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Paper



Understanding vibration exposure in wheelchair users: Experimental insights

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ABSTRACT

Addressing the complexities of manual wheelchair (MWC) vibrations is crucial for the well-being of users and their integration into society. This study investigates the experimental choices influencing the assessment of vibration exposure, aiming to contribute for enhanced MWC developments and standardized design principles. By conducting a comprehensive full factorial experiment, the impact of various factors, including four MWC loads, two speeds, five floor types, and two MWC models was examined. Notably, findings highlight the predominant influence of floor type on vibration exposure, followed by speed and, to a lesser extent, MWC properties. Furthermore, the study suggests that enlisting an able-bodied participant is more representative than using a dummy when loading the MWC, providing valuable insights into the genuine MWC/user dyad response to vibrations. This research sets the stage for a more informed and standardized approach to address the vibration exposure faced by MWC users.

1. Introduction

The human body is daily exposed to vibration during transport, work or sport activities. Depending on their properties (frequency, magnitude, duration of exposure), vibrations could alter the perceived comfort or have adverse effects on the anatomical structures. For example, studies have linked vibrations to issues such as low-back [1] and neck pains [2], which may be attributed to the deterioration of intervertebral discs [3]. To address these concerns, guidelines have been established for the prevention of workers, defining limit values to regulate exposure duration [4,5]. However, such guidelines are currently lacking for protecting the general population in everyday life, particularly in the context of manual wheelchair (MWC) propulsion. MWC users experience vibration due to the interactions between the wheels and the ground. Current research indicates that, on a daily basis, the vibration exposure of MWC users surpasses the ISO standards limits set for able-bodied workers [6]. This is especially critical in this population because of their inability to perceive the vibrations they are exposed to, often due to weak or absent nervous feedback in the pelvis and lower-back regions.

Within this context, it becomes crucial to quantify the vibration exposure experienced by MWC users under both controlled and realistic conditions. However, adhering to the recommendations of the ISO standard [5] poses challenges in the context of MWC propulsion. Beyond the necessity for embedded and reliable sensors capable of measuring vibration amplitude up to $6 \text{ m}\cdot\text{s}^{-2}$ with a frequency content reaching 100 Hz, inherent precautions must be taken into account concerning the MWC users. Users may encounter difficulties in independently propelling themselves or performing MWC transfers. Consequently, existing studies have often sought to circumvent these challenges by modifying the experimental setup [7]. Vibration exposure has frequently been estimated over shorter durations than recommended and simulated on courses such as a road course [8] or a drum shock simulator [9], rather than over a real road. Similarly, numerous studies have recruited able-bodied participants [10] or used dummies [11]. This methodological diversity added to the variety of MWC settings and individual user specificities [12], has rendered drawing conclusive findings challenging.

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This complexity introduces a notable contradiction between the findings of Garcia-Mendez et al. [13] and Misch and Sprigle [14]: the former asserted that manual wheelchair (MWC) users are overexposed to Whole-Body Vibrations (WBV), while the latter expressed doubt. On one hand, Garcia-Mendez et al. [13] assessed vibration exposure during the daily life of 37 MWC users over a span of 2 weeks, with an average reported speed of $0.7 \text{ m}\cdot\text{s}^{-1}$. Measurements were conducted beneath the seat cushion, along the anteroposterior and vertical directions, yielding an average acceleration of $0.8 \text{ m}\cdot\text{s}^{-2}$. On the other hand, Misch and Sprigle [14] estimated the vibration exposure of a robotic dummy on a lightweight MWC propelled at a controlled speed ($[0.8 - 1] \text{ m}\cdot\text{s}^{-1}$) across four floor types (tile, decorative interior brick, poured concrete sidewalk, and expanded aluminum grates). Measurements were collected at the top surface of the seat cushion along the vertical direction, with reported acceleration values ranging from $0.2 \text{ m}\cdot\text{s}^{-2}$ to $0.4 \text{ m}\cdot\text{s}^{-2}$. Therefore, the two studies varied not only in chair characteristics but also in load, speed, floor type, sensor positioning, and measurement direction. While speed [15] and floor type [16] have previously been identified as factors influencing vibration exposure, the use of a robotic dummy instead of an actual MWC user may be subject to dispute. From a dynamical standpoint, a robotic dummy, composed of rigid bodies, may overlook the damping properties inherent in the human body.

Examining the vibration exposure of MWC users necessitates experimental choices. Establishing standardized experimental methods is crucial to comprehend how MWC users experience vibrations and the potential impact on their comfort and health. These advancements could eventually lead to the development of standards for MWC design or of innovative designs for urban and interior flooring. Building upon the exploration of the vibration response of a MWC under diverse propelling conditions [17], this study aims to establish guidelines for standardizing methods to assess vibration exposure in manual wheelchair users. To achieve this a comprehensive full factorial experiment has been devised to explore the influence of MWC loads, speeds, floor types, and MWC models. Additionally, recognizing the importance of metrological recommendations, the study explores various sensor locations on both the MWC and the users.

2. Method

2.1. MWC load

Conducting studies with manual wheelchair (MWC) users is not always feasible. Consequently, researchers often examine vibration during MWC propulsion using dummies or able-bodied subjects. To assess the validity of this approach, an experiment was conducted with dummies, able-bodied participants, and MWC users.

The dummy used was a 60 kg model compliant with ISO-7176-11:1992 [18]. Such a dummy was designed to simulate an average adult's weight and mass distribution for wheelchair testing. It consists of two blocks made of metal and wood, which can accommodate additional masses to replicate a total weight of around 75 kg. These blocks are connected by hinges, allowing for basic articulation to mimic human movement. The dorsal part is constructed with less rigid materials to better simulate the flexibility of the human body, ensuring realistic and reliable testing conditions.

Both able-bodied and MWC participants were aged between 18 and 60 years and had no musculoskeletal issues in their shoulders or back. Able-bodied participants, who did not typically use MWCs for mobility, included 18 individuals (9 men and 9 women) with an average age of 34 ± 15 years, height of 172 ± 8 cm, and weight of 63 ± 6 kg. The MWC user group consisted of eight individuals who relied on MWCs as their primary mode of transportation.

To assess whether trunk control affects vibration transmission, MWC users were divided into two categories based on whether their pathology impacted trunk control. The first category, consisting of users with lower limb amputations, included three male participants (46 ± 12 years

old, 172 ± 7 cm, and 68 ± 18 kg): two transfemoral amputees and one transtibial amputee. The second category, consisting of users with muscle control deficits, included five participants (four men and one woman, aged 52 ± 18 years, measuring 176 ± 12 cm, and weighing 78 ± 20 kg): two with spinal cord injuries (T4-T5 and T11-T12), one with hemiplegia, one with cerebral palsy, and one with T11 to L3 arthrodesis. The experiment was in line with the ethical agreements (IRB00012476-2021-05-02-84).

