Vibration transmissibility and apparent mass changes from vertical whole-body vibration exposure during stationary and propelled walking

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A B S T R A C T

Whole-Body Vibration (WBV) is an occupational hazard affecting employees working with transportation, construction or heavy machinery. To minimize vibration-induced pathologies, ISO identified WBV exposure limits based on vibration transmissibility and apparent mass studies. The ISO guidelines do not account for variations in posture or movement. In our study, we measured the transmissibility and apparent mass at the mouth, lower back, and leg of participants during stationary and propelled walking. Stationary walking transmissibility was significantly higher at the lumbar spine and bite bar at 5 and 10 Hz compared to all higher frequencies while the distal tibia was lower at 5 Hz compared to 10 and 15 Hz. Propelled walking transmissibility was significantly higher at the bite bar and knee at 2 Hz than all higher frequencies. These results vary from previously published transmissibility values for static participants, showing that ISO standards should be adjusted for active workers.

1. Introduction

The 6th survey on European working conditions showed that, in 2015 20 percent of employees were exposed to vibration during work. According to Krajnak (2018), workers of transportation, agriculture, construction, and manufacturing industries are exposed to Whole-Body Vibration (WBV) on a daily basis. Krajnak (2018) explains that WBV is associated with various adverse health conditions from fatigue, motion sickness and headaches, to musculoskeletal disorders, digestive issues, neuropathies, cardiovascular disease, type II diabetes, and potentially cancer. Due to the high percentage of workers exposed to WBV and the associated detrimental effects, daily vibration exposure limits were established for the safety and comfort of humans exposed to WBV (EU Directive, 2002/44/EC). The Directive imposes that the ISO 2631-1 (1997) is used to evaluate the risk. The 2631-1 standard describes methods for direction, location, and duration of vibration measurement as well as evaluation methods and frequency weighting. Yet, when the application of these methods is addressed, the standard specifically explains that the effects of WBV on standing, reclining or recumbent persons is unknown the assessment is, "usually carried out using the same evaluation method as for seated persons" (ISO 2631-1, 1997).

The ISO standard derived the frequency weighting curves from the laboratory tests existing in the literature. Experiments are conventionally divided into force-motion (impedance or apparent mass) and motion-motion (transmissibility) test: the mechanical response of the biological tissues is in turn used to estimate the risk of injury at the workplace (Mansfield, 2005). Different studies showed that the human response to vibration is different not only between the seated and standing position, but also changes based on posture and muscular activity (Hinz and Seidel, 1987; Fairley and Griffin, 1989; Mansfield and Griffin, 2000; Matsumoto and Griffin, 2002; Tarabini et al., 2013). The surveys and epidemiologic studies about pathologies/pain occurrence related to WBV exposure in various workplaces (Bongers et al., 1988; Bovenzi and Betta, 1994; Bovenzi et al., 2002; Hulshof and van Zanten, 1987) reveal a lack of correlation between the exposure and the pathologies. This can be partially due to the effect of posture (Tarabini and

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Valsecchi, 2017) and to the presence of motion during the vibration exposure.

The dependence of the biomechanical response of standing people from different parameters is well documented in the literature. Some of the laboratory WBV studies showed that the transmissibility to the head and body segments of standing people is dependent upon amplitude, frequency and direction of vibration as well as the posture (Garg and Ross, 1976; Harazin and Grzesik, 1998; Paddan and Griffin, 1993). Continued study of the transmissibility of standing people has revealed that the body's response to WBV is heavily affected by postural differences (Harazin and Grzesik, 1998; Subashi et al., 2008; Tarabini et al., 2013). In addition to transmissibility, the apparent mass of humans exposed to WBV both while standing and seated has also been shown to change with frequency and posture, and is non-linear (Fairley and Griffin, 1989, 1990; Matsumoto and Griffin, 2003; Tarabini et al., 2014, 2016). Most of the studies (Nawayseh and Griffin, 2006; Tarabini et al., 2013, 2014, 2016) evidenced that the response of the standing human body depends on the knee joint angle. Seeing that there were already evident differences in the human response to WBV with different standing postures, it was thought that there may be a need for the investigation of the human response to vibration while walking.

To date, very little research has been published on the topic of the human response to WBV during dynamic activities. One of the first studies was by Smeathers (1989) to measure transmissibility of the spine during walking and running. Since then, Morgado Ramirez et al. (2013, 2017) have studied the transmissibility of the spine during daily activities, and during walking to understand the difference in transmissibility between regions of the spine. Munera et al. (2016) have also approached transmissibility of WBV at several bodies' location during half-squat exercises. However, a study to understand the apparent mass and transmissibility of the whole body at different locations during walking has yet to be published.

