Vestibular contributions to metabolic cost in older and younger adults during treadmill and overground walking

Master's Thesis

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15-10-2025

Vestibular contributions to metabolic cost in older and younger adults during treadmill and overground walking

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In partial fulfilment of the requirements for the degree of:

Master of Science in Biomedical Engineering
Track: Neuromusculoskeletal Biomechanics

at the Delft University of Technology,
to be defended publicly Wednesday, October 22, 2025.

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Abstract

The metabolic cost of walking reflects gait efficiency and is influenced by biomechanical factors as walking speed. While most walking research uses treadmills, older adults are found to exhibit a greater elevation in the cost of walking on the treadmill compared to overground walking than younger adults. A possible cause for this elevation could be the higher stability demands in older adults during treadmill walking, possibly influenced by a higher vestibular contribution. This study investigated whether increased vestibular demands for balance control contribute to this elevated cost in older adults. Ten younger (mean age 26.4 years) and ten older adults (mean age 68.6 years) completed 5-minute treadmill and overground walking trials at preferred and slow fixed speeds. Metabolic cost was measured, and vestibular contributions to balance were assessed via electrical vestibular stimulation, which induced virtual movements and evoked balance correcting responses measured by inertial sensors on the back and ankles. Treadmill walking increased the cost of walking significantly by 15-23% compared to overground walking, with no significant age effect. Vestibular stimulation increased metabolic cost significantly in both overground and treadmill walking and age groups. Assessment of the vestibular contributions to kinematic measures revealed a significant increase in vestibular contribution to balance at slower walking speeds, but no significant effect of age and no large effect of treadmill or overground. Indicating that the measured participants cannot conclude that the elevation of cost in treadmill walking in older adults is due to the vestibular contribution to balance.

Acknowledgments

This thesis would not have been possible without the support and guidance of many wonderful people. First, I would like to thank Patrick Forbes and Matto Leeuwis, not only for supervising this thesis, but for the entire year of working together, from my Erasmus MC internship to this final project. Thank you for your mentorship, time, and for helping me learn and grow. Thank you, Eline van der Kruk, for trusting me to take the lead in my project, for your sharp questions, and for reminding me to believe in myself.

I'm grateful to Matto, Patrick, Hill and Wessel for assisting during the experiments. To Martijn for making the pacing cart run smoothly, and Liam Foulger for helping me understand the code and set-up. I also want to thank all participants for their time and contribution. Wessel, thank you for your help, your encouragement, for cheering me on, wiping my stress-tears, and celebrating every success together. I love you. Fabiën, I want to thank you for your support, for looking out for me and for the evenings working together in your room. Very thankful for our friendship. I want to thank my dear girlfriends that I met on the first day of this Master's, for the coffee dates and the looking out for each other from the first day of this master's until the last and hopefully far beyond. To my parents and family, thank you for your support and faith in me. And to the Blijdorp family, especially Wim and Cato, thank you for your love and for helping with participant recruitment. I would not have been able to find all 20 participants without your help. Thank you to the BODIES lab for their input and discussion to improve this project. I want to thank the Forbes lab and everyone on the 14^{th} floor in Erasmus MC Rotterdam of the Neuroscience department for giving me a home away from home. It truly felt like a family, from the walks in the park, pupquiz calendar questions, getting dirty chai lattes and eating Koekela cakes when there was something to celebrate.

I am proud of finishing my Master's, but it is bittersweet as I won't be seeing the people involved daily anymore. I will cherish these last two years and everyone I've met along this journey. Thank you.

1 Introduction

The metabolic cost of walking, defined as the energy expended per unit distance walked, is a fundamental parameter in assessing the efficiency of human locomotion (1, 2). Since walking is a primary component of the daily energy expenditure (3), it is important to understand the factors that influence the cost of walking. Individuals naturally adjust their gait parameters, such as walking speed, step width, and cadence, to minimise the cost of walking (4-6). Most human walking studies have been conducted on treadmills, as they offer a controlled, repeatable environment and facilitate the use of precision tools such as motion capture and instrumented force plates (7). However, concerns remain regarding how well treadmill walking represents overground locomotion. It has been observed that there are differences in ground reaction forces, spatiotemporal parameters, and joint kinematics between treadmill and overground walking (8-10). This raises concerns about the generalizability of treadmill-based findings to overground walking. To this point, a consistent finding in the literature is that older adults exhibit a higher cost of walking on a treadmill compared to overground walking (11-14) and a higher cost in treadmill walking compared to younger adults (13, 15-18). This difference occurs despite the similarity in the cost of walking in younger and older adults during treadmill walking (13). The underlying mechanisms contributing to the elevated cost of walking found in older adults during treadmill walking remain unknown and need further understanding.

A possible contributor to this phenomenon could be the decline in systems critical for sensing movement and maintaining postural stability. Specifically, the vestibular system, located in the inner ear, is fundamental as it encodes head movements in space and drives whole-body balance responses (19, 20). Its contribution is especially crucial in dynamic environments where postural stability is challenged (21, 22). Given that treadmill walking inherently alters sensory input (such as vision) and spatial constraints compared to overground walking (23), it is plausible that older adults experience greater vestibular-related postural challenges in this environment. This challenging context, coupled with age-related lateral balance issues (24), forces older adults to adopt compensatory strategies to ensure steadiness. The mechanism where instability leads to an increased cost of walking is supported by Brown et al., who observed that instability contributes to a higher cost of walking and a slower walking speed in older adults (25). Since humans are known to prioritise stability over walking efficiency depending on the environmental risk (26), older adults likely execute more frequent or more intense stabilisation efforts when walking on a treadmill compared to walking overground and to younger adults.

A common method for probing vestibular function is electrical vestibular stimulation (EVS), which applies mild currents to the mastoid process. This activates all vestibular afferents (27, 28), modulates ongoing vestibular activity, and evokes corrective balance responses (29-31). By integrating this into a fully portable system with inertial measurement units, it is possible to characterise vestibular-evoked responses across the stride cycle as done by Foulger et al. (32) during overground and treadmill walking without having to deal with static equipment. This method captures the phasic modulation of vestibular-evoked responses across the stride cycle, with the highest correlation observed during the stance phase or double support phase of walking (19, 21, 33-38) and a decrease in vestibular control of balance with increasing locomotor cadence (30) and speed (31, 36).

Despite extensive research on the cost of walking, ageing, walking conditions (treadmill vs. overground), and the vestibular system separately, no comprehensive causal relationship has yet been established between the combined effects of age, walking condition, and the vestibular system's contribution to balance on the metabolic cost of walking. The existing literature often focuses on these factors in isolation or in limited combinations, which means that the current evidence does not allow us to conclude that the underlying factors contributing to the cost of walking differences on the treadmill are caused by the difference in reliance on vestibular signals.

Building on this gap, this master's thesis aims to investigate the combined effects of age, walking condition, and vestibular contributions to balance on the cost of walking to understand the elevation of the cost of walking in older adults during treadmill walking. Younger and older adults were compared during both treadmill and overground walking, while the vestibular system's role was assessed using electrical vestibular stimulation and inertial measurement units measuring body responses, following the approach of Foulger et al. (32) and measuring their metabolic rate. It was hypothesised that older adults would show a higher cost of walking on the treadmill, alongside a stronger vestibular contribution to balance, as indicated by a greater peak coherence between the vestibular stimulation and body responses compared to younger adults. Indicating the underlying cause for the elevation of the cost of treadmill walking in older adults.

2 Methods

2.1 Participants

A total of 20 adult participants were enrolled in this study, including 10 older adults (4 females and 6 males, mean age 68.6 ± 3.57 years, height 177.1±5.9 cm, weight 76.7±8.2 kg) and 10 younger adults (4 females, 6 males, mean age 26.4 ± 2.55 years, height 182.1±9.2 cm, mass 78.9±16.4 kg). The older adults had to be 65 years or older, and the younger adults 30 or younger. The sample size per group was matched to a study by Das Gupta et al., where the metabolic cost of walking in older and younger adults was assessed in both treadmill and overground walking (13). Exclusion criteria were chronic heart diseases, diabetes, prior lower limb surgeries or prosthesis, neuromuscular injuries, recent falls within the past 6 months, or participation in specialised strength or endurance training. The inclusion criteria required participants to be fit and physically active, carrying out their normal day-to-day activities without assistance. Standard anthropometrics were measured, like body mass, height, and ankle height from the lateral malleolus. All participants signed a written consent form. The ethical review committee of the faculty of Mechanical Engineering of the TU Delft approved the project.

2.2 Setup

Before the start of the trials, each participant was equipped with a portable system including a stimulation box for the electrical vestibular stimulation and the IMU's placed on the forehead, lower back and both ankles, as discussed in detail below. Additionally, a COSMED K5 mask was placed covering the mouth and nose to measure rates of oxygen (O_2) consumption and carbon dioxide (CO_2) production.

2.2.1 Portable system

To simultaneously measure movement kinematics and deliver stochastic electrical vestibular stimulation (EVS), a portable experimental system was developed (Figure 1) using a reconfigurable I/O device (MyRIO 1900, National Instruments). The system was controlled using a laptop (Dell, Vostro 5468) running LabVIEW 2019b (National Instruments) and was based upon the experimental set-up used by Foulger et. al. (32). The stochastic EVS signal was sent from the myRIO device to the stimulator box (STIMSOLA, Biopac Systems Inc., CA, United States) at a rate of 200 Hz to deliver the electrical stimulus to participants (see Electrical Vestibular Simulation). Both the myRIO-1900 and the stimulator were powered with a 12 V battery (TalentCell, China) weighing approximately 1.5 kg. All the equipment, the myRIO-1900, stimulator box, and battery, were placed in a crossbody bag (UNIQLO, Yamaguchi, Japan) worn by the participant on their chest (see Appendix A, Figure 6A for the organisation of the bag).