2.2. Factorial design

2.2.1. Context

A full factorial experiment was carried out for each examined MWC load (able-bodied participants, MWC users with lower limb amputation, MWC users with a muscle control deficit, and a dummy) to elucidate the impact of three experimental factors (speed, type of floor, and MWC) on the assessment of vibration exposure. Hereafter, these three factors will be denoted as f_{Speed} , f_{Floor} , and f_{MWC} .

Each examined experimental factor was characterized by two discrete levels (-1 and 1), representing "low" and "high" levels of induced vibration, respectively. Subsequently, a specific experiment was conducted to assess vibration exposure for every conceivable combination of levels across all factors. The assumption of linearity was taken into account, given the examination of two levels. This complete factorial experiment enables an exploration of the impact of each experimental factor, as well as the interactions among these factors, on vibration exposure.

2.2.2. Propulsion speed

To generate "low" and "high" levels of vibration, two propulsion speeds were selected. Considering the established influence of speed on vibration exposure [19], two speeds were chosen to represent a *slow* ($0.8 \text{ m}\cdot\text{s}^{-1}$) pace and a *fast* ($1.6 \text{ m}\cdot\text{s}^{-1}$) pace within the daily speed range [7]. Regular audio signals and floor markings were used to ensure that the propulsion speed matched the instructions, requiring the experimenter to be at the designated mark at each signal.

2.2.3. Type of floor

To describe floor, a common method is to compute a roughness indicator based on the ISO 8608 standard [20]. However, for the purpose of using the floor characterization as input in a manual wheelchair (MWC) model, we estimated a vibrating index via a preliminary standardized acceleration measurement. This vibrating index was derived from the root-mean-squared (RMS) level of the acceleration norm measured using a rolling calibration object. The calibration object utilized was a chipboard panel measuring $55 \times 22 \times 2 \text{ cm}^3$, to which four skateboard wheels were attached. The panel was then loaded with two evenly distributed 3-kilogram weights to prevent bouncing and was pulled over a 10 m straight line at a speed of $0.8 \text{ m}\cdot\text{s}^{-1}$. The resulting acceleration was measured using an inertial measurement unit (BlueTrident sensors, Vicon, Oxford, UK, 1125 Hz, ± 16 g) fixed to the panel.

Various types of flooring commonly encountered in daily life were measured using the calibration object. Based on the results, floor types inducing the lowest (floor 1) and highest (floor 5) RMS values were identified, providing crucial insights into the potential range of vibration exposure for manual wheelchair (MWC) users. Consequently, floors types 1 and 5 were included in the full factorial design to represent "low" and "high" levels of vibration, respectively. Additionally, floor types 2, 3, and 4 were also tested at a "slow" speed to gather supplementary data for comparison with existing literature and to provide additional information on mid-level vibration scenarios. These mid-level floor types were selected based on their common occurrence in daily life and literature, ensuring comprehensive coverage of vibration characteristics. The resulting vibration indices were $[2.0, 4.2, 8.6, 11.2, 13.0] \text{ m}\cdot\text{s}^{-2}$ for floors 1 to 5, respectively.



Fig. 1. Photographs of the two indoor floor types (Floors 1 and 2: marble tiles and terracotta floor tiles) (a), and of the three outdoor floor types (Floors 3 to 5: graveled concrete slab, light gray asphalt, and coarse gravel concrete) (b) used to study vibration exposure during daily propulsion.

2.2.4. MWC type

To identify how MWC design affects vibration properties, two daily MWCs with the most distinct dynamic behavior were selected. For this purpose, the vibration behavior of various MWC were characterized: the RMS value of the 3D vibration response at the seat was computed during empty propulsion over floor 2 (refer to Fig. 1) at a constant speed of $0.8 \text{ m}\cdot\text{s}^{-1}$ (previously defined as a “slow” speed). Over the tested MWC, a folding standard model (Vermeiren, D200 30°) and a rigid lightweight model (Invacare, Kuschall K-Series) were selected. The resulting vibration indices were determined to be $0.86 \text{ m}\cdot\text{s}^{-2}$ and $1.10 \text{ m}\cdot\text{s}^{-2}$ for the standard and lightweight MWCs, respectively. During measurement both MWC were equipped with a standard 5 cm thick Invacare foam cushion.

2.2.5. Measurement protocol

Each MWC load was propelled by an assistant along a straight 10 m path. The MWC loads included able-bodied participants, MWC users with lower limb amputation, MWC users with a muscle control deficit, and the dummy. The pushing action involved using two fingertips to minimize interaction with the MWC and potential biases in data collection.

For each MWC load, 14 propulsion conditions were explored. The factorial design encompassed eight conditions (with two levels for three factors), and an additional six conditions were examined for comparison with existing literature (one propulsion speed \times three floor types \times two MWC types). Each condition was repeated three times, resulting in a total of 42 trials collected for each MWC load.

2.3. Experimental setup

For each trial, vibration exposure was measured at various locations on the MWC and the participant or the dummy, using 3D accelerometers (BlueTrident sensors, Vicon, Oxford, UK, 1125 Hz, $\pm 16 \text{ g}$). Specifically, three accelerometers were affixed to the MWC: (1) beneath the seat cushion, at the participant’s left ischium; (2) at the hub of the right rear wheel (referred to as the *frame* henceforth); and (3) under the footrest (refer to Fig. 2-a). Additionally, three extra accelerometers were attached to (1) the forehead, (2) the cervical spine (spinous process of the seventh cervical vertebrae, C7), and (3) the lumbar spine (spinous process of the fifth lumbar vertebrae, L5) of the participant (refer to Fig. 2-b). In the case of the dummy, accelerometers were affixed

