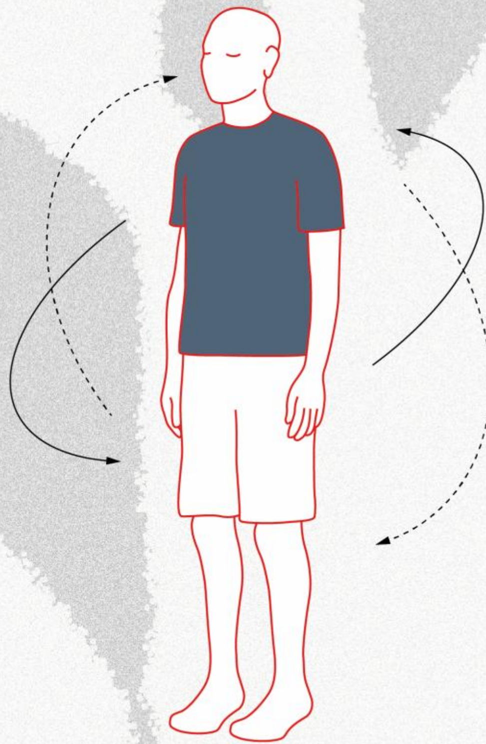
The metabolic cost of postural variability and vestibular contributions during standing balance



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The metabolic cost of postural variability and vestibular contributions during standing balance

By

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Abstract

Maintaining standing balance requires continuous sensorimotor control and is associated with a measurable metabolic cost. While minimization of energy usage is thought to influence human movements, it remains unclear how this applies to standing balance and how postural variability contributes to metabolic cost, because previous studies could not separate the effects of biomechanics from those of variability. This study aimed to quantify the metabolic cost of standing balance and to determine how this cost relates to postural variability independently of biomechanical alterations.

Twenty healthy participants performed seven standing conditions in which postural variability was manipulated using electrical vestibular stimulation (EVS) with varying frequencies and amplitudes. With EVS, postural responses were evoked without mechanically destabilizing the body. Metabolic cost was measured using indirect calorimetry, while postural variability was quantified using center of pressure (CoP) velocity and upper-body kinematics.

Free-standing resulted in an approximately 8% higher metabolic cost compared to externally supported standing, indicating the energetic cost of active balance control. Increasing postural variability through low-frequency EVS significantly elevated metabolic cost and revealed a positive relationship between variability and metabolic cost. In addition, vestibular stimulation and its reflexive muscle activity did not increase metabolic cost when postural variability remained unchanged. Furthermore, movements with comparable variability resulted in similar metabolic costs, regardless of whether they were reflexively induced or self-initiated.

These findings demonstrate that the metabolic cost of standing is primarily determined by postural variability rather than by vestibular input or reflexive muscle activity alone. The results support the hypothesis that humans adopt a preferred level of postural variability that is close to the metabolic minimum, consistent with principles of optimal control.

1. Introduction

Although maintaining standing balance may appear effortless, it requires continuous energy expenditure to counteract the downward pull of gravity and keep the body upright. Vestibular, visual and somatosensory inputs are integrated into an internal representation of body motion and orientation that generates corrective motor commands [1]. Since upright stance in humans is mechanically unstable, continuous sensorimotor feedback control is required to maintain balance. This instability, which can be increased by both external (e.g. unstable support surface) and internal (e.g. breathing) factors, creates postural sway during quiet standing [2], [3]. While it is known that the magnitude of sway is dependent on these factors, it remains unclear what determines the preferred sway magnitude during human standing.

Optimal control models suggest that a performance criterion, such as a minimization of energy use, is optimized in sensorimotor control of humans [4]. Previous research in movement science has shown that minimization of metabolic cost tends to be an important factor while performing movements [5], [6], [7]. For example, humans have a preferred step width during locomotion that minimizes their energy expenditure [7]. In addition, continuous adjustments are made to the characteristics (e.g. step frequency) of movements to find the lowest metabolic cost [5], [8]. In standing, this optimization strategy also appears to contribute to the balancing behaviour. Leeuwis et al. [9] showed that during standing, humans tend to adopt the anteroposterior lean angle that minimizes energetic costs. Consistent with this idea, Houdijk et al. [10] reported that an increase in postural sway was associated with increased metabolic cost, further supporting the theory of energy minimization during standing. In that study, however, postural sway was increased by inducing biomechanical changes, namely by having participants stand in tandem stance or on a balance board. As a result, it remained unclear whether the higher metabolic cost arose from the increased postural sway, the greater instability of postures, or both.

An alternative approach to evaluating the influence of postural variability on the metabolic cost of standing is to use targeted perturbations that avoid altering the biomechanics of standing. Electrical vestibular stimulation (EVS) offers such an approach by selectively targeting vestibular signals, thereby enabling controlled changes in both vestibular input and postural variability without mechanically destabilizing the body . EVS activates the vestibular afferents through electrodes placed on the mastoid bones behind the ears and is perceived as a virtual head movement, primarily along a roll rotation axis [11]. The body responds as if a real head movement has occurred and compensates for the virtual imbalance, causing whole-body sway in the opposite direction. EVS is characterized by its amplitude and frequency, and induces responses in standing participants that depend on the low-pass filter properties of the human body. Specifically, while muscle activity can be reliably evoked over a wide stimulation bandwidth (0 - 20 Hz), changes in ground reaction forces and body sway are observed up to 10 Hz and 2 Hz, respectively [12]. This indicates that higher-frequency components are attenuated at the level of whole-body motion and that the frequency characteristics of the stimulus can be used to target different reactions of the body. In addition to the frequency characteristics of the EVS signal, its amplitude determines the magnitude of the evoked responses, with larger amplitudes producing greater muscle activity [13]. Despite the EVS-evoked muscle activity and postural responses, vestibular reflexes are absent when the state of balance is sensed by the somatosensory system through external support [14], suggesting that vestibular feedback is suppressed when muscle output is no longer relevant to maintaining balance [15].

The aim of this study was to quantify the metabolic cost of standing balance and to establish how that cost relates to postural variability. First, the cost of standing balance was calculated by comparing metabolic rates of free-standing and externally supported standing, where balance-correcting muscle activity is removed. Next, by using low-frequency EVS across increasing amplitudes, postural variability was manipulated without changing the biomechanics of the body, extending earlier approaches that could not manipulate variability independently of mechanics [10]. With this approach,

an isolated correlation between postural variability and metabolic cost could be made. To assess whether variability itself drives the metabolic cost or whether other factors account for the increased cost, three targeted comparisons were made. First, to isolate the metabolic cost of the vestibular activity evoked by the stimulation, the metabolic cost of EVS was assessed when participants were externally supported. Under these conditions, postural responses to EVS are suppressed such that there should be no added metabolic cost due to the reaction of the body to EVS [14]. Second, to investigate the metabolic cost of EVS-evoked muscle reflexes, participants were exposed to high-frequency EVS, as under these conditions muscle reflexes were expected without increasing postural variability [12]. Third, to assess whether the origin of variability influences its metabolic cost, energetic cost of EVS-evoked reflexive movements were compared with energetic cost of self-initiated movements, in which it was needed to achieve a matched postural variability. Following these targeted comparisons, the correlation between postural variability and metabolic cost was shown to be unconfounded, after which a comparison with the correlation in the study of Houdijk et al. [10] showed that their correlation was confounded by the induced biomechanical changes.