Electrical vestibular stimulation

A Binaural bipolar electrical vestibular stimulation (EVS) was applied to the participants' mastoid process via rubber electrodes (9 cm^2) coated with conductive gel (Spectra Gel 360, Parker) that were secured to the head with tape (3M Durapore) and an elastic net bandage (Elastofix, C-25m stretched). The electrical vestibular stimuli were delivered as a stochastic signal (stochastic EVS; 0-20 Hz, amplitude peak ± 4.5 mA, root mean square 1.25 mA) created using LabView 2019b (National Instruments) and delivered with a constant current isolated stimulator (STIMSOLA, Biopac Systems Inc., CA, United States). This type of signal was chosen to replicate the study that inspired this experiment (32) and the prior studies that have successfully evoked vestibular balance responses during walking (19, 35).

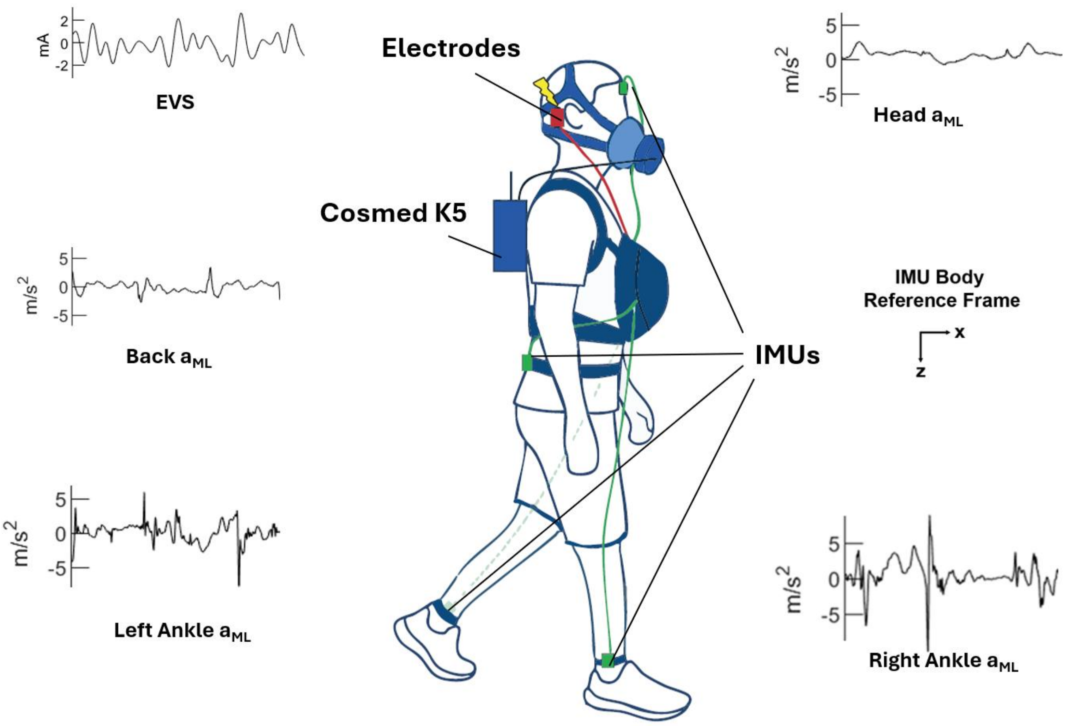

Figure 1: Experimental set-up with the participant. Behind the ears on the mastoid process are the electrodes for the electrical vestibular stimulation (EVS) placed (red). The IMU's are placed on the forehead, lower back and ankles (green). The EVS and IMU come together in the cross-body back, which is 17cm wide, 32.5cm tall, and 10.5cm thick. The cross-body bag contained the myRIO-1900, EVS stimulator and battery. On the back was the Cosmed k5 with the harness, measuring the oxygen in and carbon dioxide out, during breathing by the connected mask that covered the nose and chin. Raw traces of the applied EVS and mediolateral linear accelerations (a_{ML}) from the IMUs are shown for a single stride of walking at 78 steps/min and 0.8 m/s. The IMU body reference frame is presented as oriented while the participant is standing still, and would move with the local body segment orientation.

IMUs

To measure the kinematics of the head, lower back and both ankles, four IMUs (MPU 6050; accelerometer range = ± 16 g; gyroscope range = ± 200 deg/s) were used to measure the angular velocities and linear acceleration of these body parts (Appendix A, Figure 7A). The IMU on the head was placed at the centre of the forehead. The IMU on the lower back was placed over the third lumbar spinous process to estimate the resultant whole-body balance responses evoked by EVS during locomotion. This location, which approximated the body centre of mass, corresponded with previous studies (21, 32, 39) and was measured for later use in the signal analysis per participant (112.5 ± 5.4). The IMUs on the ankles were placed just above the lateral malleoli of the right and left ankles to detect gait events, such as heel strike and toe-off, and to measure vestibular-evoked responses in the lower limbs. This location was again measured for later use in the analysis (11.1 ±1.2 cm). All the IMUs were attached to the skin using double-sided hair body garment tape on the 3D printed IMU cover (Appendix A, Figure 7A) and taped over with kinesiology tape (Sport Support) to ensure there was no tugging of the wires. Lastly, Velcro straps were used to keep the IMUs on the lower back and ankles in place. Before the data collection, a calibration of each IMU was done to correct the gain of the accelerometers and any offsets (i.e. velocity offsets) in the accelerometers and gyroscopes to the reference frame with X forward and Z upwards. The IMUs were then calibrated once before the start of each experiment to estimate the correct body-reference frame (See Protocol).

2.2.2 COSMED K5

To measure and analyse the gas exchange rates, the COSMED K5 (9260AA6, COSMED, Italy) was used (40). This is a wearable metabolic analyser consisting of a device that can be worn as a backpack using a harness, along with a silicone mask that covers both mouth and nose to do a breath-by-breath analysis (Appendix A, Figure 8A). There were different sizes of masks, varying from extra small, small, to medium, which were checked to be a correct fit with every participant. The COSMED K5 was calibrated each morning before the experiments by using a certified gas mixture (16% O2, 5% CO_2), a CO_2 scrubber (C04408-01-07, COSMED, Italy), and a 3 L calibration syringe (C00600-01-11, Hans Rudolph, United States), and cleaned after every use, following the manufacturer's guidelines.

2.3 Protocol

2.3.1 EVS familiarisation

The participants were advised to eat a light meal before, refrain from alcohol and nicotine for 2 hours before, and coffee for 4 hours before the experiment. When participants arrived, their standing height, weight, and ankle height were measured. Participants were familiarised with the stimulation signal by progressively increasing amplitudes in steps of 1 mA, starting from a signal with a peak amplitude of 1 mA and increasing it until reaching the maximum peak of 4.5 mA. Placement of the rubber electrodes was adjusted when participants indicated any stinging sensations due to uneven conductivity.

2.3.2 IMU to body calibration

The IMU calibration followed the protocol of Foulger et al. (32). Once the IMUs were attached to the participants, a calibration to a standard body reference frame was performed, with X: forward (anterior), Y: right (lateral), Z: down (inferior) (as seen in Figure 1). Two static poses were recorded per IMU to define the body reference frame. Starting with defining the Z axis as pointing downwards, where participants were asked to stand upright with their head pitched perpendicular to gravity, following Reid’s plane. Reid´s plane was chosen for the calibration pose so that the head could be oriented with respect to the net response evoked by the EVS to give feedback to maximise the vestibular evoked response. The second pose was each respective body segment ~90 degrees forward down to define the approximate X axis forward. Next, while standing upright, both legs were flexed to a ~90-degree angle per leg, supported by a beer crate (Appendix A, Figure 9A) to define the approximate X axis in the IMUs placed on the ankles. The approximate X axis and the true Z axis were defined as being oriented opposite to the net acceleration vector caused by the gravitational field. The Y axis was computed as the cross product of the Z axis and the approximate X axis, and the true X-axis was subsequently obtained as the cross product of the Y and Z axes to ensure the resulting axes are orthonormal.

2.3.3 Preferred walking speed estimation

Next, participants were asked to walk overground at their preferred walking speed (PWS) in a 26-meter-long hallway, using a pacing cart to estimate the walking speed (see Appendix A, Figure 10A) that was held by the experimenter who walked behind the participant at the same speed. This was done twice, and the mean of both trials was taken as the overground PWS. To set the PWS on the treadmill, the protocol of Jordan et. al. (41) was followed. Participants started walking on the treadmill at a relatively slow speed of 2.7 km/h, which was increased by 0.1 km/h until the participant indicated they were walking at their treadmill PWS. This chosen speed was then increased by 1.5 km/h and subsequently decreased in steps of 0.1 km/h until the participant indicated their PWS. This procedure was followed twice, and the mean of both trials was used as the treadmill PWS, which was the speed that was set in a control trial on the treadmill (see Treadmill trials). The participant was blinded to the set speed during this process.

2.3.4 Resting metabolic rate

The participant's resting metabolic rate was measured by having participants stand upright and unsupported for 5 minutes. This measurement was subtracted from the gross metabolic cost of walking to determine the net cost of walking (NCoW) in J/kg*m. After this final preliminary measurement, the primary walking trials were performed, where half of the participants started with treadmill walking and the other half with overground walking and the order of the trials within conditions was randomised.

2.3.5 Overground trials

During overground trials, participants were asked to walk outside in a straight path on a 65-meter-long terrain. At the end of this terrain, participants were instructed to make a quick turn without using too many extra steps. Whenever the weather was unsuitable for continuing outside, a 45 m inside walkway was used for the overground trials (Appendix A, Figure 11A). The same pacing cart that was used to estimate the PWS was used to ensure that participants walked at the correct speed. An experimenter walked beside and behind the participant, out of their field of vision, and provided verbal instructions to help the participant maintain the desired walking speed (Appendix A, Figure 11A).