While the research thus far serves as a valuable foundation for understanding the human response to WBV in static scenarios, not all workers remain static throughout an entire workday. However, the current literature and subsequent safety standards have not been adapted for moving participants. Since the scientific literature has shown the effect of postural changes in static people (both seated and standing), it is expected that the human response to WBV in dynamic conditions will be different from that of static conditions. Accordingly, it is important to have measurement and safety standards which reflect the actual behavior of a participant's apparent mass and transmissibility during the dynamic movement experienced in a working environment. The objective of this study is to investigate how vertical WBV affects the human body response at various frequencies during walking. The apparent mass and transmissibility were evaluated while walking in two different conditions. First, during stationary walking (also known as walking 'in place' or 'on the spot') atop a vibrating platform, and second, during propelled walking on a treadmill mounted on a vibrating platform. The results of this study may provide a new approach to future research on WBV in dynamic participants and cause a reevaluation of the current vibration exposure limits or frequency weighting curves.

2. Materials and methods

2.1. Participants

This two-part experiment consisted of testing seven male participants (23 ± 4 years old, 1.79 ± 0.05 m tall, 73.9 ± 9.7 kg; mean ± SD) during stationary walking, and nine male participants (29 ± 7 years old, 1.78 ± 0.07 m tall, 77.8 ± 9.9 kg; mean ± SD) during propelled walking on a treadmill. Exclusion criteria included: diagnosis of a physician to have diabetes, vibration-induced pathologies, a concussion, or a lower body musculoskeletal injury within the previous six months, suffering from motion sickness, or allergy to the required adhesive tape. The personal data was collected following the General Date Protection Regulation criteria and all the participants signed an informed consent. Tests were performed according to the University Ethics Guidelines and the standards of the Declaration of Helsinki.

2.2. Stationary walking

2.2.1. Experimental conditions

Participants walked stationary on a rigid platform driven by an electrodynamic shaker LDS V930 (LDS, England) which provided the vertical WBV as shown in Fig. 1. The same setup was used by Tarabini et al. (2013, 2014). Walking trials were performed in three repetitions of seven randomized tests, which lasted 90 s each, with 1 min of rest between each test. The seven tests consisted of six different vibration frequencies: 5, 10, 15, 20, 25, 30 Hz, and one test without vibration. The vibration amplitudes and frequencies are similar to those used in the literature which study the response of the human body to vertical WBV (Matsumoto and Griffin, 1998; Nawayseh and Griffin, 2003; Tarabini et al., 2014). The maximum frequency was limited by the bandwidth of the force measurement system (30 Hz). The vibration amplitude of 2 m/s^2 was chosen to maximize the signal to noise ratio while allowing for the measurement of acceleration at the mouth, given the low transmissibility expected at high frequencies. According to the EU directive 2002/44/EC, the daily exposure limit value of employees for WBV is 1.15 m/s^2. Hughes and Ferrett (2008) and Griffin (2004) demonstrated that the acceleration exposure can be higher with shorter exposure times. Being that our trials lasted only 90 s, the exposure level was well below the EU directive threshold.

Since stationary walking results in much lower contact forces than propelled walking, participants were instructed to walk without shoes to avoid any differences due to shoe design. All three repetitions of the seven tests were performed on the same day. After one series was completed, participants were given five minutes of rest before beginning the next repetition.

2.2.2. Acceleration

Participants were fitted with three ADXL335 (Analog Devices, MA, USA) triaxial accelerometers to measure the vibration transmitted to their body; one applied to the distal tibia, one applied to the lumbar spine (L5), and one mounted to a bite bar held in the mouth. Platform vibration was measured using a B&K 4508B (Brüel & Kjær S&V, Denmark) accelerometer. Acceleration signals were sampled using NI 9234 acquisition cards (National Instruments, TX, USA) with a frequency of 2048 Hz, connected to a PC via a NI 9172 (National Instruments, TX, USA) cDAQ chassis.

2.2.3. Inertial force

The ground reaction force of the participant was measured by four PCB 212B (PCB Piezotronics, NY, USA) load cells, placed between the platform and the shaker actuator. All the signals from the load cells, platform control accelerometer, and accelerometers were acquired simultaneously.