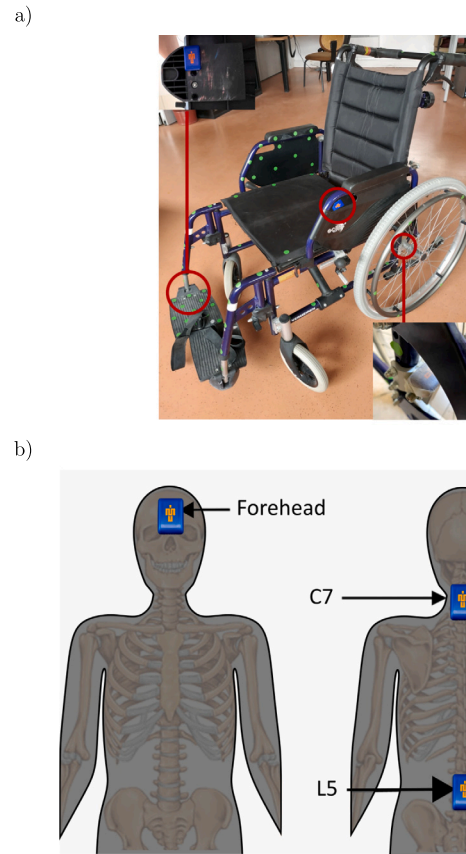


Fig. 2. Location of the 3D accelerometers fixed on the MWC (a) and the user (b).

at 10 cm from its lower end, 15 cm from its upper end, both to match the positions of the C5 and L5 vertebrae on the human body, and on the top of it. It is important to note that the accelerometers were secured using adhesive tape and further secured with adhesive masking tape. In the subsequent sections, the axes x , y , and z denote the anteroposterior, mediolateral, and vertical directions, respectively, with respect to the participant or the dummy.

2.4. Description of the vibration exposure

2.4.1. Temporal domain

To quantify vibration exposure, the energy content in the acceleration signal was assessed using the RMS value. Since MWC vibration exposure is predominantly concentrated along the vertical direction [7], the RMS value of the vertical acceleration was computed as

$$RMS_i = \sqrt{\frac{1}{N} \sum_{n=1}^N |s_n|^2}, \quad (1)$$

where s_n represents a given discrete signal of N points collected at the i -th accelerometer’s location. The discrete signal s_n results from filtering the raw signal through a zero-phase-shift band-pass filter ([4–80] Hz, 4th order Butterworth). The chosen [4–80] Hz frequency range aims to focus on frequencies relevant to human health and comfort [5].

In addition to the physical assessment of vibration exposure, it is crucial to gain insight into the potential health risks for MWC users. To achieve this, the ISO2631 standard [5] recommends calculating the coefficient A_{wz}^i (effective value of the weighted acceleration) at the interfaces between the MWC and the participant or the dummy (i.e., at the seat and the footrest) as

$$A_{wz}^i = \left[\frac{1}{T} \int_0^T a_{wz}^i(t)^2 dt \right]^{\frac{1}{2}}, \quad (2)$$

where $T = 3$ s represents the duration of measurement, and $a_{wz}^i(t)$ is the vertical weighted acceleration at position i (the seat or the footrest). The signal $a_{wz}^i(t)$ is obtained through octave band weighting of the signal s_n as proposed by the ISO2631 standard [5]. The purpose of this weighting process is to emphasize the vibration response at frequencies recognized as deleterious for the seated human body ([4-12] Hz).

Finally, to consider non-vertical solicitations (denoted as x and y), vibration exposure was also computed in the three directions. According to the ISO2631 standard [5], A_w^i was calculated as

$$A_w^i = \sqrt{k_{xy}^2 \times (A_{wx}^i)^2 + k_{xy}^2 \times (A_{wy}^i)^2 + k_z^2 \times (A_{wz}^i)^2}, \quad (3)$$

where i represents either the seat or the footrest, A_{wx}^i and A_{wy}^i are computed as explained for A_{wz}^i , $k_{xy} = 1.4$, and $k_z = 1$.

The subsequent analysis will primarily focus on the parameters RMS_i . A_{wz}^i and A_w^i will be used to discuss and compare the results with the existing literature.

2.4.2. Spectral domain

The impact of vibration exposure on human health is significantly influenced by the frequency characteristics of the stimulus. Therefore, the spectral centroid μ_i of the vibration exposure (i.e., the spectral barycenter) has been calculated to depict the frequency content as

$$\mu_i = \frac{\sum_{n=1}^N f_n |P_n|}{\sum_{n=1}^N |P_n|}, \quad (4)$$

where P_n represents the power spectrum of the signal s_n , f_n is the discrete frequency vector, and i denotes the accelerometer's location. The discrete signal s_n is obtained by filtering the raw signal through a zero-phase-shift band-pass filter ([4-80] Hz, 4th order Butterworth).

2.5. Data analysis

Initially, the non-weighted parameters RMS_i were examined and compared to the weighted parameters A_{wz}^i and A_w^i at each investigated position i . A qualitative analysis was performed, considering that the weighted parameters were solely computed at the MWC/user interface.

Subsequently, for each MWC load (able-bodied participants, MWC users with lower limb amputation, MWC users with a muscle control deficit, and the dummy), mean values of RMS_i and μ_i were calculated across the 42 collected trials. A Wilcoxon non-parametric test was carried out to identify the effect of the MWC load on the vibration exposure measured at the location i . The significance level was set at 0.05.

Finally, an examination was conducted to understand the effect of each experimental factor (speed, type of floor, MWC) and their interactions, using full factorial designs. Using the $RMS_i^{C1..8}$ and $\mu_i^{C1..8}$ values derived from the eight tested combinations $C1..8$ (representing two levels for three factors) the main effects ($a_i^{Speed, Floor, MWC}$, $b_i^{Speed, Floor, MWC}$) and interaction effects ($a_i^{Speed-Floor-MWC}$, $b_i^{Speed-Floor-MWC}$) were estimated based on the equations

$$RMS_i^{C1..8} = a_0 + a_i^{Speed} f_{Speed} + a_i^{Floor} f_{Floor} + a_i^{MWC} f_{MWC} + \quad (5)$$

$$a_i^{Speed-Floor} f_{Speed} f_{Floor} + a_i^{Speed-MWC} f_{Speed} f_{MWC} +$$

$$a_i^{Floor-MWC} f_{Floor} f_{MWC} +$$

$$a_i^{Speed-Floor-MWC} f_{Speed} f_{Floor} f_{MWC},$$

$$\mu_i^{C1..8} = b_0 + b_i^{Speed} f_{Speed} + b_i^{Floor} f_{Floor} + b_i^{MWC} f_{MWC} + \quad (6)$$

$$b_i^{Speed-Floor} f_{Speed} f_{Floor} + b_i^{Speed-MWC} f_{Speed} f_{MWC} +$$

$$b_i^{Floor-MWC} f_{Floor} f_{MWC} +$$

$$b_i^{Speed-Floor-MWC} f_{Speed} f_{Floor} f_{MWC}.$$

The derived main and interaction effects were analyzed as a percentage of variation around the averaged parameters $RMS_i^{C1..8}$ and $\mu_i^{C1..8}$.