The overground trials consisted of four 5-minute walking trials in a randomised order. One trial was conducted at the participant's PWS to determine the baseline energetic cost of walking. A second trial was performed at the same speed while applying electrical vestibular stimulation (EVS) to assess its effect on walking cost and the body's corrective response, as measured by the IMUs. The other two trials were at the slower walking speed of 0.8 m/s (2.88 km/h) and a step frequency of 78 steps/min, where the participant maintained this cadence using a metronome played via in-ear headphones and delivered by the myRIO-1900 (National Instruments). This exact speed and cadence were chosen to follow other studies that examined vestibular responses evoked by EVS (19, 21, 22, 32, 38). An additional trial at this slower speed was performed without EVS to establish the baseline cost of walking without stimulation. The other trial with EVS was used to observe the effect of the vestibular stimulation on a slower walking speed compared to the faster walking speeds at the participant's PWS. The participant's head was kept pitched up at a ~17-19-degree angle (defined with respect to Reid's plane) to maximise the net vestibular response. Following every trial, participants were allowed to take a 5-minute break to hydrate and sit down before starting the next trial.

2.3.6 Treadmill trials

In the treadmill trials, participants were asked to walk five 5-minute trials, again in a randomised order. Four of the five were performed with the same condition as during overground walking (see Overground trials). So, two trials at the PWS set overground (with and without EVS) and two at 0.8 m/s (with and without EVS). One additional trial was performed on the treadmill in which participants walked at their PWS set on the treadmill without EVS. This trial acted as a control trial to determine whether the cost of walking differed at their treadmill PWS compared to walking at their overground PWS. Before the start of every trial, the treadmill was set to the intended speed to let the participant get familiar with walking at that speed for a few seconds (Appendix A, Figure 11A).

During the trials with EVS, the participant's head was kept pitched up at a ~17-19-degree angle to again maximise the vestibular response. This was done by marking a point on the wall for the participants to look at and by giving verbal feedback. There was the option to rest between the trials and to hydrate.

2.4 Signal analysis

The IMU data were filtered using a dual-pass lowpass fourth-order 80 Hz Butterworth filter. The stochastic EVS data were low-pass filtered at 20 Hz with a dual-pass fourth-order Butterworth filter to reduce signal noise while preserving relevant signal components. All the following analysis steps were done by following the method and MATLAB code from Foulger et al. (32).

2.4.1 IMU orientation estimation

The orientation of each body segment was determined using IMU data. Segment tilt in pitch (around Y axis) and roll (around X axis) was estimated with a complementary filter (42).

$$\theta_i = (\theta_{i-1} + \omega \cdot dt)G + (\varphi_i)(1 - G) \quad (1)$$

Where θ_i is the orientation estimate at the ith sample, ω is the gyroscope angular velocity around the same axis, dt is the sampling interval (5 ms), G is the filter weighting factor (0.995), and φ_i is the accelerometer-derived orientation. This filtering allows the short-term accuracy of the gyroscope to be fused with the long-term stability of the accelerometer, thereby minimising drift while preserving sensitivity to rapid orientation changes. If the measured net acceleration deviated more than 10% from 9.81 m/s^2, orientation was estimated exclusively from gyroscope data. Accelerometer-based estimates of pitch and roll were computed as below with a = $[\ddot{x}\ddot{y}\ddot{z}]$.

$$\text{pitch} = \varphi_i = \text{atan2d}(\ddot{x}_i, \ddot{z}_i) \quad (2)$$

$$roll = \varphi_i = -atan2d(\ddot{y}_i, \ddot{z}_i) \quad (3)$$

2.4.2 IMU linear acceleration gravity correction

Since linear accelerometers measure both gravitational and inertial accelerations, the gravitational component was removed post-recording to isolate the inertial acceleration (a^*). Using the IMU orientations estimated from the complementary filter (rotation matrix R, derived from pitch and roll while assuming no yaw), the gravity vector (g = [0, 0, –9.81]) was rotated into the sensor frame and subtracted from the raw acceleration:

$$a^* = (a - gR) \quad (4)$$

The corrected signals were then low-pass filtered at 20 Hz using a dual-pass fourth-order Butterworth filter.

2.4.3 Stride detection

Stride segmentation was performed using the ankle IMUs. Heel strike was identified as the first local minimum of the mediolateral (global Y axis) angular velocity following the mid-swing peak, and toe-off as the zero-crossing of the Y axis angular velocity preceding that peak (43). A stride was defined from right heel strike to the point immediately before the subsequent right heel strike.

Because the overground walkway was only 65 meters, participants had to make quick turns when walking for 5 minutes. These turning strides during the overground trials were excluded based on the angular velocity of the Z axis of the IMU on the lower back. A moving average (200-sample window) was used to detect rotations exceeding 20 deg/sec, which were extended by 600 samples (3 seconds) before and after the turn to ensure data contained only periods of walking without turning.

2.4.4 Time-dependent frequency analysis

To assess the phasic modulation of vestibular-evoked balance responses during walking, the time-varying coherence between the EVS signal and body kinematics was quantified. Kinematic signals were obtained from the gravity-corrected mediolateral accelerations (a_{ML}) measured at the lower back, left and right ankles. Coherence quantifies the frequency-specific relationship between an input and output signal. It is analogous to correlation in the time domain, where a value of 0 indicates no similarity and a value of 1 reflects a perfect correspondence at a given frequency (44, 45).

Time-frequency analysis was performed using Morlet wavelet decomposition (35, 44, 46). To account for the latency between EVS input and balance responses during the strides, the EVS signal was shifted 200 ms forward before analysis (17, 47, 48). The wavelet composition was applied with a 0.5 Hz resolution between 0.5 and 20 Hz to extract time-dependent coherence across the stride cycle.

Stride durations and gait events (right heel strike, toe-off, left heel strike, right toe-off) were normalised to each participant's mean values. The data were separated per walking speed (PWS and 0.8 m/s) and age groups (younger and older adults), where a comparison was made between overground and treadmill walking. For this comparison, stride durations and gait timings were additionally normalised to the means of that age group and walking speed, ensuring comparability across participants while maintaining alignment with gait events. The time-dependent coherence was calculated as follows:

$$C(\tau, f) = \frac{|P_{xy}(\tau, f)|^2}{P_{xx}(\tau, f)P_{yy}(\tau, f)} \quad (5)$$

Here, τ is the given time point in the stride cycle, and f is frequency. $P_{xy}(\tau, f)$ is the time-normalised cross-spectrum between EVS and a_{ML}, $P_{xx}(\tau, f)$ the auto-spectrum of EVS, and $P_{yy}(\tau, f)$ the auto-spectrum of the a_{ML}. Also, the peak coherence was retrieved per walking speed per overground and treadmill per age group for the statistical analysis.

2.4.5 Metabolic cost

The metabolic cost was analysed by calculating the gross and net cost of walking (CoW) in metabolic energy expended per kilogram of body mass per meter travelled, discarding the first 60 seconds of data to remain with the steady-state. The gross cost of walking (GCoW) was retrieved using Formula 6, where the energy expended per minute (EEm in Kcal/min) was multiplied by a constant 69.78 (calculated from SI definitions) and divided by the mass of the participant (kg) and by the walking speed (m/s). Then the net cost of walking (NCoW) was retrieved by subtracting the gross cost of the resting metabolic rate (GC_R, EEm multiplied by the constant and normalised by the mass) from the gross cost of the trial (GC_T) and dividing by the speed (Formula 7).

$$GCoW = \frac{EEm \cdot 69.78}{mass \cdot speed} \quad (6)$$

$$NCoW = \frac{GC_T - GC_R}{speed} \quad (7)$$

Since one of the participants' resting metabolic rate measurement failed, a linear regression model was created to calculate the baseline from the GCoW to calculate the NCoW. Here, the regression was significant, so their GCoW was used to get the predicted baseline.

2.5 Statistical analysis

First, the normality of all continuous variables was assessed using the Shapiro-Wilk test. Between-group differences in anthropometric measures (e.g., height, weight) and preferred walking speed were evaluated using independent-samples t-tests when normality assumptions were met. Otherwise, a Mann-Whitney U test was used to analyse the between-group differences. To examine the effects of age group (young versus older adults), walking condition (overground versus treadmill), and EVS on the cost of walking, a repeated measures ANOVA was performed. Here, the age group was the between-subject factor and walking condition and the presence of EVS were the within-subject factors. Separately, peak coherence values per trial, along with their corresponding frequency and timing within the stride cycle, were extracted to quantify the effect of EVS on lower back and ankle kinematics. A second repeated measures ANOVA was conducted to assess the effect of age group, walking condition and walking speed (PWS and 0.8 m/s). Here, the age group was again the between-subject factor and walking condition and walking speed were the within-subject factors. For both ANOVAs, statistical significance was defined as $p<0.05$ and post-hoc comparisons were conducted with the Holm correction when applicable. The ANOVA results were shown with the *F*-value and its degree of freedom, and the *p*-value. Analyses were performed using JASP 0.95.1 (49), and signal processing, as well as visualisation of results (connected scatterplots and boxplots), was conducted in MATLAB R2023b.

3 Results

We first compared the anthropomorphic measures and preferred walking speeds between older (OA) and younger adults (YA) (see Table 1) to test if both age groups were comparable. All data metrics were found to be normally distributed ($p > 0.05$), and no significant difference was found between the two groups in the listed parameters, except for age. Table 2B in Appendix B gives an overview of this data per participant. Four out of the 20 participants had previously experienced GVS and were all part of the young group. All young participants had walked on a treadmill before this experiment, while only five out of the ten older adults had previously walked on a treadmill.

Table 1: Age, anthropometric parameters and PWS of younger and older adults.

	Younger adults (n=10)	Older adults (n=10)	*p*-value
Age	26.4 ± 2.5	68.6 ± 3.6	<.001
Female	4 (40%)	4 (40%)	
Height (cm)	182.1 ± 9.2	177.1 ± 5.9	0.16
Weight (kg)	78.9 ± 16.4	76.7 ± 8.2	0.71
BMI (kg/m^2)	23.7 ± 4.1	24.5 ± 2.7	0.62
Ankle height (cm)	11.1 ± 1.5	11.2 ± 1.1	0.81
Overground PWS (m/s)	1.33 ± 0.24	1.33 ± 0.14	0.97
Treadmill PWS (m/s)	1.18 ± 0.17	1.18 ± 0.13	0.98

BMI body mass index, *PWS* preferred walking speed

The independent t-test was used for the numerical data

3.1 Metabolic cost of walking

We next examined the metabolic cost of overground and treadmill walking to determine differences between age groups and to assess the influence of walking condition on the metabolic cost. Figure 2 shows the comparison of the GCoW and NCoW during treadmill and overground walking at the preferred walking speed set overground (PWS_{OG}). Here, a significant elevation in metabolic cost was found during treadmill walking compared to overground walking in both age groups, as indicated by repeated measurements using AVOVA (F (1,18) = 25.38, $p < 0.001$). This elevation was found in both the net and gross costs of walking and in younger as well as older adults (increase in GCoW; YA = 16.6 ± 14.5%, OA = 14.4 ± 11.1%, NCoW; YA = 23.2 ± 20.7%, OA = 23.5 ± 17.5%). These results contradict our original hypothesis, as there was no effect of the walking condition on age in the cost of walking (F (1,18) = 1.13, p = 0.301).