2.2.4. Stride segmentation

Heel strikes were highlighted as the instants when the vertical component of the inertial force exerted by the participant exceeds 50 N. Concomitant peaks in the right distal tibia acceleration were identified to specifically extract the right heel strikes. Heel strike detection was started 10s after the trial beginning and stopped 5s before the trial ended to ensure that all analyzed strides were consistent with a normal walk. Thirty-five strides (N1) was the largest sample which could be analyzed for all the participants. Since, participants will naturally move at their most metabolically efficient stride frequency (Minetti et al., 1995), they were instructed to step at a self-selected cadence. It resulted in a walking frequency of 1.4 ± 0.2 Hz (mean ± SD) over all the investigated conditions.

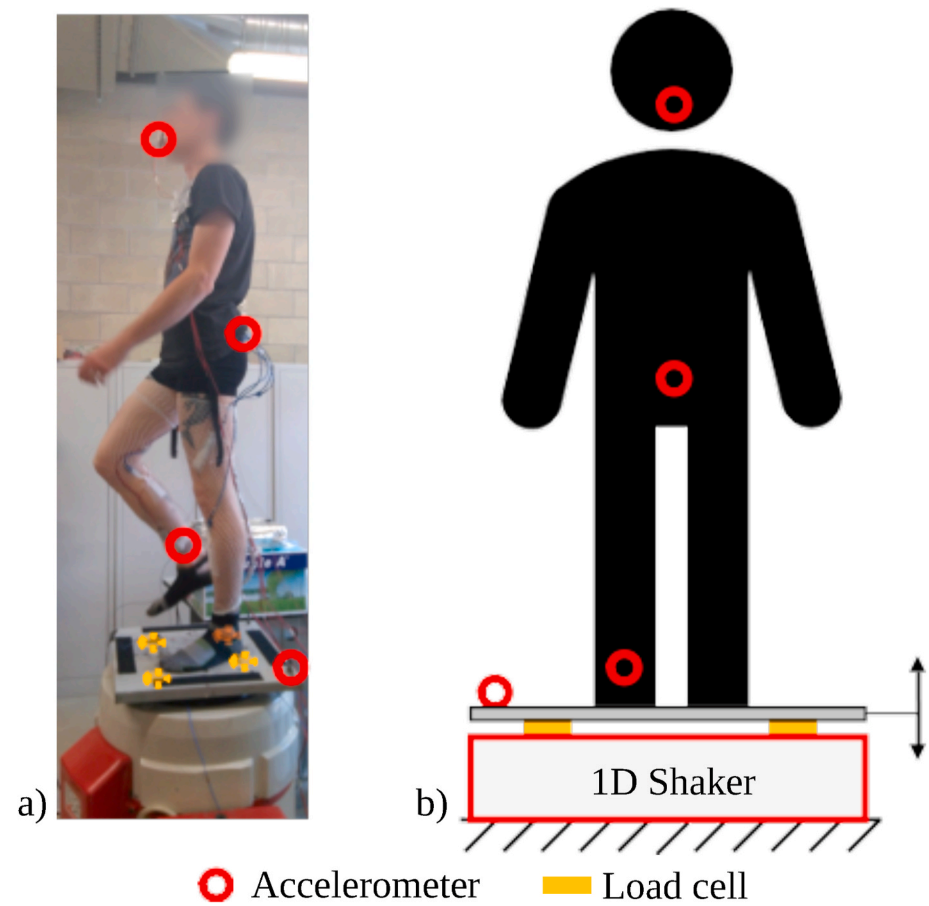

Fig. 1. Photo (a) and graphic diagram (b) of a participant performing a trial of the *Stationary Walking* experiments.

2.3. *Propelled walking*

2.3.1. *Experimental conditions*

Participants were instructed to walk atop a treadmill mounted to a six-axis vibration platform, purposely designed by MTS for Istituto Nazionale per l'Assicurazione contro gli Infortuni sul Lavoro (INAIL) Research Center of Monte Porzio Catone as shown in Fig. 2. Participants wore identical shoes and walked at freely chosen stride frequency. Walking velocity was constant in all the tests (1.25 m/s). Participants were exposed to vertical harmonic vibration with a constant amplitude of 2.5 mm which was chosen to be able to detect, with a constant Signal to Noise Ratio, the motion of the participant for the gait and muscular activation analyses (not described in this work). Walking trials were performed in one session of seven tests, which lasted 100 s each. The seven tests consisted of six different vibration frequencies: 2, 4, 6, 8, 10, 12 Hz, and one test without vibration to serve as a control. The maximum frequency was limited by the treadmill's first natural frequency (15 Hz). After participants were fitted with the sensors, they were instructed to mount the treadmill and remain still as the sensors were activated. The treadmill would then start and slowly accelerate until reaching full speed. Once the participants were walking at a constant velocity, the 100 s trial began. All tests both with and without vibration were performed on the same day. After one test was completed, the participants were given five minutes rest before beginning the next tests.