2. Methods

2.1 Participants

A total of 20 healthy participants were enrolled in the study. Exclusion criteria were a history of neurological or musculoskeletal disorders and an age under 18 years or over 60 years. Also, participants with pain or balance issues during standing were excluded. To test (self-reported) sex differences in variability and metabolic cost, 10 female and 10 male participants were included. Mean age was 27.2 ± 6.1 years, height 181 ± 10 cm, weight 77.0 ± 13.1 kg. The study was approved by the Erasmus Medical Center and all participants received the information letter for the experiment at least two days before participating. Prior to the experiment, written consent was obtained from all participants after explaining the experiment protocol.

2.2 Experimental setup

During the experiment, EVS was applied to evoke postural variability while participants were standing. Simultaneously, participants' energy expenditure was measured with a metabolic analyser. A motion capture system, together with a force plate on which the participant stood, was used to keep track of postural variability. To quantify muscle activation in the lower legs, EMG was measured from four different lower leg muscles. For trials in which participants were externally supported to offload balance-correcting muscle activity, a backboard was taken into place. The experimental setup can be seen in figure 1A and figure 1B and is discussed in detail below.

2.2.1 Electrical vestibular stimulation

Vestibular afferents were activated by applying EVS through two rubber electrodes with conductive gel (Spectra 360, Parker, USA). The electrodes, acting as a cathode and an anode, were placed binaural at the mastoid process and secured with the help of tape (3M Durapore, Solventum, USA) and an elastic headband (Elastofix, BSN medical, Germany). Three stochastic stimuli were created; one with a low frequency (0-2 Hz) and a low peak amplitude (1.5 mA), one with a low frequency (0-2 Hz) and a high peak amplitude (3.0 mA), and one with a high frequency (4-20 Hz) and a high peak amplitude (3.0 mA). The frequencies were chosen based on the study of Dakin et al. [12], which demonstrated that EVS induces body sway primarily between 0 and 2 Hz, while lower limb muscle activity in response to the EVS can be observed at the frequencies up to 20 Hz. The peak amplitudes were chosen based on pilot studies in which 3.0 mA was sufficient to generate an approximate doubling of postural variability without periods of instability, and on previous observations demonstrating that the peak amplitude shows a correlation with sway magnitude and muscle activity [13]. Furthermore, to get enough deviation from 0 mA through the signal, the standard deviation had to be above 44% of the peak amplitude [16]. The signal was generated in MATLAB (R2023b) and sent with LabView (2019b) via a D/A board (National Instruments, BCN-2110, TX, US) which was also used to trigger the measurements on all other equipment, except the metabolic analyser. EVS was delivered to the participant through a constant current stimulator (DS5, Digitimer, UK).

2.2.2 Metabolic analyser

Metabolic cost was measured using an indirect calorimetry system (system: MS-CPX, CareFusion, USA; wearable device: SBx/CPX, CareFusion, USA; mask: V-982185, Vyaire Medical, USA). Participants wore a metabolic analyser attached to a harness at their waist that was connected to a face mask where flow and gas analysis was measured. The mask covered their mouth and nose to capture all air that was breathed in and out. It was checked that no air could leak out through the sides of the mask, by ensuring breathing was impossible while covering the main outlet. Through gas analysis, oxygen consumption and carbon dioxide production was measured, after which metabolic cost was calculated breath-by-breath using the Weir equation [17]. Before the start of an experiment,

the system was calibrated by using the manufacturer's guidelines, which included a flow calibration and a gas analyser calibration, using calibration gas (15.94% O_2, 5% CO_2, Vyaire Medical, USA). Measurements of the metabolic analyser were not synchronized with the other measurement equipment and were started simultaneously by hand.

2.2.3 Motion capture

Reflective markers (16) were placed on anatomical landmarks following the simplified marker set by Tisserand et al. (13 markers) [18], with additional markers placed on the second metatarsal and calcaneus landmarks (4 markers). The sacral marker on the back of the participant was excluded, as it turned out to be not visible for the cameras because of the backboard. Eight cameras (4 Miqus M3 and 4 Miqus M5, Qualisys AB, Göteborg, Sweden) running at 100 Hz registered the location of the reflective markers in 3D. The motion capture system also consisted of a force plate on which the participant was standing, capturing ground reaction forces and moments at 1000 Hz (Plate: BMS 400600HF-1K, AMTI, MA, US; Amplifier: Optima OPT-SC, AMTI, MA, US). Participants could see a visual representation of their center of pressure (CoP) position presented on a TV screen 3.2 m in front of them. The screen also displayed the anteroposterior angle limits in which CoP had to be maintained. Every morning before experiments started, calibration of the motion capture system was done by following the manufacturer's instructions. An analog measurement board (Qualisys Analog Interface 16 Channels, 230597, Qualisys AB, Göteborg, Sweden) and a sync box (Qualisys Sync Box, 410850, Qualisys AB, Göteborg, Sweden) were used to synchronize the motion capture and force data with the EVS and electromyography measurements.

2.2.4 Electromyography

Muscle activity was measured with electromyography (EMG) of the soleus (SOL) and gastrocnemius medialis (mGAS) bilaterally through electrodes (BlueSensor, Ambu) placed on the lower legs. These muscles were known to respond robustly to the EVS signal during standing [19]. The ground electrode was placed on the right lateral malleolus. Before the EMG signals were synchronized with the motion capture data with the use of the analog measurement board, they were pre-amplified (NL844 Pre-amplifier, Digitimer) and filtered (NL135 filter, NL820 isolator, Digitimer).

2.2.5 Supportive backboard

A custom-made adjustable backboard was installed behind where the participant stood to fix their position in a subset of trials. The anteroposterior angle of the backboard could be adjusted with the help of a hinge installed on a linear guide, and was fixed in a position that matched the participant’s quiet standing angle, as checked with live force plate data. When this position was found (i.e. participant was not pushed forward and did not have to lean backward to feel supported), the participant was strapped to the backboard with two belts, one just below shoulder height, and one on pelvis height. This position allowed the participant to breathe effortlessly and to maintain their natural posture without requiring any active control to remain upright.

2.3 Protocol

2.3.1 Preparation

Participants were asked to avoid caffeinated products or large meals within 3 hours before testing took place, as this could influence metabolic cost [20], [21]. After arriving at the Erasmus Medical Center, participants first tried both low- and high-frequency EVS with two incrementally increasing levels (1.5mA to 3mA) while sitting to familiarize with the signal and assess whether they experienced any nausea or dizziness. If the stimulation was unpleasant or too disorienting, participants did not participate in the experiment. Three potential participants out of 23 screened participants chose not to start the experiment for this reason. Uncomfortable sensations behind the ears were addressed by replacing the electrodes and ensuring no excess gel was exposed from under the electrode.

Height, weight and ankle height were manually measured. Ankle height was calculated by taking the mean of the inside and outside left malleolus height and feet were placed 2 cm apart on the force plate. The anteroposterior placement of the feet on the force plate was consistent between participants such that the anterior edge of the medial malleolus was at the same location, as the center of rotation of the ankle joint was assumed to be there [22]. Subsequently, anteroposterior body angle was determined by information retrieved from ankle height, body height, gender and live CoP data, with Formula 1.

$$\theta = \cos \frac{CoP_{AP}}{(h_b * c_g) - h_a} \qquad (Formula\ 1)$$

Here, θ is the anteroposterior body angle, CoP_{AP} is the live anteroposterior location of the Center of Mass (CoM), which is assumed to be straight above the CoP. h_b is the body height, h_a is the height of the ankle. c_g is the gender coefficient, since woman and men typically have their CoM at their body height multiplied by 0.56 and 0.57 times respectively [23].