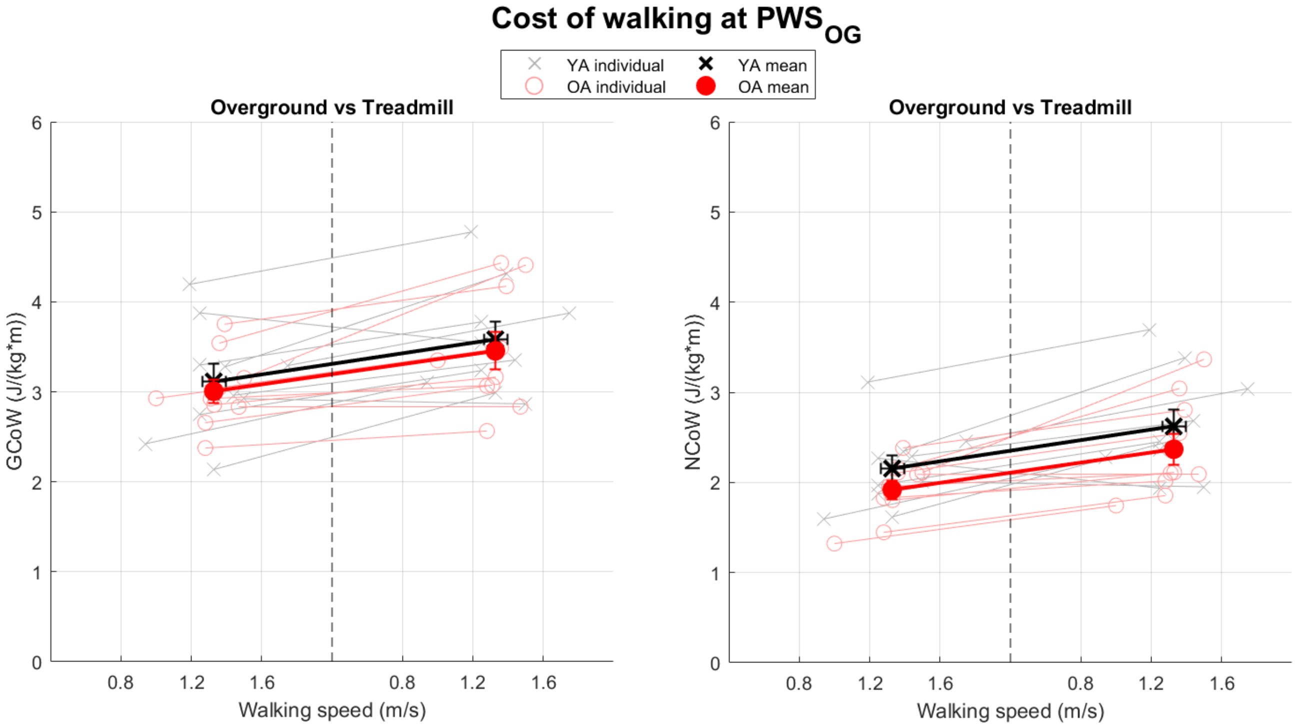

Figure 2: Connected scatter plots showing the mean cost of walking and walking speed with error bars and the single subjects between overground and treadmill walking in Gross (left) and Net (right) Cost of walking at their overground preferred walking speed (PWS_{OG}). The black crosses display the younger adults (YA) and the red circles the older adults (OA). There is a statistically significant elevation of metabolic cost in treadmill walking compared to overground walking (F (1,18) = 25.38, p < 0.001).

To analyse the effect of electrical vestibular stimulation (EVS) across walking conditions and age groups at their PWS_{OG} (Figure 3), the NCoW was compared, as it captures an isolated estimate of the cost of walking without the basal metabolism. The repeated measures ANOVA revealed a significant increase in NCoW during the EVS trials compared to the trials without EVS in both age groups (F (1,18) = 38.06, p < 0.001). This elevation was found in both overground (OG) and treadmill (TM) walking in younger as well as older adults (OG; YA = 26.0 ± 29.3%, OA = 21.2 ± 19.0%, TM; YA = 6.5 ± 9.0%, OA = 20.6 ± 21.0%). The effect of an increase in the cost of walking on the treadmill seemed to be larger in older than in younger adults (Figure 3, right panel); however, this effect was not significant, as there was no interaction between age and EVS (F (1,18) = 0.25, p = 0.621). There was also no interaction between EVS and walking conditions (F (1,18) = 0.08, p = 0.318). Resulting in no difference in age and walking condition on the cost of walking with and without EVS.

Participants also walked at a fixed walking speed of 0.8 m/s to examine any changes in the metabolic cost at a walking pace that evokes larger contributions of the vestibular system (19, 31, 32, 36, 50). The GCoW and NCoW were both significantly elevated during treadmill walking compared to overground walking (F (1,18) = 16.60, p < 0.001) with no effect of age (F (1,18) = 0.23, p = 0.637) on the cost of walking, as seen in Figure 12C in Appendix C. There was also a significant elevation of the NCoW found with EVS compared to no EVS (F (1,18) = 37.49, p < 0.001). Here, there was no effect of age (F (1,18) = 2.22, p = 0.153) and walking condition (F (1,18) = 3.79, p = 0.067) on the cost of walking with EVS (Appendix C, Figure 13A). These results found in the 0.8 m/s trials are comparable to what was found during the PWS_{OG} trials. A table with all the means and standard deviations of the cost of walking can be found in Table 3B, Appendix B.

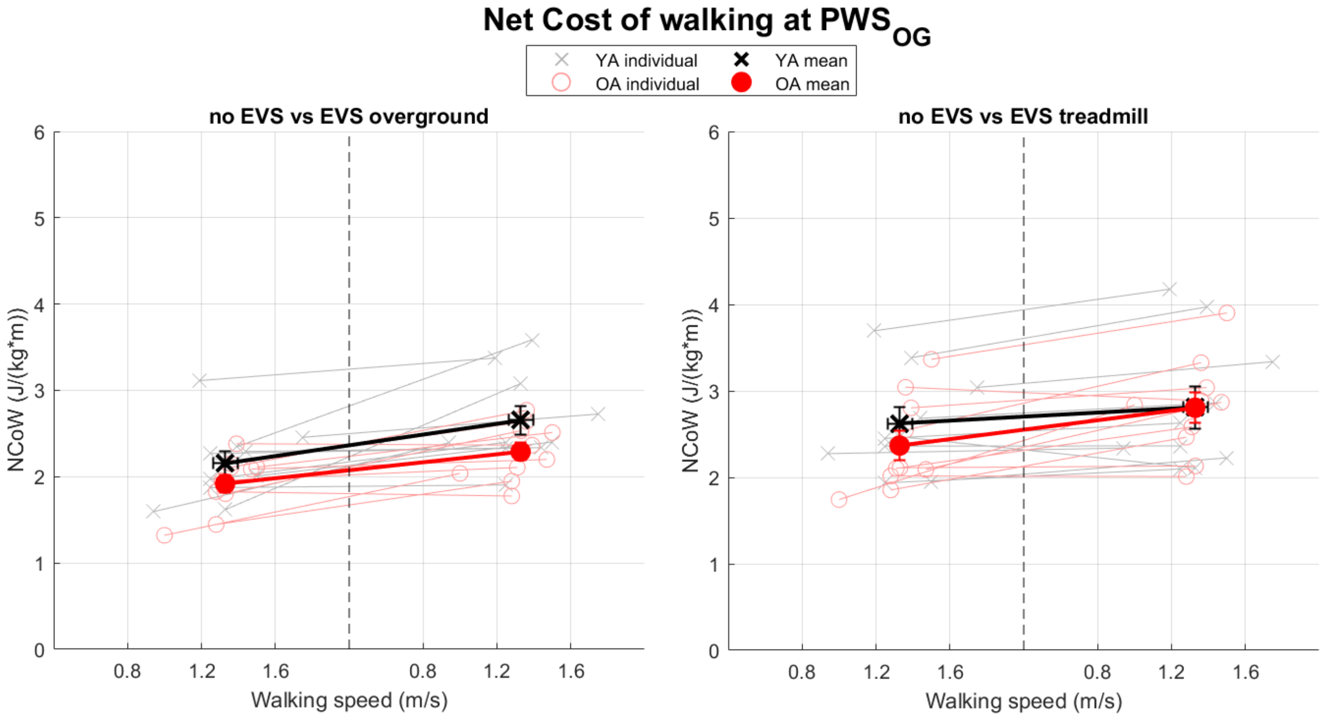

Figure 3: Connected scatter plot showing the mean with error bars and the single subjects between with and without EVS while overground (left) or treadmill walking (right) in Net Cost of walking at their overground preferred walking speed (PWS$_{OG}$). The black crosses represent the younger adults (YA), and the red circles represent the older adults (OA). There is a statistically significant elevation of metabolic cost with the presence of EVS compared to no EVS (F (1,18) = 38.06, p = < 0.001).