2.3.2. *Acceleration*

The vibration was measured by three accelerometers: two triaxial PCB 356A22 (PCB Piezotronics, NY, USA) accelerometers; one placed on the lateral epicondyle of the right knee, and one mounted to a bite bar held in the mouth, and one B&K 4518-002 single-axis accelerometer (Brüel & Kjær S&V, Denmark) was placed on the heel of the participant's right foot. Platform vibration was measured using a B&K 4508B (Brüel & Kjær S&V, Denmark) accelerometer. The acceleration signals were sampled using NI 9234 acquisition cards (National Instruments, TX, USA) with a frequency of 2048 Hz, connected to a PC via a NI 9172 (National Instruments, TX, USA) chassis.

2.3.3. *Inertial force*

The contact pressure exerted between the participant and the surface of the treadmill was measured by means of Pedar (Novel, Munich, Germany) sensorized insoles, sampled with a frequency of 100 Hz. The signals from pressure insoles were synchronized with the accelerometers post-hoc by identifying the moments where the right foot touches the ground for the first time (here forward called, "heel strike").

2.3.4. *Stride segmentation*

Before the database could be segmented into strides, pressure and acceleration signals had to be synchronized. The pressure applied to the ground was derived from the insole's output signal corrected according to (Saggin et al., 2013). The heel strikes were identified in the pressure signals as the instants where the pressure exceeded 400 kPa with an increasing slope. The average time duration (Tstep) between two consecutive heel strikes was then derived from the heel strike detection. The accelerometer heel strikes were identified in the right heel acceleration signals as the instants where the acceleration exceeds 50 m/s2 with a minimal duration of 0.8Tstep between two consecutive strikes. Accelerometer and pressure signals were then aligned according to heel strikes detections and the database was segmented into strides. Following the same stride selection as the *Stationary Walking*