3. Results

3.1. MWC load

Fig. 3 illustrates the vibration levels reaching the participant and the MWC structure in the vertical direction. The results were computed across all propulsion conditions (two speeds, five floors, and two MWC types) for the four MWC loads (able-bodied participants, MWC users with lower limb amputation, MWC users with a muscle control deficit, and the dummy). No significant differences were observed among the investigated loads. However, when examining results at the participant's level, the RMS value estimated for the dummy showed slight variations compared to the three conditions involving a participant.

Fig. 4 delves into the spectral centroid of the vibration exposure. Interestingly, the results indicated that the dummy induced a lower spectral centroid than the three other loads at the lumbar spine (12 ± 3 Hz and 26 ± 6 Hz, respectively) and the MWC seat (14 ± 5 Hz and 21 ± 6 Hz, respectively).

Based on these findings, the subsequent results will present how the propulsion conditions influence the vibration levels reaching the MWC user and the MWC structure without distinguishing the type of MWC load.

3.2. Vibration exposure through the MWC/user dyad

Table 1 provides an overview of the vibration exposure quantification using three estimators (RMS_i , A_{wz}^i , and A_w^i) across all tested experimental conditions for various MWC loads. Considering all axes, A_w^i was approximately twice as high as A_{wz}^i . Specifically focusing on the vertical axis, the weighted coefficient A_{wz}^i consistently showed lower values than RMS_i . This difference was emphasized at the footrest: $RMS_{seat} = 1.8 A_{wz}^{seat}$ while $RMS_{footrest} = 3.3 A_{wz}^{footrest}$. Despite the variations between the three estimators, Table 1 revealed a similar trend in their evolution concerning the experimental conditions.

The vibration levels tended to decrease from the floor type to the MWC, ranging from 3.1 ± 5.1 m.s⁻² at the footrest to 0.8 ± 0.8 m.s⁻² at the seat. In terms of the MWC user, although the vibration exposure decreased from the lumbar spine (1.2 ± 1.3 m.s⁻²) to the cervical spine (0.5 ± 0.3 m.s⁻²), it remained relatively constant up to the forehead (0.5 ± 0.2 m.s⁻²). Moreover, while the vibration levels were lower on the user's cervical spine and forehead compared to the MWC, the RMS values were estimated to be higher at the lumbar spine (1.2 ± 1.3 m.s⁻²) than at the MWC seat (0.8 ± 0.8 m.s⁻²).

Table 2 displays the spectral centroid of the vibration exposure for all MWC loads under various experimental conditions. Similar to the vibration levels, the spectral centroid decreased from the floor type to the MWC user's forehead (from 30 ± 11 Hz to 12 ± 3 Hz). Again, while the spectral centroid decreased from the lumbar spine (26 ± 5 Hz) to the cervical spine (12 ± 2 Hz), it remained relatively constant up to the forehead (12 ± 3 Hz). Furthermore, the spectral centroid measured at the lumbar spine (26 ± 5 Hz) was higher than at the MWC seat (21 ± 5 Hz), echoing the trends observed in the RMS values.

3.3. Speed, type of floor and MWC

Fig. 5 presents the interactions and main effects of each experimental factor (speed, type of floor, MWC) on the vibration exposure (RMS_i and μ_i). The main effects (a_i^{Speed} , a_i^{Floor} , a_i^{MWC}) and the interaction effects ($a_i^{Speed-Floor}$, $a_i^{Speed-MWC}$, $a_i^{Floor-MWC}$, $a_i^{Speed-Floor-MWC}$) are depicted as a percentage of variation around the average value of RMS_i or μ_i .

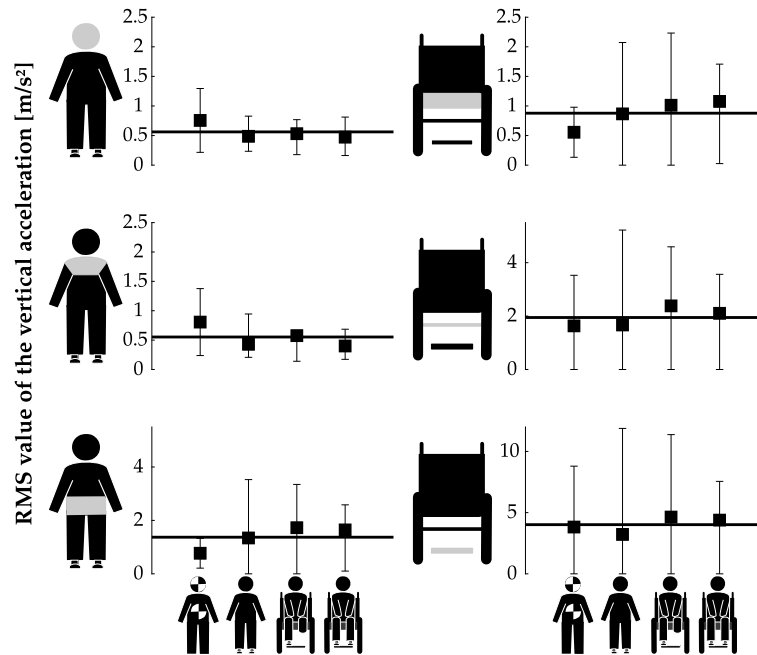


Fig. 3. Mean and standard deviation of the RMS_i value on different positions i of the participant (first column: from top to bottom; forehead, cervical spine, and lumbar spine) and MWC (second column: from top to bottom; seat, frame, and footrest). RMS_i values were computed over all the conditions of propulsion (two speeds, five floors and two MWC) for the four MWC loads (able-bodied participants, MWC users with lower limb amputation, MWC users with a muscle control deficit, and a dummy). The horizontal black lines present the average value over all the tested conditions.