In addition to their feet position, participants were also instructed to maintain a specific head position throughout the experiment. Since EVS evokes a virtual rotation around an axis that is 17.5 degrees posterior and superior to Reid's plane [11], [24], the head was tilted upward by this angle, resulting in a whole-body sensation of roll only. To get the correct head angle, the head was aligned using a custom-made level with a 17.5 degree angle, after which a laser pointer directed forward was attached to the headband of the metabolic analyser with Velcro. The participant maintained the head orientation by keeping the laser on a piece of tape on the wall that corresponded to the correct head angle. As participants performed the experiment with their eyes closed, the experimenter would provide short instructions during a trial if the head deviated too far from the target. Lastly, the mask of the metabolic analyser was placed over the mouth and nose and participants were instructed to breathe calmly through the mask.

2.3.2 Trials

Participants performed 7 trials, each consisting of a 3-minute standing task in which they were asked to stand relaxed and close their eyes within the first 30 seconds of each trial. This period ensured that participants could briefly acclimatize to the stimulation and was left out of all data analysis, as visual feedback plays an important role in maintaining human balance [25]. The anteroposterior body angle, which was measured with the help of the force plate, had to be between 0.5 and 2.5 degrees forward, as this range ensured consistency and corresponds to the energetically optimal and naturally preferred standing posture [9]. When being outside these limits, a sound was played to indicate that participants should move back inside the limits. When the laser deviated from the original place, participants were told to tilt their head slightly upward or downward. Halfway through all trials, participants rested by sitting and taking off the mask of the metabolic analyser. After every trial, participants could indicate they wanted a break, and could choose to take off the mask if desired.

Seven conditions were executed in randomized order (Figure 1C), each designed to enable comparisons addressing specific questions. The first was to determine the metabolic cost of maintaining standing balance, for which a comparison was made between a baseline condition (Baseline) and an externally supported condition (Immobile). In Baseline, participants stood freely without EVS, representing quiet standing behaviour. In Immobile, participants were externally supported without EVS, thereby removing balance-correcting muscle activity. The difference in metabolic cost between the conditions was used to isolate the energetic cost of standing balance.

To investigate the relationship between postural variability and metabolic cost, variability was manipulated using low-frequency EVS during free-standing. Participants were exposed to EVS with low frequency (0-2 Hz) and a peak amplitude of 1.5 mA (Low-freq 1.5mA), which was expected to increase postural variability compared to Baseline [12]. A similar low-frequency stimulus was applied with a peak amplitude of 3 mA (Low-freq 3mA), which was expected to further increase postural

variability [13]. Together with Baseline, these two conditions were used to assess the relationship between postural variability and net metabolic cost.

To determine whether metabolic cost could be attributed to the vestibular stimulation, an externally supported condition with EVS (Immobile low-freq 3mA) was included. Participants were exposed to EVS with low frequency (0-2 Hz) and a peak amplitude of 3 mA, as this stimulus was expected to evoke the largest responses during free-standing. However, during externally supported standing, vestibular reflexes are suppressed and no postural responses were expected [14]. Therefore, any difference in metabolic cost (gross and net) between Immobile low-freq 3mA and Immobile could be attributed to the vestibular stimulation itself.

Reflexive muscle activity was also evaluated as a potential contributor to metabolic cost, independent of postural variability, by exposing participants to EVS with high frequency (5–20 Hz) and a peak amplitude of 3 mA (High-freq 3mA) during free-standing. Under these conditions, coherent muscle activity was expected without an increase in postural variability compared to Baseline [12]. Comparing this condition to Baseline allowed assessment of the metabolic cost (gross and net) associated with reflexive muscle activity in the absence of increased variability.

Finally, to examine whether the origin of postural variability influences its metabolic cost, a self-initiated movement condition (Self movement) was included, which always took place right after Low-freq 3mA. During Self movement, participants stood freely without EVS while mimicking the movements experienced during the previous trial of Low-freq 3mA. By comparing Self movement with Low-freq 3mA, it was possible to assess whether similar postural variability results in different metabolic costs (gross and net) depending on whether it is reflexively induced or voluntarily generated.

2.4 Data analysis

Data from all measurement equipment was collected through Qualisys Track Manager (2024.2) (motion capture, force plate, EVS and EMG) and JLab Lab Manager (V5.32.0) (metabolic analyser) and was processed and analysed with MATLAB (R2023b) as described below.

2.4.1 Time cutting

Data of the first 30 seconds of each 3-minute trial were discarded, as participants were allowed to keep their eyes open during this period to accommodate to the condition.

For the metabolic data, the first 60 seconds of every trial were discarded to ensure a steady-state measurement. Breath-by-breath indirect calorimetry measurements are actually not real-time measurements due to slow mitochondrial dynamics and long transmit times from tissue to the lungs [26], and have a time constant of around 42 seconds during gait [26]. A conservative period of 60 seconds was used for our standing experiments, aligning with previous work [9]. Together with the aforementioned time constant, it was estimated that a 30-second difference between metabolic data and the rest of the data would be realistic. However, at the end of every trial, all data was cut at the 3-minute mark as it was assumed that all body's responses were at steady-state at that point.

2.4.2 Variability metrics

Two metrics were used to quantify standing variability; mean CoP velocity and mean velocity of left and right acromion motion capture marker (ACR velocity). Similar to Houdijk et al. [10], the distance travelled by the CoP in the X–Y plane was calculated and divided by trial duration to obtain mean velocity, allowing comparison across studies independent of trial length. First, the CoP data was low-pass filtered at 20 Hz with a zero-phase 2nd-order Butterworth filter. Then, the distance travelled by the CoP between every data point was calculated using Pythagorean theorem and summed to find the total travelled distance of the CoP. This was divided by the amount of seconds in a trial to get the mean CoP velocity in mm/s.

The ACR velocity (mm/s) was used to quantify upper-body kinematic variability. Processing in Qualisys Track Manager allowed the trajectory data to have a prediction error of 20 mm (maximum inter-frame marker deviation) and a maximum residual of 5 mm (maximum camera ray intersection distance) during tracking. Gaps ≤100 ms in marker data were filled with polynomial interpolation. The distance travelled in 3D by both the left and right acromion motion capture markers were calculated separately and averaged. The mean distance was divided by the amount of seconds in a trial to get the mean ACR velocity in mm/s.