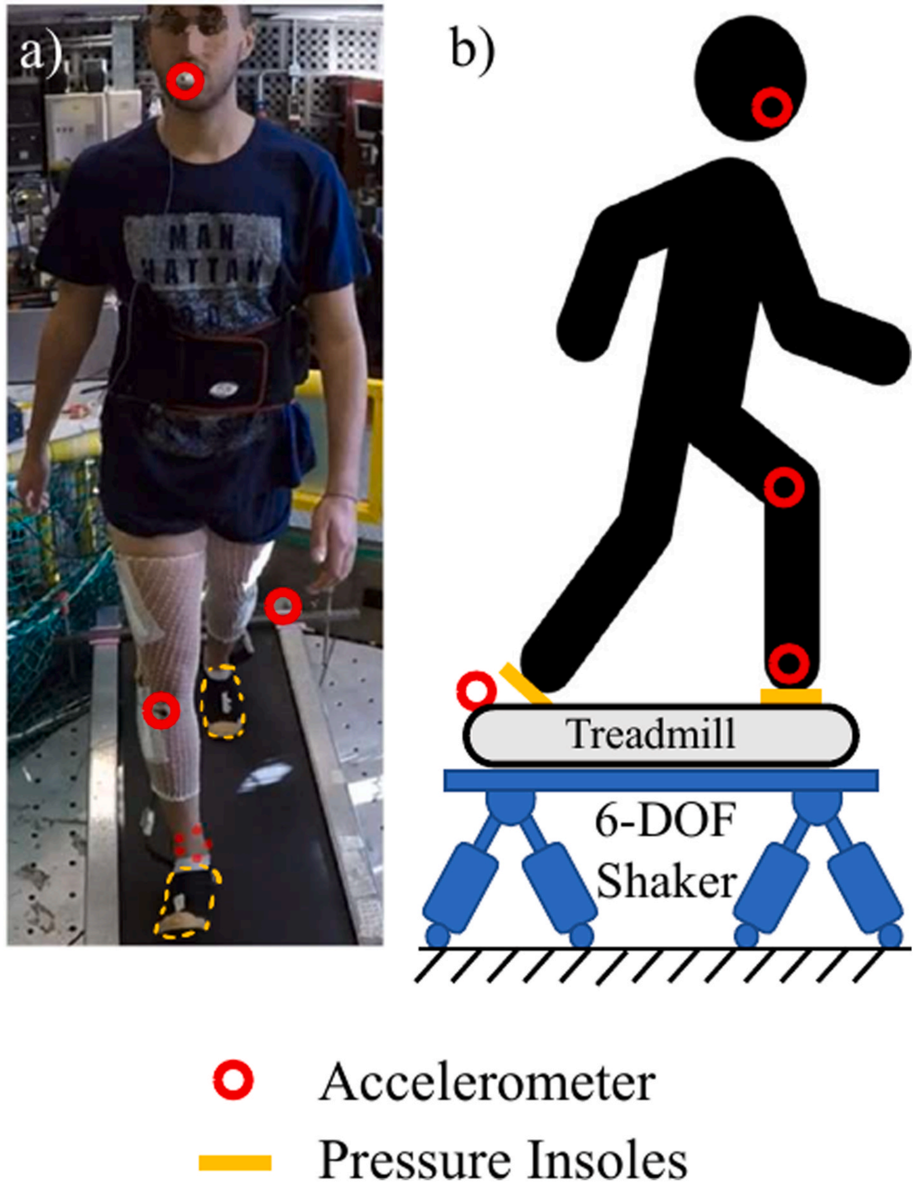

Fig. 2. Photo (a) and graphic diagram (b) of participant performing a trial of the *Propelled Walking* experiments.

experiments, 48 strides (N2) were identified.

2.4. Data analysis

The signals were not frequency-weighted. The apparent mass and transmissibility were computed according to the method used by Tarabini et al. (2013).

2.4.1. Vibration transmissibility

The vibration transmissibility was estimated from the driving point (supporting platform or treadmill) to each accelerometer located on the body (i.e. distal tibia, lumbar spine, and bite bar in the *Stationary Walking* experiment and at the lateral epicondyle of the right knee and the bite bar in the *Propelled Walking* experiment). For each location k, transmissibility is defined as

$$T^k(f) = \frac{S_k(f)}{S_{In}(f)} \tag{1}$$

where f is the frequency vector, S_k is the unweighted spectrum of the acceleration at the location k and S_{In} is the unweighted spectrum of the harmonic excitation at the driving point. The choice of not using spectral estimators was dictated by the fact that during the *Propelled Walking* experiment, the average walking frequency (1.8 Hz) was close to the first excitation frequency (2 Hz). It implies that the spectral leakage of the force exerted by the foot on the platform could lead to a biased estimation of the transmissibility and apparent mass; S_{In} and S_k were therefore computed on the entire signal and the magnitude of $|T^k(f)|$ was estimated at each excitation frequency without performing averages.

2.4.2. Apparent mass

The human body apparent mass was computed as

$$AM(f) = \frac{S_F(f)}{S_{In}(f)} \tag{2}$$

where S_F is the unweighted spectrum of the inertial force. According to Tarabini et al. (2013), the signals were not frequency-weighted, but $AM(f)$ was normalized by the participant's static weight and its magnitude and phase were estimated at each frequency of excitation. To allow comparison with the literature while standing in neutral position and while standing in neutral position with knee bent (Subashi et al., 2008), $AM(f)$ was estimated up to 20 Hz.

2.4.3. Statistical analysis

Repeated analyses of variance were carried out to highlight the effect of the excitation frequency on the vibration transmissibility and the apparent mass during stationary walking on a platform or propelled walking on a treadmill. As the two experiments were conducted differently, no statistical analysis was performed to compare the two walking conditions. When a significant effect was observed ($p < 0.05$), a multiple comparison of estimated marginal means was carried out to determine the conditions leading to significant differences.

3. Results

3.1. Stationary walking

A significant interaction effect of the excitation frequency and location occurred on the transmissibility during stationary walking ($F(10,125) = 19.0$; $p < 0.01$). The post-hoc analysis indicated that the transmissibility at the lumbar spine and the bite bar was significantly higher at 5 Hz and 10 Hz than at higher frequencies. This phenomenon occurred to a lower extent at the distal tibia for which the transmissibility at 5 Hz was interestingly estimated lower than at 10 Hz and 15 Hz. Conversely, the transmissibility to the distal tibia at 10 Hz and 15 Hz was significantly lower than at higher frequencies. A significant difference was also found in the AM at 5, 10, and 15 Hz compared to all other frequencies. A summary of the results from Stationary Walking can be seen in Table 1.

3.2. Propelled walking

A significant interaction effect of the excitation frequency and location occurred on the transmissibility while propelled walking on a treadmill ($F(5,106) = 7.2$; $p < 0.01$). The post-hoc analysis indicated that, at 2 Hz, the transmissibility to the right knee was significantly higher than at the bite bar. It also revealed that the transmissibility was significantly higher at 2 Hz than above whatever the location. A significant difference was also found in the AM at 5, 10, and 15 Hz compared to all other frequencies. A summary of the results from Propelled Walking can be seen in Table 2.