To control for the effect of using the overground PWS during treadmill trials, we performed an extra trial during treadmill walking by asking participants to walk at their treadmill PWS$_{TM}$. Participant's treadmill PWS$_{TM}$ was significantly lower than the speed set for the PWS$_{OG}$ (F (1,16) = 23.23, p < 0.001). Younger adults maintained a 10.6 ± 12.4% slower walking speed, and older adults maintained a 13.7 ± 13% slower walking speed. Here, there was no effect of age regarding the PWS (F (1,16) = 0.003, p = 0.957). Meaning that both age groups overestimated their walking speed when walking on the treadmill (Appendix C, Figure 14C). When looking at the cost of walking on a treadmill at both PWS$_{OG}$ and PWS$_{TM}$, there was no significant difference found (F (1,16) = 0.38, p = 0.545) and no effect of the different walking speeds on age regarding the cost of walking (F (1,16) = 2.25, p = 0.131) as seen in Figure 15C in Appendix C and the data in Table 4B, Appendix B. Resulting in no effect of the different walking speeds on the cost of walking.

3.2 Vestibular contribution to balance

We next estimated the influence of vestibular stimulation on kinematic responses to examine the influence of different walking speeds and conditions in the two age groups. Here we calculated the coherence between the EVS signal and the linear acceleration in the mediolateral direction (a_{ML}) recorded from the lower back, right and left ankle as a measurement of the vestibular effect on the kinematic behaviour (see methods). The time-frequency analysis showing the group-average coherence of the right ankle a_{ML} is depicted in Figure 4. This figure depicts the magnitude of the coherence across time (stride duration) and frequency during overground and treadmill trials while participants walked at their PWS$_{OG}$ and at 0.8 m/s. Across all conditions, coherence is modulated across the entire stride cycle but peaks during the stance phase and at 2-8 Hz. Furthermore, during both overground and treadmill walking, the coherence is greater during the 0.8 m/s walking trials as compared to the PWS$_{OG}$ trials.

To investigate these group-average responses more in depth and to analyse if there are differences between the walking conditions and age groups, the peak coherences were extracted from the lower back, right and left ankle a_{ML} as seen in Figure 5. To analyse these results, a repeated measurements ANOVA was done, starting with the peak coherences found of the EVS and lower back a_{ML}. Here, a significantly higher peak coherence was found during the 0.8 m/s trials compared to the PWS_{OG} trials (F (1,18) = 21.23, p < 0.001). This elevation was found in both walking conditions and age groups (OG; YA = 68.6 ± 54.3%, OA = 37.96 ± 66.7%, TM; YA = 78.6 ± 80.0%, OA = 78.3 ± 82.7%). There was also a significant main effect of walking condition found on the peak coherence (F (1,18) = 5.25, p = 0.034); however, Post Hoc comparison with Holm correction showed no difference of walking condition at the PWS_{OG} (p_{holm} = 0.406) and in the 0.8 m/s trials (p_{holm} = 0.105). This shows that the significant main effect of walking condition is likely because the results are averaged over both the speed and age trials, making the effect seem larger than it truly is. There was no significant interaction of age on walking speed (F (1,18) = 0.02, p = 0.886) and on walking condition (F (1,18) = 0.173, p = 0.683) found on the peak coherence, meaning that both age groups reacted similarly to the EVS signal when looking at the back a_{ML}.

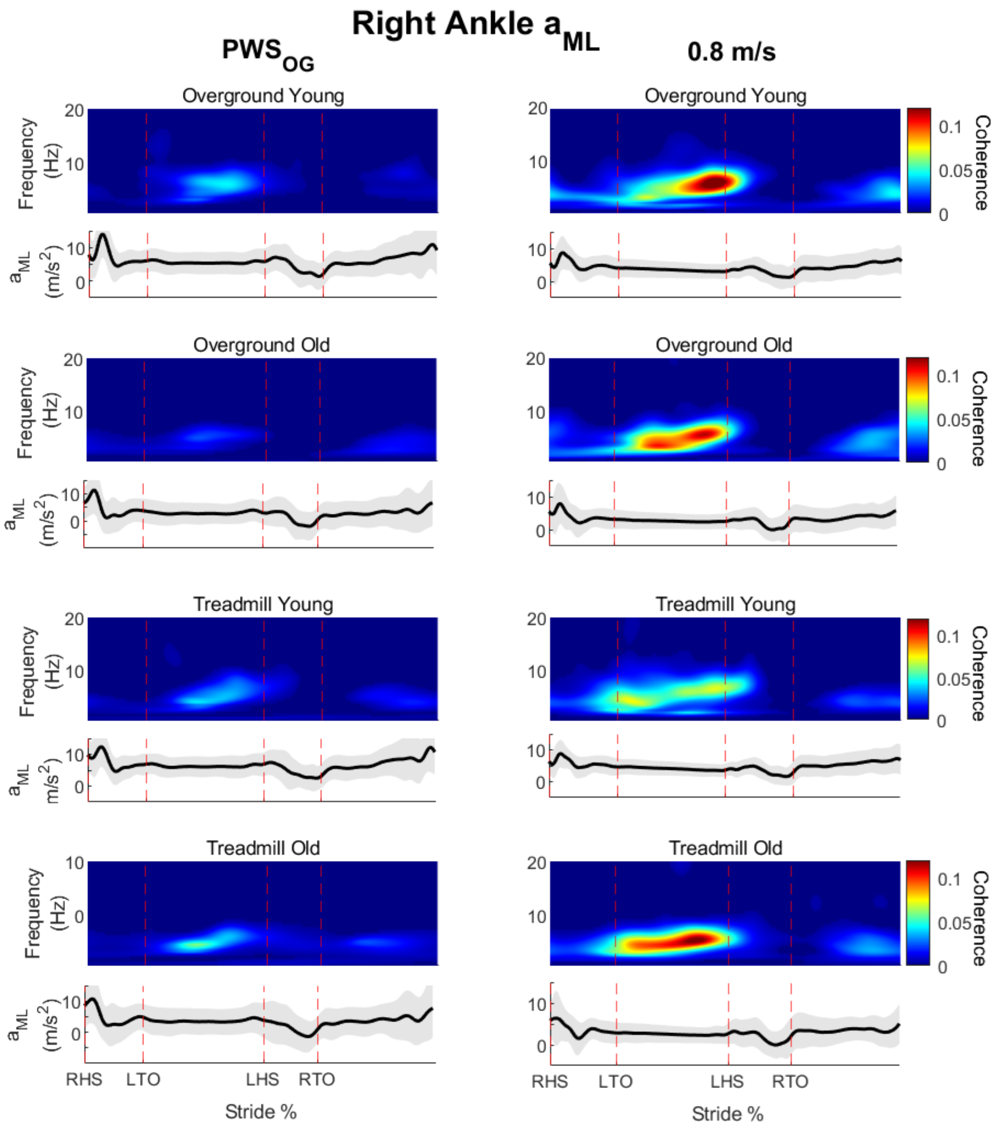

Figure 4: Participant pooled (Young = 10 and Old = 10) time-frequency analysis from the right ankle mediolateral linear acceleration (a_{ML}) for the preferred walking speed set overground (PWS_{OG}) (left column) and the 0.8 m/s trials, which had a set cadence of 78 steps/min (right column). The heatmaps display the stride-normalised time-dependent coherences, and the plots below show the average signal trace, with the shaded area representing ± standard deviation. The plots are divided into the top half, which were during overground walking in the younger (first row) and older adults (second row), and the bottom half, which were during treadmill walking in the younger (third row) and older adults (fourth row). The dashed red lines display the stride event at the percentage of occurrence during the stride cycle: RHS: right heel strike, LTO: left toe off, LHS: left heel strike, RTO: right toe off.

Analysis of the right ankle a_{ML} also showed coherence to be significantly higher at the 0.8 m/s trials compared to the PWS_{OG} (F (1,18) = 42.72, p < 0.001). In both age groups and walking conditions, this elevation was found (OG; YA = 100.5 ± 67.1%, OA = 122.3 ± 97.4%, TM; YA = 76.6 ± 104.1%, OA = 79.6 ± 79.1%) as observed in Figure 4. Here, there was no significant effect of walking condition (F (1,18) = 1.46, p = 0.242) and no effect of walking condition on age (F (1,18) = 3.11, p = 0.095). Overall, there was no difference in peak coherence between the walking conditions and age groups in the right ankle a_{ML}.