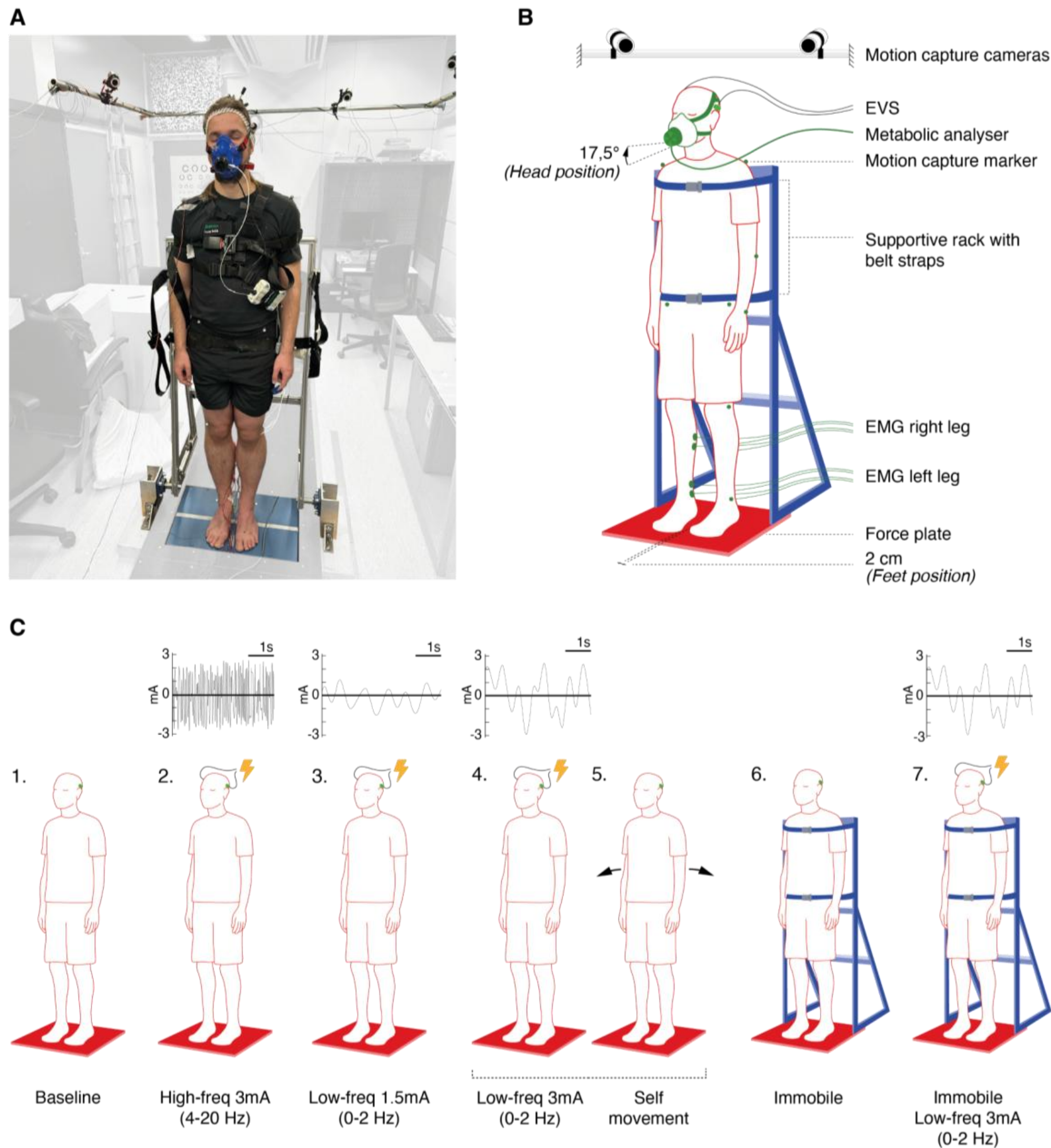


*Figure 1: (A) Photograph during pilot study with participant and experimental setup highlighted. (B) Schematic overview of participant with experimental setup and postural configuration. (C) Schematic overview of the 7 tested conditions with their abbreviations and EVS specifications. **Baseline**: a baseline condition in which participants stood freely without EVS. **High-freq 3mA**: free-standing, while being exposed to EVS with high frequency (5-20 Hz) and high peak amplitude (3 mA). **Low-freq 1.5mA**: free-standing, while being exposed to EVS with low frequency (0-2 Hz) and low peak amplitude (1.5 mA). **Low-freq 3mA**: free-standing, while being exposed to EVS with low frequency (0-2 Hz) and high peak amplitude (3 mA). **Self movement**: free-standing without EVS, while mimicking the movement of Low-freq 3mA. Self movement always took place right after Low-freq 3mA. **Immobile**: externally supported standing without EVS. **Immobile low-freq 3mA**: externally supported standing, while being exposed to EVS with low frequency (0-2 Hz) and high peak amplitude (3 mA).*

2.4.3 Metabolic measurements

The metabolic cost was calculated using breath-by-breath data that was collected in the second and third minute of the trial, since the first 60 seconds were discarded as described before. Metabolic cost (W) was calculated with the Weir equation [17] for every breath by the metabolic analyser, after which power per kilogram (W/kg) was determined by dividing it by the mass of a participant. After the metabolic cost for each breath was normalized for breathing duration, an average metabolic cost was determined, which was called gross metabolic cost. By subtracting the metabolic cost of the immobile condition without EVS for each participant individually, net metabolic cost was determined.

2.4.4 EVS coherence

To quantify the frequency-dependent relationship between the applied EVS signal and the resulting mechanical and muscular responses, coherence was calculated between the EVS input and (i) mediolateral force output and (ii) surface EMG activity.

First, the EMG signals were high-pass filtered at 10 Hz, full-wave rectified, and subsequently low-pass filtered at 100 Hz, using zero-phase 2nd-order Butterworth filters. The EVS, EMG and force signals were then cut in segments of two seconds. For each segment, signals were transformed to the frequency domain using a fast Fourier transform. Auto-spectra of the signals were obtained by averaging the squared magnitudes of the Fourier coefficients across segments. Cross-spectra between the EVS signal and the output signals were computed as the average of the product of the complex conjugate of the EVS spectrum and the spectrum of the output signal. The zero-frequency component was excluded from all spectra, and pooling spectra was performed prior to calculating the coherence using Formula 2.

$$C_{xy}(f) = \frac{|S_{xy}(f)|^2}{S_{xx}(f)\, S_{yy}(f)} \quad \textit{(Formula 2)}$$

Here, $S_{xx}(f)$and $S_{yy}(f)$ are the auto-spectra of the EVS signal and the output signal respectively, and $S_{xy}(f)$ is the corresponding cross-spectrum. Coherence between EVS and EMG, and between EVS and force, was computed for conditions in which there was vestibular stimulation, to investigate whether the human body reacted to the EVS with corresponding frequencies. Coherence levels above which coherence was considered significant were calculated using Formula 3 [27].

$$1 - (1 - \alpha)^{1-(L-1)} \quad \textit{(Formula 3)}$$

Where α is the significance level used in this study (0.05) and L is the total amount of two-second segments per condition.

2.4.5 Houdijk comparison

Lastly, the current study was compared with the study of Houdijk et al. [10], in which postural variability during standing balance was increased by changing biomechanics during standing. For both studies, linear regression plots were made between postural variability and metabolic cost. Similar to the metrics in the current study, Houdijk's study provided gross metabolic cost (W/kg) and total CoP distance, which was converted to mean CoP velocity (mm/s), for four different conditions (e.g. baseline, tandem stance (feet in front of each other), tandem stance blindfolded and tandem stance on a balance board). For the comparison with our study, Baseline and both Low-freq EVS conditions were used, as these were the conditions in which postural variability was expected. The mean values of the metrics per condition were taken, as these were the statistics available from the study of Houdijk et al. [10]. The difference in slope coefficient between both linear regression plots (postural variability vs gross metabolic cost) provided an estimate of how much the biomechanical changes added to the increased metabolic cost in the study of Houdijk et al. [10].