3.3. Comparison between stationary and propelled walking

Comparing the two experiments, stationary walking induced higher transmissibility than propelled walking up to 12 Hz. Fig. 3 reveals that the transmissibility of vibration was slightly higher than 1 below 5 Hz (i.e. the vibration was amplified), especially at the lumbar spine and the right knee for stationary walking and propelled treadmill walking, respectively. Fig. 3 also illustrates a local increase in the distal tibia and right knee transmissibility in the [10–15] Hz bandwidth, although the transmissibility remains lower than 1. The mean transmissibility decreases with the vibration at all locations, in both experiments.

Table 1
A summary of the transmissibility values estimated during the *Stationary Walking* experiments. For a given location, * indicates a significant difference with respect to the higher excitation frequencies. For a given excitation frequency, † indicates a significant difference with respect the other locations.

	5 Hz	10 Hz	15 Hz	20 Hz	25 Hz	30 Hz
T – Bite bar	0.60 (0.11) *	0.58 (0.08) *	0.40 (0.07)	0.30 (0.06)	0.18 (0.05)	0.13 (0.03)
T – Lumbar spine	1.19 (0.36) *†	0.55 (0.11) *	0.26 (0.11) †	0.12 (0.05)	0.09 (0.04)	0.07 (0.05)
T – Distal tibia	0.62 (0.08)	0.74 (0.11) *	0.66 (0.10)	0.50 (0.20)	0.39 (0.11)	0.33 (0.07
|AM|	0.92 (0.08) *	0.29 (0.04) *	0.21 (0.06) *	0.13 (0.03)	0.08 (0.01)	0.07 (0.01)

Table 2
A summary of the transmissibility values estimated during the *Propelled Walking* experiments. For a given location, * indicates a significant difference with respect to the other excitation frequencies. For a given excitation frequency, † indicates a significant different with respect to the other location.

	2 Hz	4 Hz	6 Hz	8 Hz	10 Hz	12 Hz
T – Bite bar	0.92 (0.15) *†	0.41 (0.13)	0.28 (0.11)	0.31 (0.06)	0.26 (0.06)	0.21 (0.05)
T – Right knee	1.49 (0.59) *†	0.37 (0.15)	0.29 (0.12)	0.22 (0.07)	0.27 (0.09)	0.31 (0.15)
|AM|	1.33 (0.38) *	0.90 (0.15) *	0.39 (0.07) *	0.25 (0.04)	0.19 (0.03)	0.15 (0.02)

Fig. 4 illustrates the apparent mass modulus (a) and phase (b). The lack of synchronization between the pressure and acceleration signals prevented a reliable evaluation of the apparent mass phase for the *Propelled Walking* experiments. In both walking experiments, the apparent mass modulus decreased with increased excitation frequency. The resonance frequency was not clearly identified during stationary walking because of the frequency resolution of the tests (5 Hz). The resonance seemed to appear at 2 Hz or lower frequencies when walking propelled.

Values can be compared with the apparent mass of standing people already published and measured on the same experimental setup (Tarabini et al., 2013). When participants stood on the vibrating platform without walking, resonances occurred around 5 Hz and 2 Hz when standing in neutral position and with knee bent, respectively. Similar to the magnitude, the phase decreased with increased excitation frequency, and was consistent with data of standing people. In this case, the behavior was closer to that of participants standing in neutral posture than to the behavior of participants standing with bent knees.

4. Discussion

The biomechanical response of participants walking stationary and propelled under harmonic WBV has been measured. Two new experimental setups have been designed and dedicated data processing and synchronization methods have been applied.

4.1. Vibration transmissibility

The transmissibility of the vibration during walking globally decreases with the distance from the point of interest to the ground in both the experiments. This result is in accordance with numerous studies indicating that the transmission of vibration is reduced above the knees while exposed to WBV (Matsumoto and Griffin, 1998; Munera et al., 2016) as well as to shock-induced vibrations when running (Chadefaux et al., 2019; Gruber et al., 2014). This phenomenon is supported by several hypotheses such as the dealignment of the segments (Vasconcellos et al., 2014) and the changes in the musclular mechanical properties with co-contraction (Heitmann et al., 2012). Considering running, these attenuation processes are commonly classified as passive (physiological attenuation of the anatomical elements such as the fat pad (Chu et al., 1986) or active (segment orientations (Lafortune et al., 1996), activations of the muscular system (Boyer and Nigg, 2007). We observed that the transmissibility was higher at the lumbar spine than at the distal tibia and the bite bar at 5 Hz but lower at higher frequency. While standing upright, Yang et al. (2012) also reported that the transmissibility at the pelvis was higher than at the ankle and head at low frequencies while it was lower above about 12 Hz. Yang et al. (2012) reported that as knees become progressively straighened and rigid, transmission of vibration increase, especially at the pelvis for which the resonance of the transmissibility curve increased from 0.45 to 1.5 at around 9 Hz. This outcome, which is also supported by Rubin et al. (2003), is of great interest with respect to the human health since it