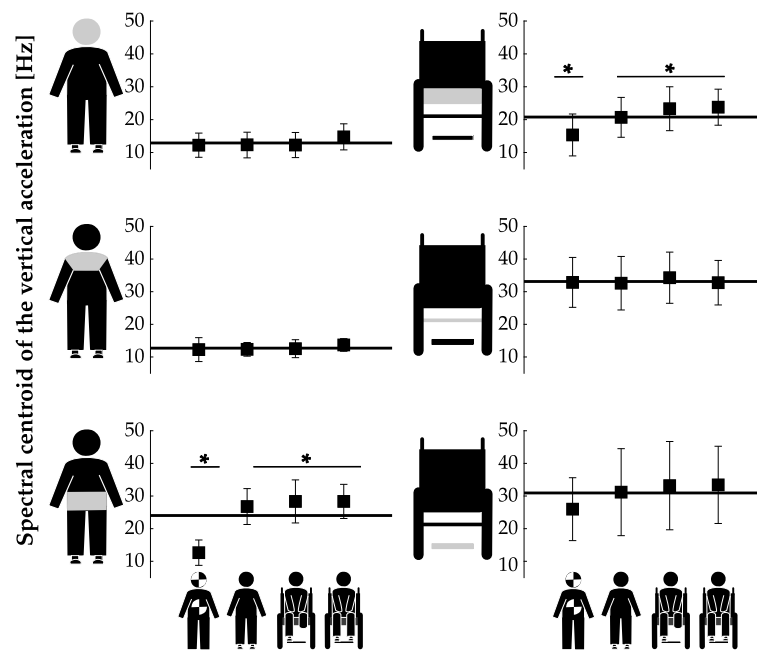


Fig. 4. Mean and standard deviation of the spectral centroid μ_i on different positions i of the participant (first column: from top to bottom; forehead, cervical spine, and lumbar spine) and MWC (second column: from top to bottom; seat, frame, and footrest). μ_i values were computed over all the conditions of propulsion (two speeds, five floors and two MWC) for the four MWC loads (able-bodied participants, MWC users with lower limb amputation, MWC users with a muscle control deficit, and a dummy). The horizontal black lines present the average value over all the tested conditions. * indicates a significant difference ($p < 0.05$) between two MWC loads.

3.3.1. Amount of vibration

Analyzing the impact of the three factors and their interactions on the RMS value at the MWC footrest, all factors and their interactions played a significant role, showing a minimum variation of 50% around the averaged value. Notably, a decrease in the main and interaction effects was observed, particularly as one moved from the lumbar spine to the user's forehead. Specifically, the main effects of speed and floor

type remained prominent, but their interaction effects decreased beyond the lumbar spine. Moreover, the main effect of the MWC, along with all interaction effects involving the MWC, significantly diminished from the user's forehead.

Table 1 provides insights into how each factor influenced the RMS value. As anticipated, higher speed resulted in an increased RMS value. Additionally, on all investigated locations, the floor with the highest

Table 1

Quantification of vibration exposure through the average and standard deviation of RMS_i (RMS value at the i th accelerometer's location along the vertical axis), A_{wz}^i (weighted estimator along the vertical axis), and A_w^i (weighted estimator along the three axes), computed for all the MWC loads, with respect to the tested experimental conditions.























Floor															
Speed [m.s ⁻¹]		0.8		1.6		0.8		0.8		0.8		1.6			
MWC															
RMS [m.s ⁻²]	Forehead	0.2 (0.0)	0.2 (0.1)	0.3 (0.1)	0.4 (0.1)	0.3 (0.1)	0.3 (0.1)	0.4 (0.1)	0.4 (0.1)	0.6 (0.2)	0.6 (0.1)	0.7 (0.2)	0.5 (0.1)	1.0 (0.3)	0.9 (0.2)
	C7	0.1 (0.0)	0.2 (0.1)	0.3 (0.1)	0.4 (0.2)	0.3 (0.1)	0.3 (0.1)	0.4 (0.1)	0.4 (0.1)	0.6 (0.2)	0.6 (0.2)	0.7 (0.2)	0.6 (0.2)	0.9 (0.2)	1.1 (0.4)
	L5	0.2 (0.0)	0.1 (0.0)	0.6 (0.2)	0.7 (0.2)	0.5 (0.1)	0.4 (0.1)	0.9 (0.2)	0.6 (0.1)	1.3 (0.3)	0.9 (0.1)	2.0 (0.6)	1.1 (0.2)	4.2 (1.3)	3.9 (1.1)
	Seat	0.1 (0.0)	0.1 (0.0)	0.5 (0.2)	0.4 (0.1)	0.4 (0.0)	0.2 (0.0)	0.7 (0.1)	0.3 (0.1)	1.0 (0.2)	0.5 (0.1)	1.4 (0.4)	0.5 (0.1)	3.0 (0.7)	1.7 (0.4)
	Frame	0.2 (0.0)	0.2 (0.0)	1.1 (0.5)	0.8 (0.2)	0.5 (0.1)	0.4 (0.1)	1.1 (0.3)	0.7 (0.1)	1.5 (0.3)	1.0 (0.1)	2.0 (0.5)	1.2 (0.2)	7.6 (1.7)	4.4 (0.9)
	Footrest	0.5 (0.1)	0.3 (0.1)	2.5 (0.8)	0.9 (0.3)	1.0 (0.2)	0.5 (0.1)	2.5 (0.5)	0.8 (0.2)	3.2 (0.6)	1.3 (0.3)	4.2 (0.8)	1.3 (0.3)	20.1 (5.0)	4.9 (1.5)
A_{wz} [m.s ⁻²]	Seat	0.1 (0.0)	0.1 (0.0)	0.4 (0.1)	0.3 (0.1)	0.3 (0.0)	0.2 (0.0)	0.4 (0.1)	0.3 (0.0)	0.7 (0.1)	0.4 (0.1)	1.0 (0.2)	0.4 (0.1)	1.5 (0.3)	1.0 (0.2)
	Footrest	0.2 (0.0)	0.3 (0.1)	0.9 (0.3)	0.7 (0.2)	0.6 (0.1)	0.5 (0.1)	1.1 (0.2)	0.6 (0.2)	1.6 (0.2)	1.1 (0.2)	2.1 (0.3)	1.0 (0.2)	6.2 (1.5)	2.4 (0.7)
A_w [m.s ⁻²]	Seat	0.2 (0.0)	0.2 (0.0)	0.8 (0.3)	0.5 (0.1)	0.5 (0.1)	0.4 (0.1)	0.9 (0.2)	0.6 (0.1)	1.4 (0.3)	0.9 (0.1)	2.0 (0.3)	1.0 (0.1)	3.9 (0.9)	2.2 (0.4)
	Footrest	0.7 (0.1)	0.6 (0.2)	2.5 (1.0)	1.7 (0.4)	1.7 (0.2)	1.1 (0.2)	3.5 (0.5)	1.7 (0.2)	5.0 (0.8)	2.6 (0.3)	7.5 (1.4)	2.8 (0.3)	15.9 (3.1)	6.6 (1.0)

Table 2

Spectral centroid of the vibration exposure computed for all the MWC loads, with respect to the tested experimental conditions.