2.5 Statistical analysis

To assess differences between sexes for metabolic measurements as well as variability measurements, a two-way repeated-measures ANOVA was performed with self-reported sex as a between-subject factor and condition as a within-subject factor.

Afterwards, when two conditions were compared, pre-planned paired samples t-tests were used to analyse differences for metabolic measurements and/or variability measurements. To check whether normal distribution criteria were met, Kolmogorov-Smirnov tests were computed before t-testing for all conditions on metabolic and variability data. Paired samples t-tests were executed for (i) gross metabolic cost for Baseline vs Immobile, (ii) gross metabolic cost for Immobile vs Immobile low-freq 3mA, (iii) gross metabolic cost and CoP velocity for Baseline vs High-freq 3mA and (iv) gross metabolic cost and CoP velocity for Low-freq 3mA vs Self movement. For the comparison between Baseline, Low-freq 1.5mA and Low-freq 3mA, a repeated-measures ANOVA was performed to test for differences in net metabolic cost, CoP velocity and ACR velocity. Post-hoc analysis with Holm correction was conducted when a significant main effect was found. It was decided to perform planned comparisons between specific conditions using paired samples t-tests and (a small) ANOVA as opposed to an omnibus ANOVA analysis as each condition was designed to address a specific question. For all comparisons mentioned above, gross and net metabolic cost yield identical significance, as the within-subject differences do not change between both metrics.

The correlation between variability and net metabolic cost was assessed using linear regression, restricted to Baseline, Low-freq 1.5mA and Low-freq 3mA, as these conditions were expected to show increased variability. The linear mixed effects model was used for the regression analysis, as this takes repeated measures into account. Lastly, for coherence measurements, confidence limits were calculated to determine whether coherence between the EVS signal and (i) the EMG signal and (ii) the ground reaction forces existed. For all statistical analyses, statistical significance was defined as $p < 0.05$. Statistical analysis was performed using JASP 0.95.4.0 and MATLAB (R2023b).

3. Results

All participants completed the seven three-minute trials in randomized order, while being instructed to stand relaxed with eyes closed and head slightly tilted upward. Participants were stimulated with EVS during four of the trials, while data from the metabolic analyser, motion capture, force plate and EMG was captured continuously during all trials.

For all conditions, gross metabolic cost, net metabolic cost, CoP velocity and ACR velocity was calculated. Kolmogorov-Smirnov tests indicated that all metrics for all conditions that were used in this study were found to be normally distributed *($p > 0.05$)*, except for the variability metrics during self-initiated movement. As this was the only condition that did not pass the normality test and due to the robustness of these parametric tests, it was chosen to continue with the pre-planned t-tests. For every condition, mean values with standard deviations of all metrics are shown in Table 1.

Table 1: Average and standard deviation of all participants for metabolic and variability metrics, separated per condition. The immobile condition does not have a net metabolic cost, as it is defined as zero.

	BASELINE	HIGH-FREQ 3MA	LOW-FREQ 1.5MA	LOW-FREQ 3MA	SELF	IMMOBILE	IMMOBILE LOW-FREQ 3MA
GROSS METABOLIC COST (W/KG)	1.56 ± 0.18	1.53 ± 0.10	1.65 ± 0.19	1.69 ± 0.26	1.68 ± 0.22	1.44 ± 0.18	1.45 ± 0.16
NET METABOLIC COST (W/KG)	0.11 ± 0.14	0.09 ± 0.10	0.20 ± 0.15	0.25 ± 0.21	0.23 ± 0.18	-	0.01 ± 0.09
COP VELOCITY (MM/S)	15.6 ± 4.08	16.7 ± 3.38	28.3 ± 11.5	35.6 ± 16.9	42.5 ± 25.7	3.23 ± 2.17	3.10 ± 1.04
ACROMION VELOCITY (MM/S)	10.3 ± 2.68	10.6 ± 2.06	17.0 ± 7.76	19.4 ± 11.3	27.5 ± 14.8	4.34 ± 1.38	4.86 ± 3.12

Repeated-measures ANOVA revealed that (self-reported) sex had no significant influence on all of the measured metrics (see Table 2). Also, no interaction was found between conditions and sexes. And lastly, as expected, it was shown that the different conditions had a significant effect on all metrics *($p < 0.001$)*.

Tabel 2: Two-way repeated-measures ANOVA with condition as a within-subject factor and (self-reported) sex as a between-subject factor for all metabolic and variability metrics. Degrees of freedom F-statistic: $^{a}(1,18)$ $^{b}(6,108)$ $^{c}(5,90)$.

	GROSS METABOLIC COST	NET METABOLIC COST	COP VELOCITY	ACR VELOCITY
SEX	$F^a = 0.830, p = 0.374$	$F^a = 1.137, p = 0.300$	$F^a = 0.110, p = 0.744$	$F^a = 0.471, p = 0.501$
CONDITION	$F^b = 15.64, p < 0.001$	$F^c = 12.27, p < 0.001$	$F^b = 32.64, p < 0.001$	$F^b = 25.66, p < 0.001$
CONDITION *SEX	$F^b = 0.359, p = 0.903$	$F^c = 0.257, p = 0.935$	$F^b = 0.210, p = 0.973$	$F^b = 0.346, p = 0.911$

3.1 Metabolic cost of standing balance

We next examined the difference in metabolic cost between the free-standing baseline condition and the immobile condition to determine the cost of keeping the body in a balanced state during standing. Since immobile standing offloads most of the balance-correcting muscle activity, this condition was used for the comparison. A significant difference was found in gross metabolic cost (identical significance for net metabolic cost) between Baseline and Immobile *(8%, $t(19) = 3.780$, $p = 0.001$)*, which is shown in Figure 2. The difference, 0.11 W/kg, can be interpreted as the amount of energy it requires to keep the body balanced against gravity's downward pull during standing.

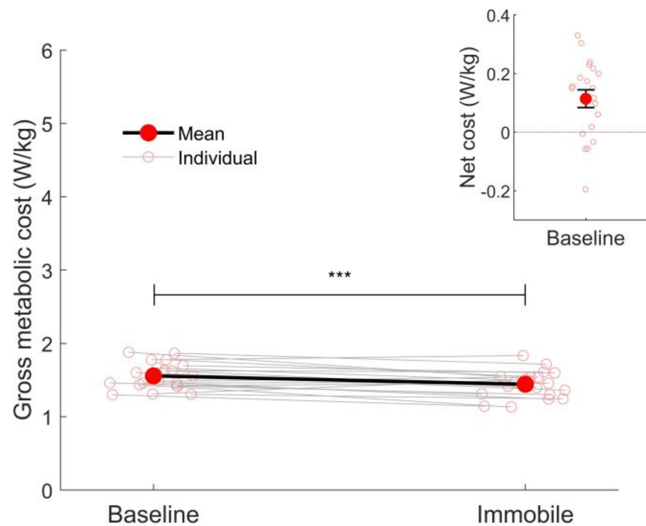


*Figure 2: Paired data plot showing mean and individual data of gross metabolic cost of Baseline and Immobile. Net metabolic cost of Baseline is shown with error bar in top right. Since net metabolic cost is the gross metabolic cost minus the metabolic cost of the immobile condition, both figures have the same meaning and significance. Labels: ***: $p \leq 0.001$.*

3.2 Postural variability vs metabolic cost

To analyse the relationship between postural variability and metabolic cost, comparisons between both Low-freq EVS conditions (1.5 mA and 3 mA) and Baseline were made for net metabolic cost, CoP velocity and ACR velocity (Figure 3). Repeated-measures ANOVA revealed there was an effect of condition on the net metabolic cost *($F(2,38) = 7.068$, $p = 0.002$)*. Post-hoc analysis showed a significant increase in net metabolic cost between Baseline and Low-freq 1.5mA *(79%, $t(19) = -3.115$, $p_{holm} = 0.011$)* as well as between Baseline and Low-freq 3mA *(115%, $t(19) = -3.333$, $p_{holm} = 0.010$)*. No difference was found between Low-freq 1.5mA and Low-freq 3mA *($t(19) = -1.071$, $p_{holm} = 0.298$)* (Figure 3A).