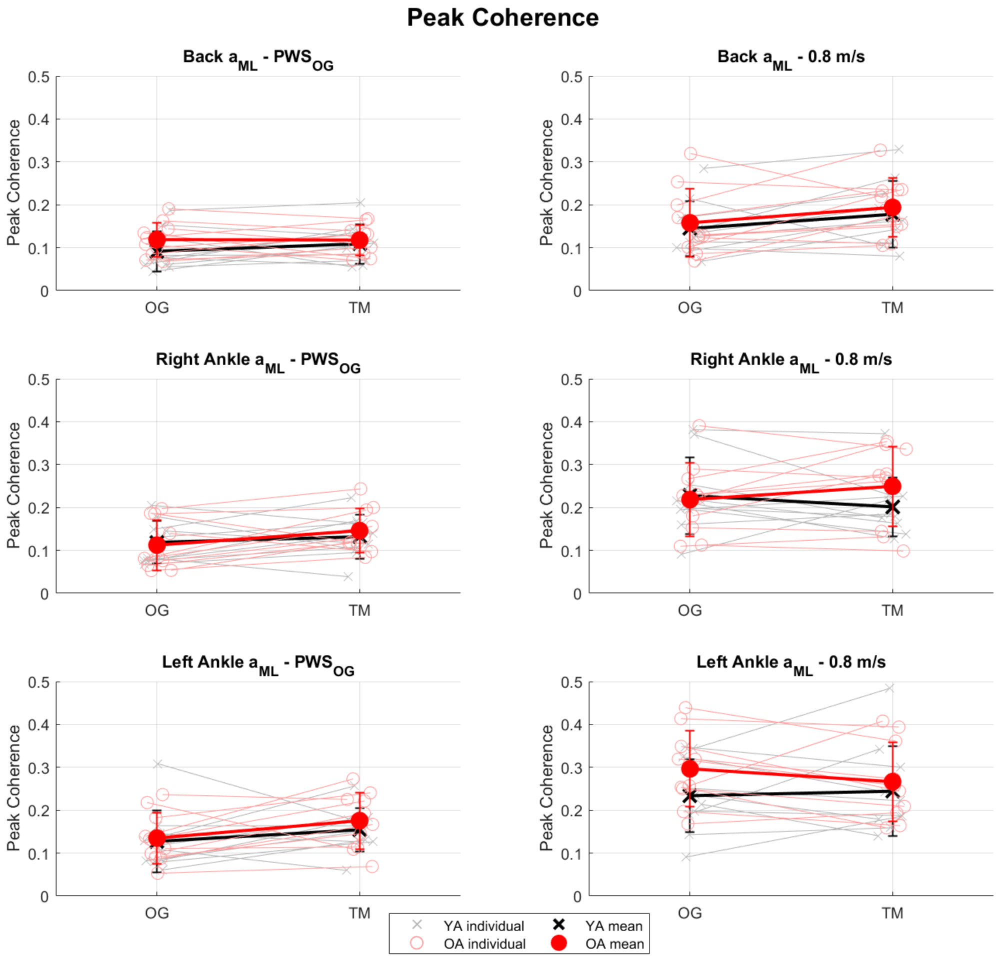

Figure 5: Connected scatterplot showing the peak coherence between the Electrical Vestibular Stimulation (EVS) signal and the back mediolateral linear acceleration (top), right ankle a_{ML} (middle), and left ankle a_{ML} (bottom) in younger (YA; black cross) and older adults (OA; red circle) and their mean and standard deviation. The right column shows the peak coherences during the preferred walking speed set overground (PWS_{OG}), and the right column shows the 0.8 m/s walking at 78 steps/min. There is a statistically significant elevation of peak coherence during the 0.8 m/s walking compared to the PWS_{OG} (p < 0.001) in both the back, right and left ankle a_{ML}.

Lastly, the left ankle a_{ML} also showed a significantly higher peak coherence at the 0.8 m/s walking trials compared to the PWS_{OG} trials (F (1,18) = 31.21, p < 0.001). Here, the elevation was again found in both walking conditions and age groups (OG; YA = 102.0 ± 70.7%, OA = 150.5 ± 104.2%, TM; YA = 64.4 ± 62.9%, OA = 72.2 ± 91.7%). There was also a significant interaction effect found between walking condition and speed (F (1,18) = 6.81, p = 0.018); however, the Post Hoc comparison with Holm correction only showed significant differences between the walking conditions at different walking speeds, and not at the same walking speed (PWS_{OG}: p_{holm} = 0.106, 0.8 m/s: p_{holm} = 0.575). Overall, these results suggest that the peak coherence was affected by the imposed walking speed, but not by the walking condition. There was also no condition effect found on age (F (1,18) = 0.18, p = 0.674). All the means of the peak coherences can be found in Appendix B, Table 5B.

4 Discussion

This study investigated the combined effects of age, walking condition (treadmill versus overground), and vestibular contributions to balance on the cost of walking. While the treadmill consistently increased the cost of walking compared to overground, the expected age-related difference was not observed. Furthermore, the application of EVS increased the cost of walking in both age groups, without indicating an age-specific effect. Vestibular coherence analyses similarly revealed no age effect but confirmed the influence of walking speed on vestibular contributions to walking balance. The peak coherence decreased with walking speed in both age groups. Here, no large effect of walking condition, as was found in the cost of walking.

When participants walked at their PWS_{OG}, both younger and older adults exhibited significantly higher NCoW and GCoW on the treadmill compared to overground walking. This finding is partly consistent with previous studies reporting an increased metabolic cost in treadmill walking in older adults compared to overground (11, 13, 14, 17). However, the current study did not observe a higher cost of treadmill walking in older adults compared to younger adults, as other studies did (13, 15-18). Instead, both age groups showed a mean elevation of ~15% of the mean GCoW and ~23% of the mean NCoW when walking on the treadmill as compared to overground walking. Therefore, the observed increase in walking cost for younger adults contrasts with the initial expectations, aligning instead with findings by Martin et al. (51), who also reported elevated treadmill walking costs in younger adults. However, based on work from Das Gupta et al. (13), we expected a larger elevation of older adults during treadmill walking. One possible explanation for the absence of this larger elevation could be due to the age range of the older participants measured in our study. Studies that found a higher cost of treadmill walking in older adults measured adults with a mean age of ~75 years old (13, 15, 17), but our older adults were 68.6 ± 3.6 years old. However, Ciprandi et al. (18) also had a population of ~68 years old and found this elevation in older adults during treadmill walking compared to younger adults. In addition, Malatesta et al. (16) measured both 25-, 65- and 80-year-olds and found a difference in the cost of treadmill walking between the 25- and 65-year-olds, but only from a walking speed of 1.33 m/s. This was the mean PWS_{OG} in both our age groups, too, so when comparing the data from this study to the study from Malatesta et al. and Ciprandi et al., a similar result should have been observed. This is, however, not the case. Meaning that both age groups showed a similar increase in walking cost on the treadmill compared to overground walking, deviating from previous studies that reported a stronger age effect.

The observation that preferred walking speed overground (PWS_{OG}) was similar between age groups was expected and consistent with previous studies (13, 52, 53). The estimated preferred walking speed on the treadmill (PWS_{TM}) was 1.18 m/s in both age groups, significantly lower than the PWS_{OG}. Because differences between overground and treadmill walking speeds are well established (54-56), an additional treadmill trial was conducted in which participants walked at their individually determined PWS_{TM}. This allowed us to compare walking costs at PWS_{TM} and PWS_{OG} on the treadmill. If walking speed had influenced energetic cost, a difference between these two trials could partially explain the elevated cost typically observed during treadmill walking, as participants would not be walking at their most efficient speed when using the PWS_{OG} on the treadmill (2). However, no significant differences in gross or net cost of walking (GCoW, NCoW) were found between these conditions for either age group. Therefore, the higher energetic cost of treadmill walking cannot be attributed to participants walking at their overground rather than their treadmill preferred speed.

The application of EVS led to a significant increase in the cost of walking across both age groups and walking conditions at the PWS_{OG} and the 0.8 m/s walking speed, reflecting the additional energetic demands of compensating for perturbed vestibular input. Participants were stimulated with a stochastic 0-20 Hz signal, which is known to evoke muscle responses in the lower limbs (57) and could be the cause for the elevated cost of walking. It has been found that a low-frequency EVS signal affects the cost in standing balance, but that the sway variability itself, rather than the vestibular activity, is the primary determinant of the cost of standing (58). In the case of walking, the increase in metabolic cost with EVS may be due to the corrective responses to the sway variability induced by vestibular activity, rather than solely to the vestibular cost. For future research, it is interesting to see if it would be possible to estimate the corrective cost on its own.

An analysis of the vestibular-evoked coherence between the EVS signal and linear acceleration in the mediolateral direction was used to investigate vestibular contributions across the walking conditions. The coherence was mostly present during the stance phase, which parallels existing literature done on both overground (32-34, 50) and treadmill walking (19, 21, 34-38). The peak coherence was found to have a significant elevation during the 0.8 m/s trials in the back, right ankle and left ankle a_{ML} compared to the PWS_{OG} trials. The peak coherence is known to decrease with walking speed in both overground (31, 32, 50) and treadmill walking (19, 36). So, it is not unusual that the PWS_{OG} trials would result in a lower peak coherence than in the 0.8 m/s trials in both walking conditions. There was some effect of the walking condition on the peak coherence, but the post hoc correction showed that this effect was established due to the large speed effect rather than the condition effect. Iles et al. also compared treadmill to overground walking with EVS and found no significant difference between the two walking conditions (34). They only tested younger adults, but it compares to what this study observed, as there is no large effect of the walking conditions on the peak coherence. In correlation with the metabolic results, no effect of age was found on the peak coherence. There has been no work yet in comparing older and younger adults in walking with EVS, but there is work on standing balance. Dalton et al. (47) observed higher peak coherence in older adults; these adults were approximately 10 years older than the older adults in this study, which could be the difference in results found. It is also known that EVS has a higher peak value in standing balance than in walking (34), making it plausible that the interaction between age and peak coherence can be more pronounced in standing balance due to the amplification of the vestibular contribution during standing compared to walking.

Several limitations should be considered when interpreting the results of this study. First, the sample size was limited to 10 older and 10 younger adults to match previous studies in the field of the cost of walking (11, 13, 16) and vestibular contribution to walking (32). While this allowed consistency with prior research, a post-hoc power analysis showed a small effect size (<0.25). A larger population could maybe increase the significance between the age groups, but the effect size should be taken into consideration for future research on this work. Second, although overground walking was conducted outdoors on level, linear paths, this environment still represents a semi-controlled scenario compared to natural daily walking, as unforeseen distractions occurred during the trials, such as people occasionally walking and driving by, and varying weather conditions. So did the temperature slightly differ between the experiments (range of 17-25 degrees Celsius). Also, treadmill trials were conducted in a basement with limited ventilation, which we tried to improve with a fan, but still introduced potential differences between the treadmill and overground walking trials. Next, the study population consisted of healthy, young adults, and the older participants were relatively young. While this minimised confounding factors such as musculoskeletal disease or balance impairments, it limits the generalizability of the findings to clinical populations or older adults with mobility limitations. Also, the use of the pacing cart to estimate participants' PWS_{OG} and to regulate treadmill speed introduced some variability due to the limited precision of this cart. It is not certain that the participants all walked at the same speed throughout all the trials. Finally, this walking speed was set before the start of the trials; this could have impacted the natural reaction on the EVS in older adults. Brown et al. found that the cost of walking increases, and the walking speed decreases when stability is impaired in older adults (25). It would be interesting in future research to test whether older adults have a higher cost of walking and see the reaction in walking speed compared to younger adults. This is difficult to pursue in treadmill walking, but it would be interesting to check in overground walking.

5 Conclusion

This study investigated whether the increased metabolic cost of walking commonly observed in older participants when walking on a treadmill compared to overground walking is driven by an elevated balance response facilitated by the vestibular system's contribution. The results confirmed that treadmill walking increases the cost of walking compared to overground walking. However, there was no age effect, meaning that this data did not show an interaction of age on walking condition on the cost of walking. There was also an effect of electrical vestibular stimulation found on the cost of walking. The peak coherence analysis showed that there was a difference in speed, but not a large effect of walking condition, suggesting that the increased cost of treadmill walking is not due to vestibular contributions in this measured population.