Floor															
Speed [m.s ⁻¹]		0.8		1.6		0.8		0.8		0.8		1.6			
MWC															
μ [Hz]	Forehead	11.4 (2.7)	8.3 (1.6)	11.5 (3.1)	9.1 (1.4)	11.9 (2.6)	8.8 (1.2)	15.1 (4.0)	9.8 (1.7)	12.4 (2.9)	8.9 (1.4)	15.8 (3.1)	10.3 (1.5)	19.7 (5.0)	14.3 (3.1)
	C7	11.4 (1.2)	9.5 (1.7)	11.6 (1.7)	11.6 (2.3)	12.3 (1.4)	10.4 (1.4)	14.3 (2.1)	11.7 (2.1)	12.6 (1.7)	10.6 (1.8)	14.5 (1.7)	12.0 (1.7)	15.4 (3.9)	14.9 (2.7)
	L5	26.2 (4.3)	23.1 (4.7)	30.2 (5.5)	24.7 (3.3)	22.2 (3.6)	22.1 (3.1)	27.1 (4.1)	25.7 (3.0)	24.6 (3.7)	22.7 (3.1)	23.8 (3.2)	22.6 (2.7)	38.4 (5.7)	30.6 (2.7)
	Seat	19.6 (3.1)	15.0 (2.9)	24.5 (4.1)	20.3 (2.4)	18.0 (2.4)	16.1 (1.8)	23.1 (2.9)	20.2 (1.9)	20.4 (2.9)	17.4 (2.0)	20.4 (2.4)	18.8 (2.0)	33.7 (3.0)	28.4 (3.0)
	Frame	38.6 (3.1)	30.2 (3.9)	42.3 (3.2)	25.1 (1.5)	31.6 (2.9)	24.9 (1.1)	38.2 (3.2)	28.4 (1.0)	35.3 (3.2)	24.8 (0.9)	32.0 (2.6)	23.8 (1.2)	45.7 (2.1)	32.2 (1.5)
	Footrest	35.0 (3.7)	13.8 (2.9)	45.5 (3.7)	21.9 (3.2)	29.7 (2.4)	14.7 (2.9)	37.6 (2.6)	24.7 (4.3)	35.3 (2.7)	19.6 (3.8)	33.4 (2.7)	21.4 (3.7)	51.9 (3.9)	37.8 (5.3)

vibrating index (floor 5) induced an elevated RMS value. Lastly, the MWC type, while considered the least influential of the three factors, notably affected the vibration exposure measured on various MWC parts more than on the MWC user. Results suggest that the lightweight MWC led to a higher RMS value compared to a standard MWC.

3.3.2. Spectral centroid

The variations in the spectral centroid were observed to be lower than the variations in the RMS values, with spectral centroid variations not exceeding 40% (see Fig. 6). The interaction between the three factors had a comparatively lesser impact on the spectral content than on the amount of vibration.

Table 2 provides insights into how each factor influenced the spectral centroid. The speed predominantly affected the signals measured at the

MWC footrest, seat, and lumbar spine, with a higher speed leading to a higher spectral centroid. Interestingly, the type of floor specifically affected the frequency content at the MWC seat, C7, and the forehead, with floor 5 inducing a higher frequency content than floor 1. Finally, a lightweight MWC implied a higher spectral centroid than a standard MWC.

4. Discussion

4.1. Vibration transmission through the MWC/User dyad

The study revealed that the characteristics of vibrations, encompassing both vibration exposure and spectral content, are markedly influenced by the measurement position, whether on the MWC or the user.

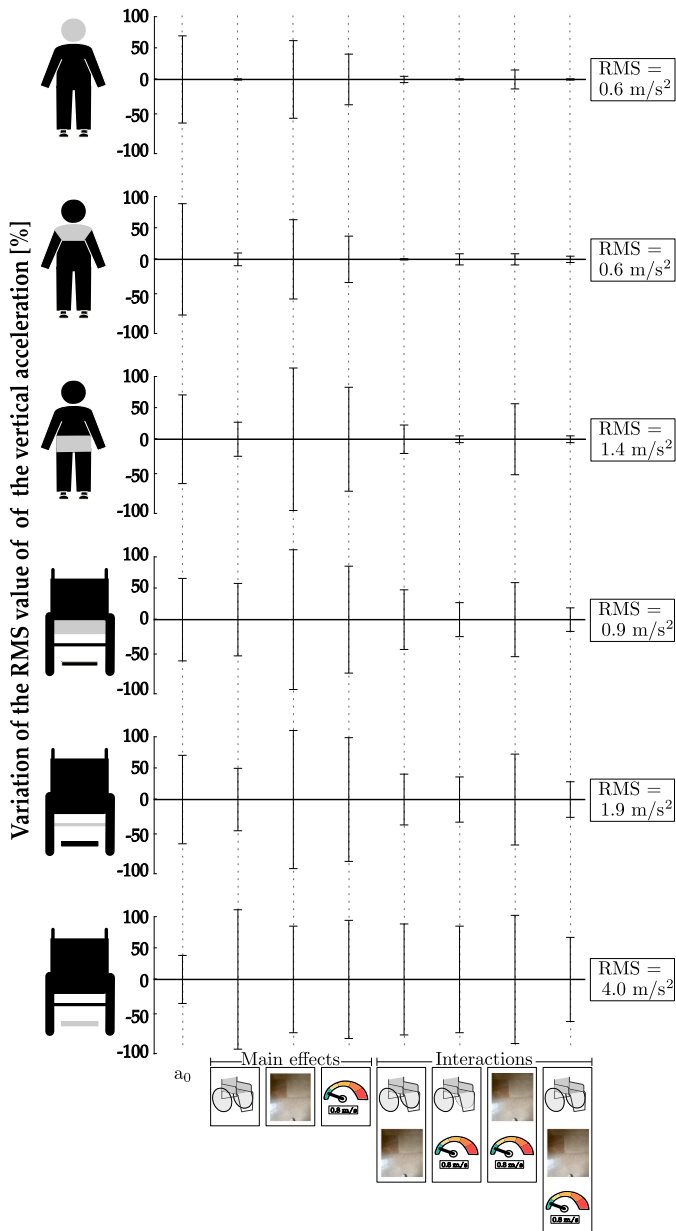


Fig. 5. Percentage of variation of the RMS values $RMS_i^{C1..8}$ around the averaged values on different positions of the participant and MWC (from top to bottom; forehead, cervical spine, and lumbar spine and MWC seat, frame, and footrest) according to the different factors of influence. MWC, Floors, speed and their interactions. The framed value shows the averaged parameters.

Consistent with existing literature, the vibrations exhibited the highest amplitude at the footrest and the lowest at the user's cervical spine and forehead, where the estimations were comparable [7]. Notably, the higher vibration level at the footrest compared to the MWC frame is likely attributable to the free end of the footrest or to the front wheels, which exhibit less vibration dampening than the rear wheels due to their rigidity.