For variability, the metric based on the CoP velocity, as measured on the force plate, was evaluated first. Repeated-measures ANOVA revealed there was an effect of condition on the CoP velocity *($F(2,38) = 18.16$, $p < 0.001$)*, after which post-hoc analysis showed a significant increase in CoP velocity between Baseline and Low-freq 1.5mA *(81%, $t(19) = -6.102$, $p_{holm} < 0.001$)*. An increase was found between Baseline and Low-freq 3mA as well *(128%, $t(19) = -5.169$, $p_{holm} < 0.001$)*, while no difference was found between Low-freq 1.5mA and Low-freq 3mA *($t(19) = -1.918$, $p_{holm} = 0.070$)* (Figure 3B). Consistent with Dakin et al. [12], low-frequency EVS induced postural variability, although doubling the peak amplitude did not result in a further significant increase.

Variability based on ACR velocity showed the same pattern, as repeated-measures ANOVA revealed there was an effect of condition on the ACR velocity *($F(2,38) = 9.196$, $p < 0.001$)*. Post-hoc analysis showed a significant increase in mean ACR velocity between Baseline and Low-freq 1.5mA *(65%, $t(19) = -4.968$, $p_{holm} < 0.001$)* as well as between Baseline and Low-freq 3mA as well *(89%, $t(19) = -3.469$, $p_{holm} = 0.005$)*. No difference was found between Low-freq 1.5mA and Low-freq 3mA *($t(19) = -1.017$, $p_{holm} = 0.322$)*, indicating consistency between the two variability metrics.

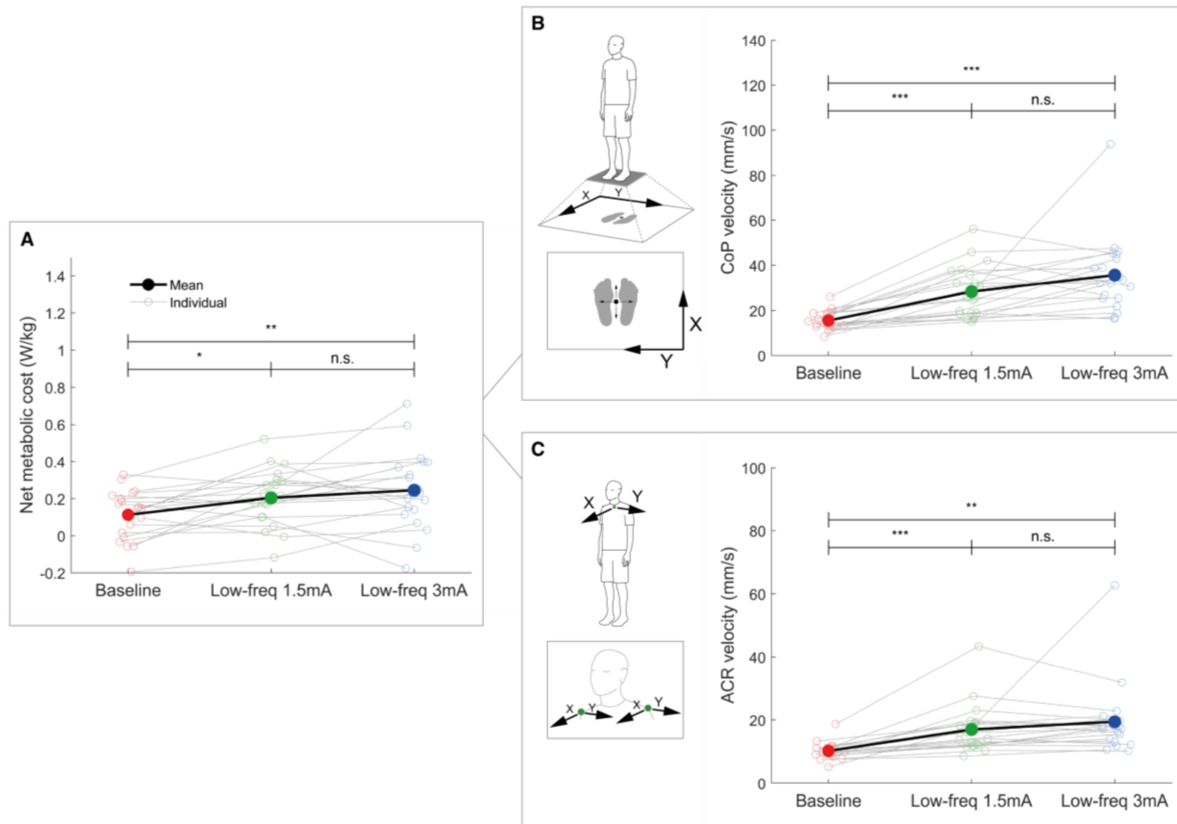


*Figure 3: Paired data plots showing mean and individual data of (A) net metabolic cost, (B) mean CoP velocity and (C) mean acromion marker velocity during Baseline and both Low-freq EVS conditions. Labels: n.s.: not significant, *: $p \leq 0.05$, **: $p \leq 0.01$, ***: ≤ 0.001, ACR: acromion.*

After an increase from Baseline to the Low-freq EVS conditions was determined for net metabolic cost, CoP velocity and ACR velocity, a linear regression was conducted with data from postural variability and net metabolic cost. For both metrics of variability, a separate linear mixed effects model was performed, which included data of all trials from Baseline, Low-freq 1.5mA and Low-freq 3mA (Figure 4). There was a significant positive correlation between CoP velocity and net metabolic cost *($\beta = 0.007 \pm 0.0009$, $t(58) = 7.894$, $p < 0.001$)* (Figure 4A) as well as between ACR velocity and net metabolic cost *($\beta = 0.011 \pm 0.0016$, $t(58) = 6.765$, $p < 0.001$)* (Figure 4B). Since both variability metrics showed similar t-values in regression plots and p-values between conditions, it was decided to continue only with CoP velocity as a variability measurement.