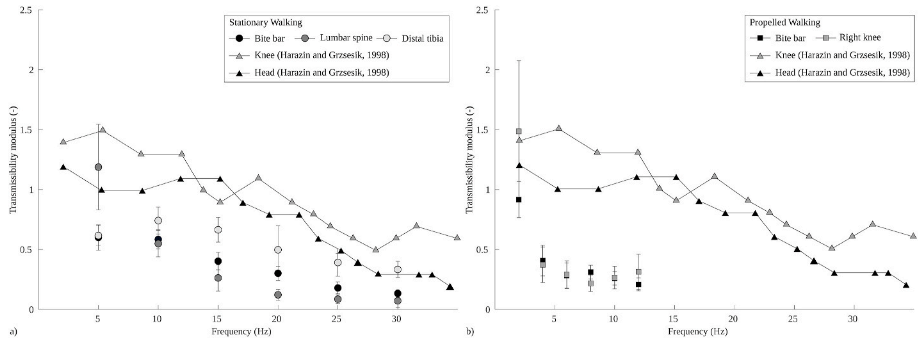

Fig. 3. The mean ± SD of the transmissibility estimated at each excitation frequency across all participants measured at the distal tibia, the lumbar spine, and the bite bar during the *Stationary Walking* experiments (a) and at the knee, and bite bar during the *Propelled Walking* experiments (b).

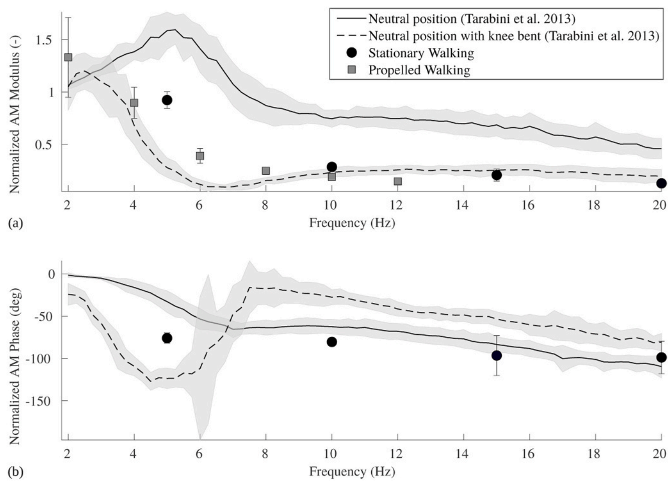

Fig. 4. The mean ± SD of the apparent mass modulus (a) and phase (b) of participants walking stationary (circles) and propelled (squares). These results are plotted over previously published data (Tarabini et al., 2013) illustrating the apparent mass (AM) with the SD (grey shaded area).

affects fundamental anatomical regions such as the pelvis and the lumbar spine.

The vibration transmissibility decreased with increased frequency of excitation, which is consistent with static posture in previously published literature (Matsumoto and Griffin, 1998; Nawayseh and Hamdan, 2019). In our tests, the input vibration was amplified below 5 Hz at the lumbar spine and the knee. A noteworthy outcome was the local increase of vibration transmissibility in the 10–15 Hz bandwidth regarding the distal tibia and the knee. This observation may be related to the resonance frequency of shank elements which have been reported slightly higher between 15 and 30 Hz (Wakeling et al., 2002). While a thorough investigation would be required to accurately evaluate the influence of shank elements, excitation in this bandwidth could be a potential source of discomfort or ailment and should therefore be considered carefully.

4.2. Apparent mass

The apparent mass is consistent with the data reported in the literature. Although no accurate value can be provided because of the discrete characteristic of our tests, the resonance frequency appeared below 5 Hz. These observations may coincide with previous literature which reported that the apparent mass of a standing person presents a resonance at around 5.5 Hz and sometimes another peak between 9 and 14 Hz (Matsumoto and Griffin, 1998). Interestingly, Matsumoto and Griffin (1998) also reported that with knee bent, the resonance frequency of the apparent mass decreased to about 2.75 Hz with low magnitudes above 7 Hz, which is in accordance with data reported by Tarabini et al. (2013). With respect to these results, our findings indicate that the apparent mass of the human body exposed to WBV while walking is closer to standing with knee bent than in a neutral posture.