Our findings at the seat cushion ($A_{wz} \approx 0.3 \pm 0.0 \text{ m.s}^{-2}$) concurred with those of Misch and Sprigle [14] ($A_{wz} \approx 0.2 \text{ m.s}^{-2}$) under similar conditions (a lightweight MWC seat on terracotta floor tiles). However, a discrepancy arises from our measurements being conducted under the seat cushion, similar to Garcia-Mendez et al. [13], while Misch and Sprigle placed their sensors at the top surface of the seat cushion. Despite the hypothesis proposed by Misch and Sprigle suggesting that the measurement location explains the observed differences, the present

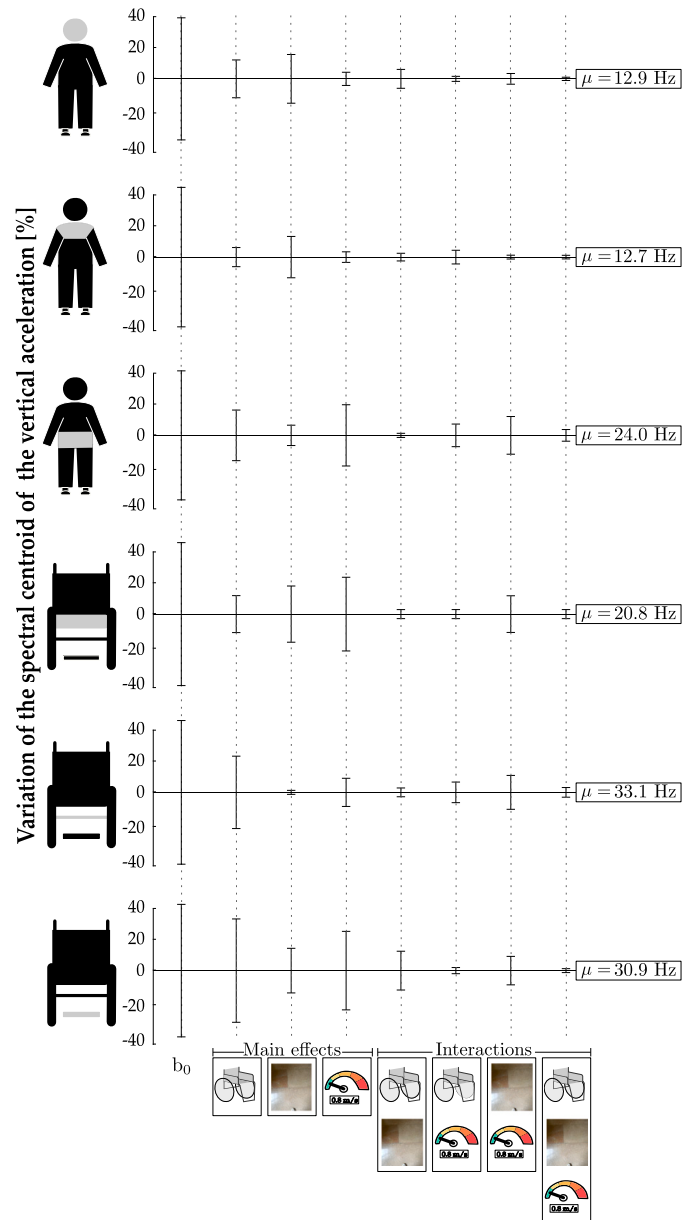


Fig. 6. Percentage of variation of the spectral centroid $\mu_i^{C1..8}$ around the averaged values on different positions of the participant and MWC (from top to bottom; forehead, cervical spine, and lumbar spine and MWC seat, frame, and footrest) according to the different factors of influence. MWC, Floors, speed and their interactions. The framed value shows the averaged parameters.

results indicate that the type of loading (human versus a dummy or a robot) also highly contributes to this difference. The observed difference between the two studies is also likely induced by the multi-axis measurements undertaken by Garcia-Mendez et al., while Misch and Sprigle focused solely on the vertical axis. This study addressed vibration exposure under all three axes, revealing that A_w is twice as high as A_{wz} .

Surprisingly, the vibration level at the lumbar spine was measured higher than at the seat. Given that the spine exhibits resonance between 4 and 20 Hz [21], vibrations entering at these low frequencies are likely to be amplified. Additionally, the contribution of the MWC backrest could be considered. A higher spectral centroid was indeed observed at the lumbar spine than at the seat. As the human lower body functions as a low-pass filter [21], the heightened vibration content could only be transmitted through the MWC. To accurately estimate MWC users'

vibration exposure, it is therefore crucial to account for the backrest contribution.

Finally, the vibration levels measured at the cervical spine and the forehead were similar, with almost no impact from the three tested factors (speed, type of floor, MWC). Regardless of the properties of the entering vibrations, the trunk is assumed to play a role in maintaining vibration levels at the upper body. Vibrations at the forehead and the cervical spine were predominantly concentrated at frequencies below 20 Hz. This aligns with previous findings on vibration transmission in seated humans [22,23]. These results suggest that the human body possesses the capability to amplify input vibrations, especially below 20 Hz. Beyond this threshold, the human body acts as a low-pass filter, progressively damping the input vibrations.

4.2. MWC load

While conducting investigations using an empty MWC is inherently not representative of a realistic situation, involving actual MWC users poses significant challenges. Focusing on understanding the MWC response to vibration, a prior study emphasized that using a dummy leads to a higher vibration level and lower frequency content [17]. To delve deeper into this issue, the current study compared various MWC loads to determine whether a dummy or an able-bodied participant could offer a viable solution. Despite no significant difference being observed in the vibration level, estimations made with the dummy were readily distinguishable from those made with a human body, particularly at the MWC seat and overall the participant. The spectral centroid was significantly lower at the MWC seat and the participant's lumbar spine for a dummy compared to a human body. Nevertheless, estimations for the upper body were similar across all conditions involving a human body. The results can be attributed to the simplistic nature of the dummy used. This dummy primarily represented the distribution of human mass but failed to simulate the damping and stiffness characteristics of a real human. Using a more advanced dummy could yield results that better mirror those observed in actual MWC users. Before using a dummy to estimate vibration characteristics, it is crucial to ensure that the dummy produces results that are representative of real MWC users.

These differences might be attributed to the dummy's feet, simulated by two 3 kg isolated masses placed on the footrest with no connection to the dummy's body [5]. Consequently, the dummy's damping properties are likely reduced compared to a human body. While the dummy used in the present study may be representative of the MWC/User dyad at the forehead and cervical spine levels, our recommendation is to investigate the MWC loaded by an able-bodied participant or a MWC user. However, the 2012 version of the dummy (ISO-7176-11:2012) recommends having the feet tied to the body. Therefore, it would be relevant to test this configuration as well to validate this hypothesis.