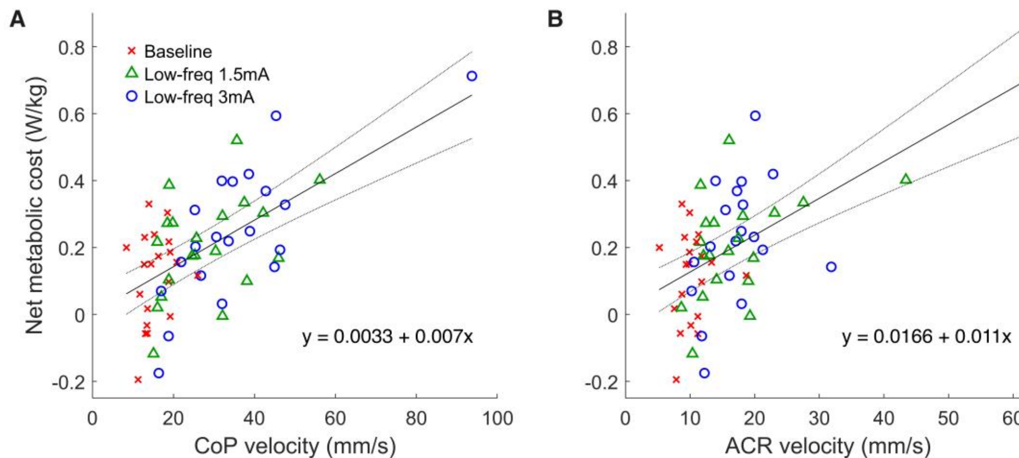
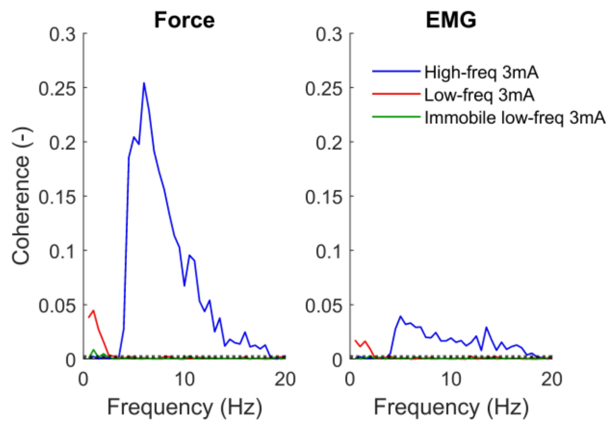


Figure 4: Linear regression plots with confidence intervals (p = 0.05) showing the correlation between net metabolic cost and (A) mean CoP velocity ($\beta = 0.007 \pm 0.0009$, $t(58) = 7.894$, $p < 0.001$) and (B) mean acromion marker velocity ($\beta = 0.011 \pm 0.0016$, $t(58) = 6.7652$, $p < 0.001$). All individual trials from the Baseline and both Low-freq EVS conditions are included. The outlier on the right upper side of both figures was caused by a participant who exhibited high levels of postural sway but performed the task as intended and did not lose balance. A statistical check revealed that excluding this participant would have yielded similar p-values and slopes ($\beta = 0.007 \pm 0.0012$, $t(57) = 5.948$, $p < 0.001$, for CoP velocity | $\beta = 0.013 \pm 0.0026$, $t(57) = 4.853$, $p < 0.001$ for ACR velocity) in the linear regression. Labels: ACR: acromion.

3.3 Metabolic cost of vestibular contributions

Having identified a correlation between standing variability and metabolic cost, we further assessed whether variability itself drives this metabolic cost or whether other factors account for the increased metabolic cost. The vestibular stimulation itself or EVS-induced muscle responses could be other reasons for an elevated cost. First, for (i) High-freq 3mA, (ii) Low-freq 3mA and (iii) Immobile low-freq 3mA, coherence was calculated between the EVS signal and (i) mediolateral force and (ii) EMG of the right gastrocnemius medialis (Figure 5), as this muscle showed the largest coherence in previous studies [19]. Coherence for both soleus muscles and the left gastrocnemius medialis was comparable and is shown in Appendix C. Coherence calculations assessed whether there were corresponding bodily reactions to the EVS. As expected during external support [14], EVS-evoked responses were absent in Immobile Low-freq 3mA (Figure 5). With this established, it was determined whether the vestibular stimulation itself drove the metabolic cost by comparing the gross metabolic cost between both immobile conditions, one with and one without EVS (Figure 6). Since there was no coherence found with the EVS signal, difference in metabolic cost between both immobile conditions could only be explained by the cost of EVS itself. However, no significant difference was found in metabolic cost (gross and net) *($t(19) = -0.520$, $p = 0.609$)*, indicating that vestibular stimulation itself did not account for the increased metabolic cost during the free-standing low-frequency EVS conditions.



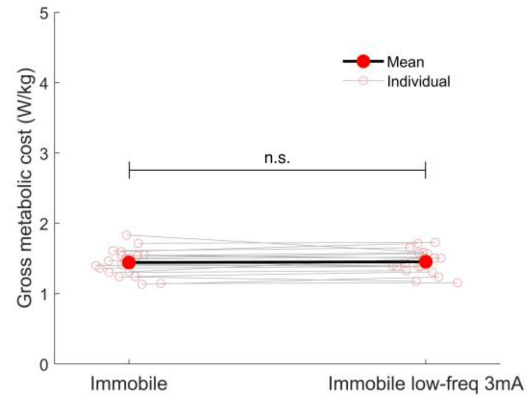


Figure 5: Coherence between EVS and (i) mediolateral force and (ii) EMG of the right gastrocnemius medialis for High-freq 3mA, Low-freq 3mA and Immobile Low-freq 3mA. The horizontal dotted lines (coherence = 0.002) in the plots represent the level above which the coherence is significant.

Figure 6: Paired data plot showing mean and individual data of gross metabolic cost of immobile conditions with and without EVS. Labels: n.s.: not significant.

After it was established that the vestibular activity created by the EVS signal itself had no significant influence on the metabolic cost, a comparison was made between Baseline and High-freq 3mA (Figure 7). Contrary to the immobile condition, bodily responses corresponding to high-frequency EVS have been reported by Dakin et al. [12] and were also found in the current study, as High-freq 3mA coherence was continuously above significance levels in the region between 4 and 19 Hz, for both EMG and force (see Figure 5). Similar to Dakin et al. [12], this bodily response to the high frequency EVS did not induce a significant increase in postural sway, measured in CoP velocity *($t(19) = -1.450$, $p = 0.163$)* (Figure 7A). To investigate whether the bodily muscle response to the EVS would increase the metabolic cost, a metabolic cost comparison was made. No significant difference metabolic cost (gross and net) was found *($t(19) = 0.848$, $p = 0.407$)* (Figure 7B), meaning that the EVS-evoked muscle reflexes were not the reason for the increased metabolic cost during the free-standing low-frequency EVS conditions.

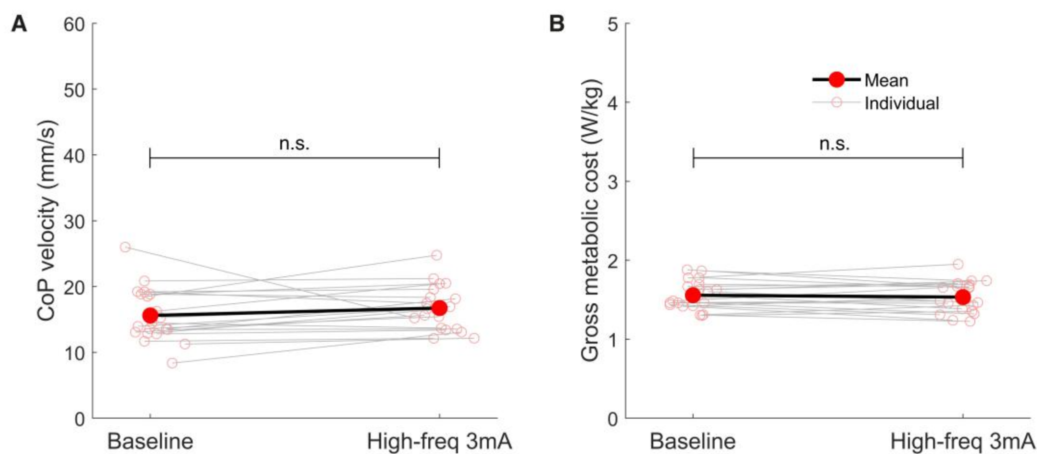


Figure 7: Paired data plots showing mean and individual data of (A) mean CoP velocity and (B) gross metabolic cost for Baseline and High-freq 3mA. Labels: n.s.: not significant.