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Appendix A: Materials

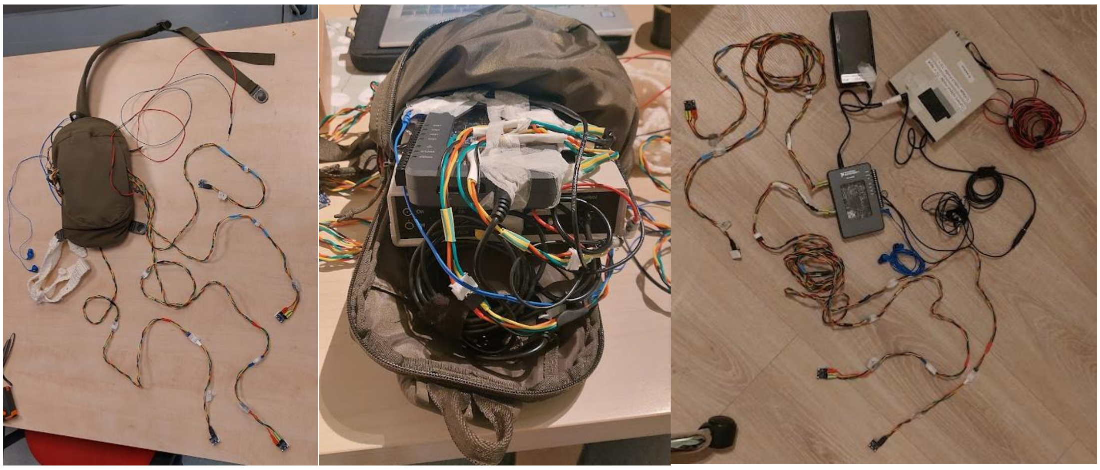

Figure 6A: Portable system showing the wires coming out of the bag, including IMUs, earplugs and EVS wires together with a handmade extra band to secure the bag in place (left), organisation inside the bag including the Labview MyRIO, stimulation box and battery (middle), all parts and their connection as it was organised in cross-body bag (right).

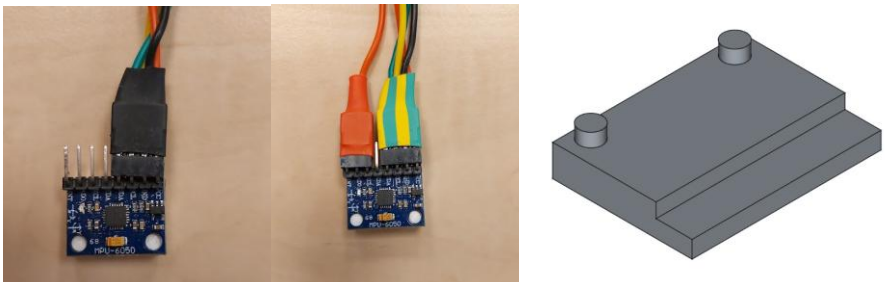

Figure 7A: IMUs and their connection to the wires, including power, ground, signal 1 (acceleration), and signal 2 (gyroscope). Two out of the four IMUs included an extra power wire (middle). The right shows the IMU cover, which was 3d printed and acted as a protection of the skin against the IMU back.

Figure 8A: COSMED K5 (40)

Figure 9A: The beer crate and pose that was used to calibrate the ankle IMUs to estimate the X-direction. This pose was repeated with the right ankle.

Figure 10A: Pacing cart that was used to measure the overground preferred walking speed and to control the speed during the overground walking trials.

Figure 11A: Treadmill (left), impression of outside trails with one of the experimenters acting as subject (middle), inside back-up walking way when it was raining (right).

Appendix B: Participants

Table 2B: Age, anthropometric parameters, preferred walking speed and temperature during the experiment per participant.

Participant		Sex	Age (years)	Mass (kg)	Height (cm)	PWS Overground (m/s)	PWS Treadmill (m/s)	Temperature
YA	1	M	27	95	188	1.25	-	19
	2	F	23	68	170	1.39	-	25
	3*	M	24	70	192	1.75	1.43	22
	4	M	30	85	195	1.50	1.10	22
	5	M	25	82	180	0.94	0.92	17
	6	M	25	85	186	1.25	1.06	18
	7	M	28	113	185	1.44	1.39	20
	8	F	28	63	168	1.25	1.14	23
	9	F	30	60	173	1.33	1.17	18
	10	F	24	68	184	1.19	1.22	20
mean		*4F, 6M*	*26.4*	*78.9*	*182.1*	*1.33*	*1.18*	
OA	1	M	65	81	176	1.28	1.08	24
	2	M	70	75	178	1.36	0.94	24
	3	F	70	70	176	1.31	1.17	22
	4	M	76	87	178	1.33	1.22	22
	5	F	65	67	174	1.00	1.06	20
	6	M	72	67	175	1.39	1.22	20
	7*	F	65	79	164	1.28	1.11	20
	8*	M	67	81	185	1.47	1.42	23
	9	F	67	70	180	1.50	1.25	19
	10	M	69	91	185	1.36	1.28	22
mean		*4F, 6M*	*68.6*	*76.7*	*177.1*	*1.33*	*1.18*	

** Walked inside due to the rain.*

Table 3B: Mean gross and net cost of walking (GCoW and NCoW) and standard deviation of the younger (YA) and older adults (OA) per walking conditions and with and without electrical vestibular stimulation (EVS).

Age group	Walking condition	Stimulation	GCoW (J/kg*m) PWS_{OG}	NCoW (J/kg*m) PWS_{OG}	GCoW (J/kg*m) 0.8 m/s	NCoW (J/kg*m) 0.8 m/s
YA	Overground	No EVS	3.11 ± 0.62	2.15 ± 0.45	3.86 ± 1.12	2.28 ± 0.87
		EVS	3.61 ± 0.57	2.65 ± 0.53	4.11 ± 1.00	5.54 ± 0.81
	Treadmill	No EVS	3.58 ± 0.61	2.63 ± 0.58	4.31 ± 0.95	2.73 ± 0.81
		EVS	3.77 ± 0.82	2.81 ± 0.77	4.82 ± 0.86	3.25 ± 0.75
OA	Overground	No EVS	3.01 ± 0.40	1.92 ± 0.33	3.94 ± 0.80	2.16 ± 0.86
		EVS	3.38 ± 0.50	2.29 ± 0.33	4.37 ± 0.64	2.59 ± 0.48
	Treadmill	No EVS	3.46 ± 0.67	2.37 ± 0.54	4.47 ± 0.72	2.69 ± 0.57
		EVS	3.89 ± 0.71	2.81 ± 0.56	5.31 ± 0.78	3.53 ± 0.64

Table 4B: Mean gross and net cost of walking (GCoW and NCoW) and standard deviation in younger (YA) and older adults (OA) for the overground (PWS_{OG}) and treadmill preferred walking speed (PWS_{TM}) when walking on the treadmill.

Age group	**Preferred walking speed**	**GCoW (J/kg*m)**	**NCoW (J/kg*m)**
YA	PWS_{OG}	3.58 ± 0.61	2.63 ± 0.58
	PWS_{TM}	3.52 ± 0.62	2.42 ± 0.51
OA	PWS_{OG}	3.46 ± 0.66	3.37 ± 0.54
	PWS_{TM}	3.65 ± 0.57	3.43 ± 0.43

Table 5B: Mean peak coherences and standard deviation of the back, right and left ankle linear acceleration in mediolateral direction (a_{ML}) in the younger (YA) and older adults (OA) in the different walking conditions at the overground preferred walking speed (PWS_{OG}) and 0.8 m/s trials.

Age group	Walking condition	Trial type	Peak coherence Back a_{ML}	Peak coherence Right ankle a_{ML}	Peak coherence Left ankle a_{ML}
YA	Overground	PWS_{OG}	0.09 ± 0.05	0.12 ± 0.05	0.13 ± 0.07
		0.8 m/s	0.14 ± 0.06	0.22 ± 0.08	0.23 ± 0.08
	Treadmill	PWS_{OG}	0.11 ± 0.05	0.13 ± 0.05	0.15 ± 0.05
		0.8 m/s	0.18 ± 0.07	0.20 ± 0.07	0.24 ± 0.10
OA	Overground	PWS_{OG}	0.12 ± 0.04	0.11 ± 0.06	0.13 ± 0.06
		0.8 m/s	0.16 ± 0.08	0.21 ± 0.08	0.29 ± 0.08
	Treadmill	PWS_{OG}	0.12 ± 0.03	0.15 ± 0.05	0.17 ± 0.06
		0.8 m/s	0.19 ± 0.07	0.24 ± 0.09	0.26 ± 0.09