4.3. Factors effect

Consistent with previous findings highlighting the pivotal role of propulsion speed and floor characteristics in the vibration exposure of MWC users [17], this study reaffirms that escalating the propulsion speed and roughness of the ground surface correlates with heightened vibration levels [24,16,25]. Moreover, this influence extends to the MWC user, particularly affecting the vibration level at the user's forehead, predominantly governed by speed and floor type.

Beyond vibration levels, alterations in propulsion speed or floor type directly impact the spectral content, shifting it towards higher frequencies with increased speed or floor roughness. Furthermore, higher speed predominantly affects vibrations at the MWC seat and lumbar spine, while increased floor roughness has a more pronounced impact on the cervical spine and forehead. Given that the frequency range affecting human health and comfort is [4 - 80] Hz, with a specific risk between 4 and 12 Hz for the seated human body [5], controlling both propelling speed and floor type during MWC testing becomes crucial.

The lightweight MWC induced higher vibration exposure than a standard MWC, likely due to mass differences and their foldable/rigid properties. Interestingly, MWC variations only influenced vibrations measured on the MWC itself. Despite distinct modal properties of each MWC [12] and varied properties of entering vibrations, the human body's response to vibration remained similar up to 80 Hz, irrespective of the MWC. This result prompts the question of whether and how the human body regulates vibration. Several processes have been suggested to be involved in cushioning shocks and vibration during walking and running: passive physiological attenuation driven by the mechanical behavior of anatomical elements [26], and active attenuation governed by changes in the orientation of body segments [27,28] or muscle activations [29,30]. If future work evidence that active control is a key-factor in MWC propulsion, increased fatigue could be expected, posing a paradox between promoting physical activity and mitigating the risks of vibration exposure. As highlighted in sports activities [17], redefining action and limit values and establishing dedicated guidelines for MWC design may be necessary.

4.4. Vibration exposure and MWC health risk

One of the primary challenges in evaluating the connection between vibration exposure and health risks involves establishing recommended levels and durations of exposure limits. Guidelines have been instituted to protect the health of workers exposed to whole-body vibration [4,5]. These guidelines specify *action* and *limit* values set at 0.5 m/s² and 1.15 m/s² over an eight-hour period. Exceeding the *action value* triggers the implementation of a preventive plan, while surpassing the *limit value* necessitates immediate intervention from the employer to mitigate vibration exposure.

Assuming the applicability of these limit values to manual wheelchair users, it becomes possible to calculate the propulsion time limit associated with the tested propulsion conditions. For example, the indoor floor generating the least vibration among the tested floors would require exposure for over 24 hours to exceed the permitted threshold by the standard. In contrast, the floor causing the highest vibration levels, which is the closest to the floor commonly encountered in streets, would breach the authorized threshold in just 2 hours and 30 minutes. In their study, Misch and Sprigle [14] observed a higher exposure time limit due to the prevalence of indoor flooring surfaces or smooth surfaces in the tested floors. This difference in results between Misch and Sprigle's study [14] and that of Garcia-Mendez [13] could, in part, be attributed to the floor condition. These results emphasize the crucial importance of floor quality. A real urban floor is likely to be in poorer condition than the floors examined in the current study. This suggests that the minimum exposure time would decrease and approach the average daily locomotion time of MWC users, estimated at approximately 1 hour [31].

However, for manual wheelchair users, the daily exposure duration is notably variable, especially considering engagement in physical activities and sports. Sports activities involve propulsion at higher speeds, consequently leading to increased exposure to vibrations [32]. Therefore, enhancing manual wheelchair design to improve vibration transmission could prove beneficial in mitigating certain health risks and reducing fatigue associated with wheelchair use. Additionally, such improvements might enhance sports performance, enabling athletes to train more effectively and delaying the onset of fatigue induced by vibration exposure [33].

The type of floor emerged as the most influential factor affecting the vibration exposure encountered by manual wheelchair (MWC) users. It would be valuable, therefore, to identify the predominant surface types encountered by MWC users and the duration of propulsion on each. While some studies have characterized aspects such as the number of trips, duration, and speed of MWC use [31], they often lack information about the nature of the terrain. Since all tested floors exhibit equivalent frequency content, the impact on user risks is solely determined by the

floor-generated level. Therefore, having knowledge of surface types and their respective durations would facilitate a straightforward estimation of vibration exposure. This is particularly relevant, given that certain studies have successfully linked ground roughness to vibratory levels [16], a correlation also observed in the present study. Regardless of the MWC type and measurement position, a linear relationship was noted between the RMS level obtained from the calibration rolling object on the ground and the levels measured at different MWC and user positions.

5. Limitations and perspectives

The main limitation of this study lies in the generalization of results obtained from measurements on a small number of participants. Given the variability in the pathologies of MWC users, it would be interesting to investigate how their specificities affect the propagation of vibrations. It is also known that posture can affect vibration transmission [34]. In this study, the participants' posture was standardized but not controlled, with all participants using identical MWC settings. Such research would allow for adjusting the threshold values proposed by standards to accommodate these specific populations. In this study, the users were propelled by a third person. When users propel themselves, the contact with the handrim, the backrest and the varying postures they adopt could influence the nature of the vibration [35]. Indeed, the third person induces a mechanical coupling with the MWC and eventually additional damping phenomena depending on his grip. Moreover, the mass distribution on the MWC is also directly driven by the propelling condition, affecting the vibration exposure content [17]. Future studies may thus be interested in analyzing how posture and self-propulsion affect vibration transmission. Finally, additional studies are needed to enhance our understanding of how vibration spreads and is absorbed by both the MWC and the user, particularly in light of various pathologies.

6. Conclusion

This study aimed to establish guidelines for standardizing methods to assess vibration exposure in manual wheelchair users. Through a comprehensive full factorial experiment involving various propulsion conditions (two speeds, five floors, and two MWC) and four MWC loads (an ISO-7176–11:1992 dummy, able-bodied participants, lower limb amputee MWC users, and muscle-injured MWC users), outcomes emphasized the crucial role of controlling floor type and speed. These parameters significantly impact the vibrations reaching the users' cervical spine and forehead. While the type of MWC showed less influence, it still plays a role in determining the vibration content at the MWC/User interfaces. Additionally, enrolling an able-bodied participant is more suitable than using a dummy when considering how to load the MWC, providing a more realistic approximation of the MWC/user dyad response to vibration.

Ethical approval

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Declaration of competing interest

None declared.

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