The phase of the apparent mass could only be measured during the *Stationary Walking* Experiments. The phase was approximately −80° at 5 Hz and decreased almost uniformly until −115° at 30 Hz. This aspect might be interesting in the field of human-structure interaction given that coupling between the human body and a structure occurs at low frequencies (Van Nimmen et al., 2017) and that there is a current lack of experimental data describing the apparent mass of walking participants. The phase indicates that, above 5 Hz the walking participant has a damper-like behavior showing that magnitude decreases with increased frequency.

4.3. Comparison between stationary and propelled walking

When comparing the two experiments, higher transmissibility as well as apparent mass were obtained during the *Stationary Walking* experiments than during the *Propelled Walking* experiments. As it has already been reported that standing shod or barefoot does not greatly affect the apparent mass (Tarabini et al., 2013), the observed difference is most likely related to variations in the gait kinematics with respect to the two experiments. Participants impacted the ground barefoot with the forefoot during stationary walking while they impacted shod with the rearfoot or midfoot when walking on a treadmill. Consequently, a hypothesis would be that the vibration was cushioned by soft tissues such as the gastrocnemium muscles and by the knee and ankle joints when walking on a treadmill (Nordin et al., 2017). A second hypothesis would be that, since they were limited to the plate, participants had a more constrained gait with less knee flexion during stationary walking than on a treadmill. It therefore induced a higher transmissibility and apparent mass since, as already shown (Tarabini et al., 2013), the apparent mass is higher when standing in neutral position than with knee bent. Those results are also supported by several studies indicating that bending the knees induces higher damping of the vibrations especially through muscular activation (Harazin and Grzesik, 1998; Wakeling et al., 2002). Finally, the modulus of the apparent mass was estimated between its value while standing in neutral posture and with knee bent. It suggests that walking may be seen as a series of consecutive neutral and bent knee postures.

4.4. Limitations

The main limitation of our study is that the response has been determined only at discrete frequencies steps of 2 Hz and 5 Hz. The choice was dictated by the necessity to limit the total duration of tests and prevented the identification of a resonance during stationary walking.

Another limitation arises from the fact that the transmissibility in the two tests was measured at different locations on the body. This limitation was due to EMG electrodes which were placed on the participants during the treadmill tests (results not reported here) and due to the different experimental setups used in the two laboratories.

The last limitation is that this study focused on WBV along the vertical direction, while lateral and fore-and-aft vibration might have relevant effects on human body integrity, as testified by the 1.4 multiplying factors that have to be applied to the horizontal vibration for the computation of the WBV exposure according to the ISO 2631-1 (1997).

4.5. Implications in standardization

The current standards for the evaluation of vibration-related risks have been derived from the biomechanical response of the human body to vibration. Our study has proven that the response to vibration while walking differs from that measured in a standing posture. The transmission of vibration to human body segments and the human body apparent mass are dependent not only on the global body posture, but also on the dynamical aspects of the activity.

The values of transmissibility and apparent mass will be the basis for the development of biomechanical models as in (Chadefaux et al., 2020). Models, in turn, will open new perspectives in the design of effective protection elements. Future works will consist of investigating the effect of WBV exposure on biomechanical and physiological features. For this purpose, the study will be enhanced by a kinematicanalysis and a monitoring of the lower-limb muscles activation while exposed to vibration during gait. In addition to exposure to vertical vibrations, the fore-and-aft and medio-lateral directions will also be investigated to complete the understanding of the phenomenon and to get as close as possible to actual occupational exposures. Finally, this generic overview of WBV exposure will contribute to proposing a review of the current legislation framework, with weighting curves or multiplication factors derived from the biomechanical response.

5. Conclusion

This study has experimentally described the vertical WBV transmissibility and apparent mass during walking. The WBV transmissibility generally decreased with the distance from the driving point and with the frequency of the vibration. The behavior was similar in stationary and propelled walking, with transmissibility magnitudes that were dependent on the flexion of the lower-limb joints. The apparent masses during stationary and propelled walking were intermediate between those reported in the literature for standing participants in neutral posture and with bent knees. The phase of the apparent mass demonstrated that the human body, above 5 Hz, acts as an energy dissipator and not as a solid mass. These findings confirm that the current ISO regulations in place for testing and safety regulations in human participants exposed to WBV are not properly adapted to walking people. Further studies are necessary to identify the biomechanical response along the fore-and-aft and medio-lateral directions.

This will provide a more complete understanding of the human response to WBV and contribute to a proposed review of the current legislation framework.

Data availability

The data used to support the findings of this study are available from the corresponding author upon request.

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Declaration of competing interest

The authors declare that they have no known competing financial interest or personal relationships that could have appeared to influence the work reported in this paper.

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