occurred at the medial malleolus, the protocols adopted in the current study should not be used in the treatment of people with ankle pathology.

4.9 Conclusions

Resonance was observed only at the medial malleolus. Above the knee joint, lower WBV frequencies and amplitudes yielded better transmissibility. Single-leg standing yielded the highest WBV transmission to the hip without augmenting transmission to the head. Nevertheless, signal transmissibility depended on the interaction between frequency and posture and, to a lesser extent, the interaction between amplitude and posture. Signal purity was well conserved during WBV transmission.

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Contributors

Study design: FMHL, CYT, and MYCP. Study conduct: FMHL. Data collection: FMHL. Data analysis: FMHL. Data interpretation: FMHL, CYT, TCYK, and MYCP. Drafting manuscript: FMHL. Revising manuscript content: FMHL and MYCP. Approving final version of manuscript: FMHL, CYT, TCYK, and MYCP. FMHL takes responsibility of the integrity of the data analysis.

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