Appendix C: Graphs

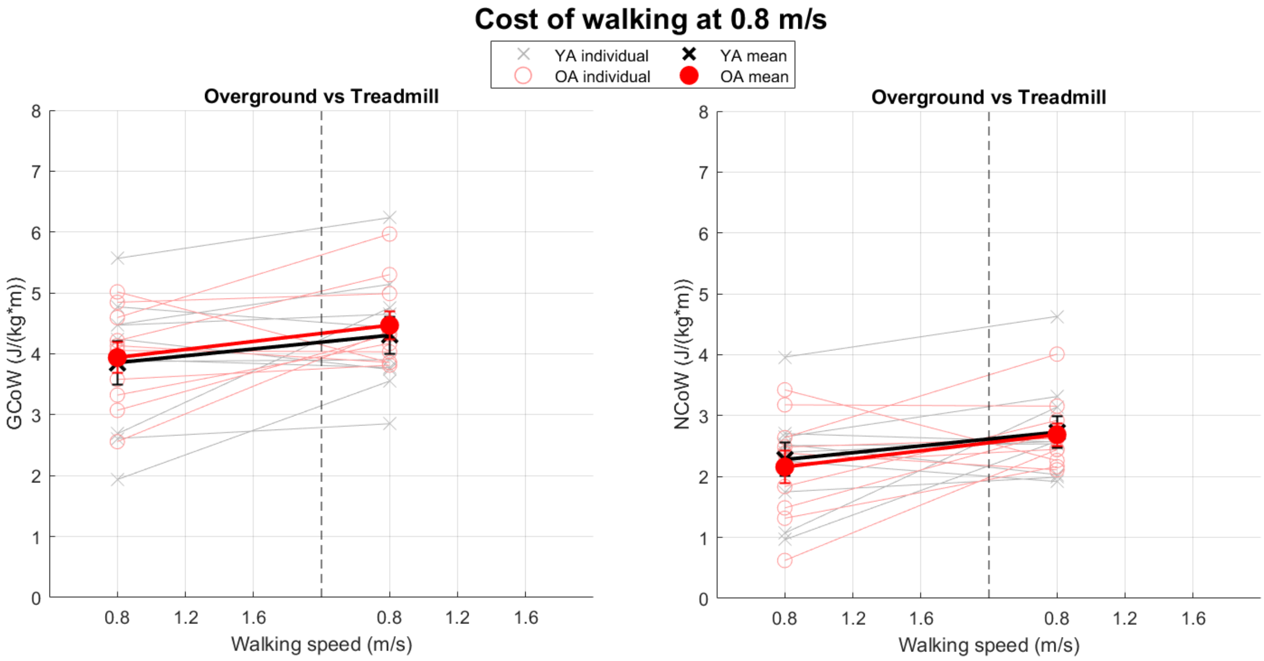

Figure 12C: Connected scatter plots showing the mean cost of walking and error bars, and walking speed with the single subjects between overground and treadmill walking in Gross (left) and Net (right) Cost of walking at a prescribed speed of 0.8 m/s and cadence of 78 steps/min. The black crosses display the younger adults (YA) and the red circles the older adults (OA).

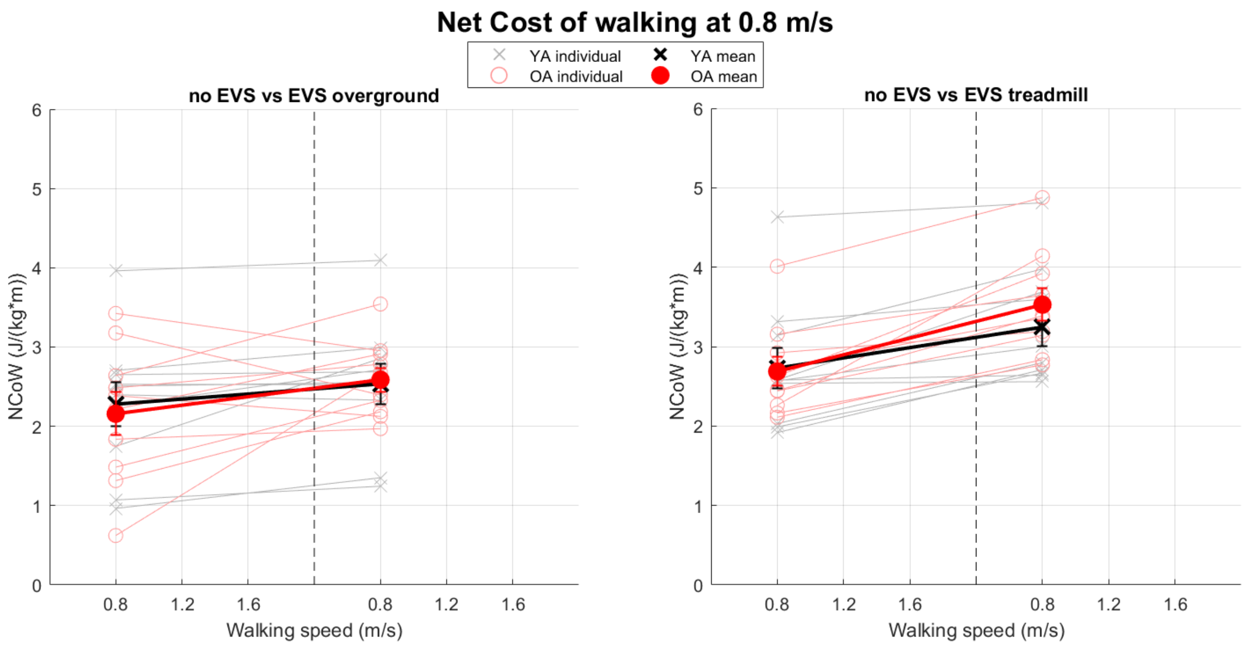

Figure 13C: Connected scatter plot showing the mean with error bars and the single subjects between with and without EVS while overground (left) or treadmill walking (right) in Net Cost of walking at their overground preferred walking speed (PWS_{OG}). The black crosses represent the younger adults (YA), and the red circles represent the older adults (OA). There is a statistically significant elevation of metabolic cost with the presence of EVS compared to no EVS.

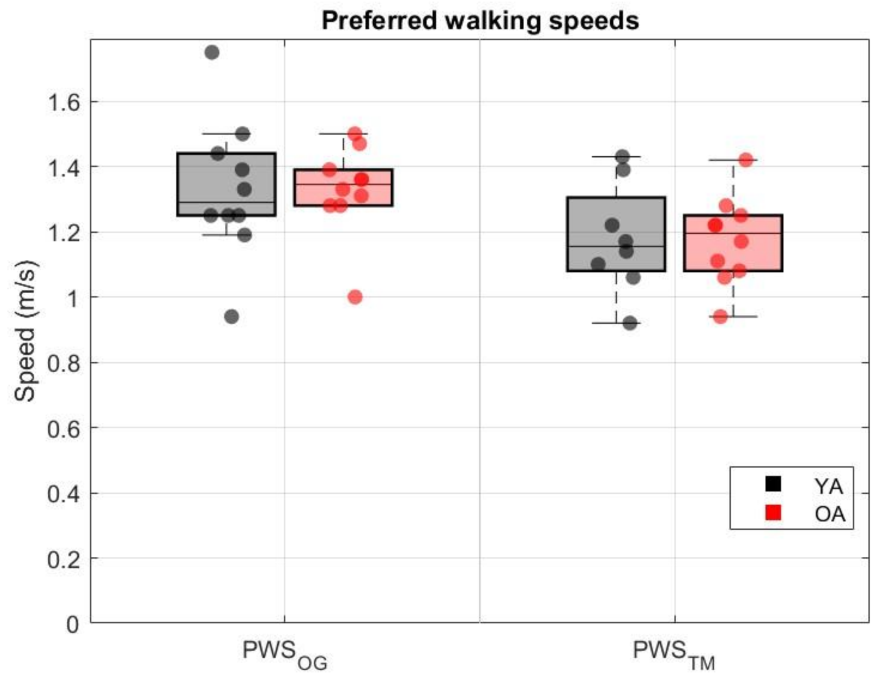

Figure 14C: Box plots showing the preferred walking speed overground (PWS_{OG}: left) and treadmill (PWS_{TM}: right) in m/s for the younger (YA) and older adults (OA).

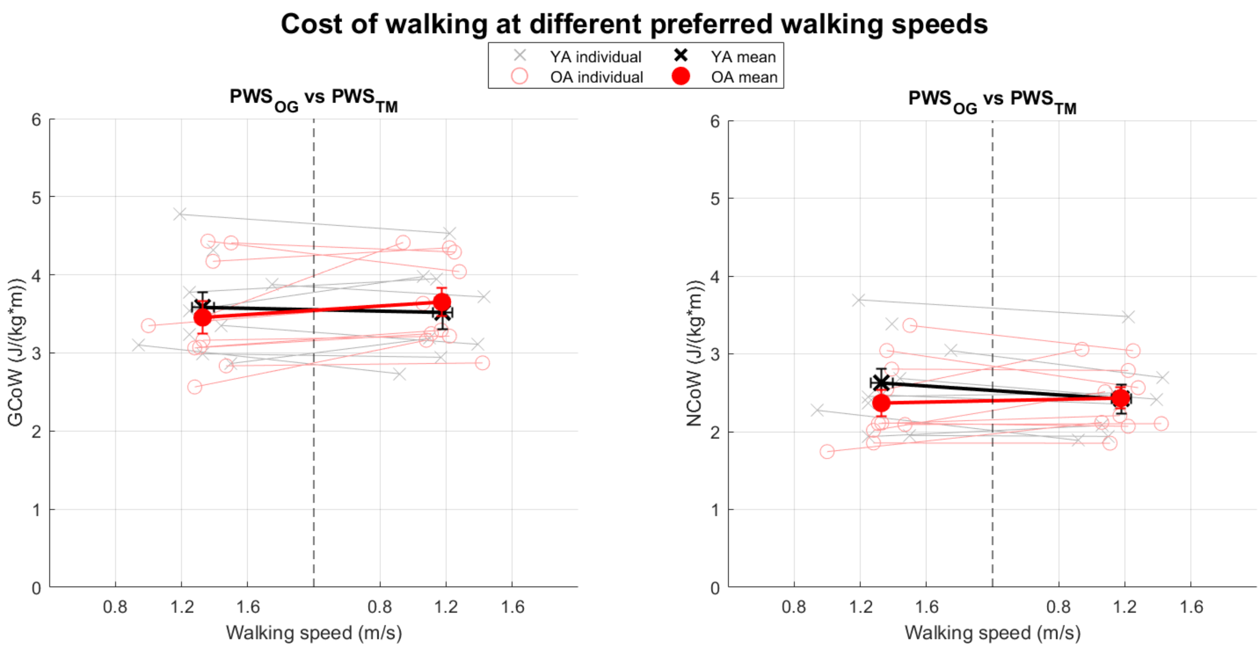

Figure 15C: Connected scatter plots showing the mean cost of walking and walking speed with error bars and the single subjects between the overground (PWSOG) versus treadmill preferred walking speed (PWS_{OG}) while walking on the treadmill in Gross (left) and Net (right) Cost of walking at their overground preferred walking speed (PWS_{OG}). The black crosses display the younger adults (YA) and the red circles the older adults (OA).

This report has been corrected and paraphrased with the aid of Grammarly and ChatGPT.