3.4 Metabolic cost of reflexive vs self-initiated movement

Afterwards, Low-freq 3mA was compared to Self Movement (Figure 8), in which participants had to mimic the movements of Low-freq 3mA. No significant difference in variability between both conditions was found *($t(19) = -1.558$, $p = 0.136$)* (Figure 8A), which indicated that the goal of mimicking the movements was achieved. After this confirmation, a comparison between the same conditions for metabolic cost (gross and net) showed no significant difference as well *($t(19) = 0.316$, $p = 0.755$)* (Figure 8B), suggesting that the metabolic cost of postural sway primarily depends on the magnitude of variability instead of the origin of movement.

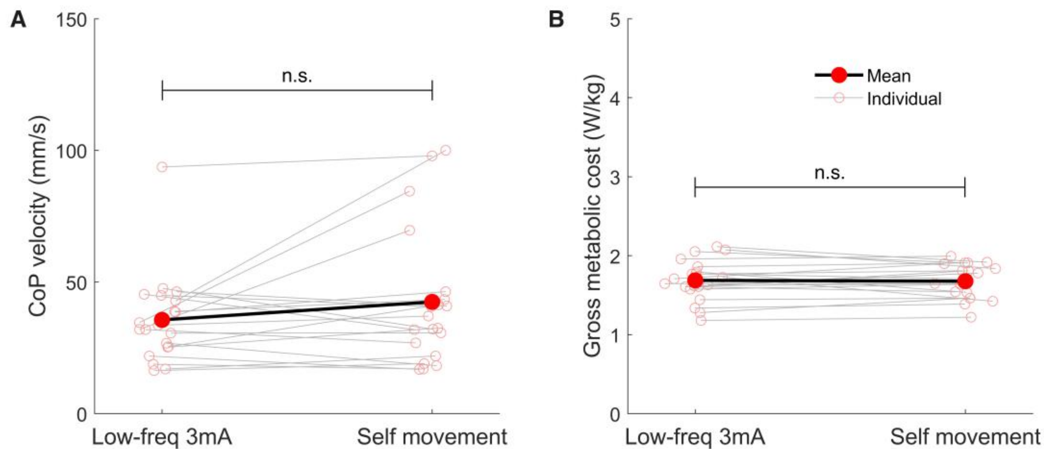


Figure 8: Paired data plots showing mean and individual data of (A) mean CoP velocity and (B) gross metabolic cost for Low-freq 3mA and Self movement. Labels: n.s.: not significant.

3.5 Houdijk comparison

Finally, for the comparison of this study with the study of Houdijk et al. [10], linear regression plots (postural variability vs metabolic cost) of both studies were made in which the mean values per condition of identical metrics were used (i.e. CoP velocity (mm/s) and gross metabolic cost (W/kg). The regression plot with the four conditions from Houdijk et al. [10] showed a slope coefficient of 0.051, meaning that for every mm/s the mean CoP increased, the gross metabolic cost increased with 0.051 W/kg (Figure 9). When doing the same for our study with Baseline and both Low-freq EVS conditions, the slope coefficient was 0.007 (Figure 9), which suggested that the study of Houdijk et al. [10] overestimated the metabolic cost of increasing postural variability. It is plausible that the increasing metabolic cost in the study was not only originating from an increasing postural variability, but also from the induced biomechanical changes.

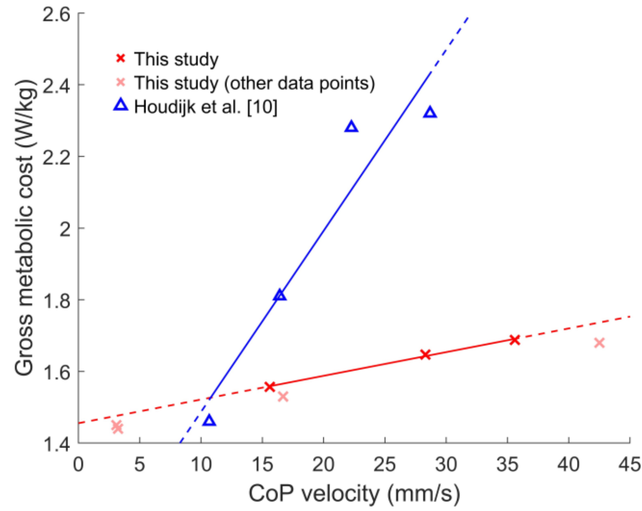


Figure 9: Linear regression plots showing correlation between mean CoP velocity and gross metabolic cost for the study of Houdijk et al. [10] and this study. The study of Houdijk et al. included a baseline condition, a tandem stance condition, a blindfolded tandem stance condition and a tandem stance on balance board condition. For the regression line of our study, Baseline and both Low-freq EVS conditions were included. The other conditions are shown in the figure as well. There is a substantial difference in slope coefficients (0.007 vs 0.051); an increasing postural variability resulted in a more increasing metabolic cost in the study of Houdijk et al. compared to this study.

4. Discussion

The aim of this study was to investigate the metabolic cost of standing balance and to determine the relationship between standing variability and metabolic cost. By using EVS postural sway was manipulated while maintaining a constant state of the biomechanics of standing. To provide insights into the metabolic cost, this study sought to distinguish the energetic contributions of balance control, vestibular stimulation itself, reflexive muscle activity and voluntary movement. The results demonstrate a positive relationship between postural variability and net metabolic cost. Neither vestibular stimulation nor reflexive muscle activity increased metabolic cost in the absence of increased sway. Furthermore, movements of comparable variability induced similar energetic costs, irrespective of whether they were reflexive or self-initiated. Overall, this study demonstrates that the metabolic rate of humans during standing is dependent on postural variability. An increase in postural variability compared to quiet standing leads to an increase in metabolic rate, which indicates that humans tend to choose a preferred postural variability that is associated with a minimization of cost.

A significant increase of ~8% in gross metabolic cost was found during free-standing compared to immobile standing (Figure 2), which can be interpreted as the energetic cost associated with actively maintaining balance during standing. Two studies, which measured the difference in metabolic cost between free-standing and sitting, suggested an increase of ~7% and ~13% from sitting to standing [28], [29], which implies externally supported standing is in the same range as sitting in terms of metabolic cost. However, in our study, a smaller than natural stance width (feet 2 centimetres apart) was chosen, which might have increased the difference in metabolic cost between free-standing and externally supported standing. The other studies did not mention stance width.

A central finding in this study is the positive relationship between postural variability and net metabolic cost. Low-freq 1.5mA and 3mA induced significant increases in both postural variability (CoP velocity and ACR velocity) and net metabolic cost compared to Baseline (Figure 3). Moreover, linear mixed-effects regression showed a positive correlation between variability and net metabolic cost across Baseline and both Low-freq EVS conditions (Figure 4). Two different metrics of variability, namely mean CoP velocity and mean ACR velocity, were considered to investigate whether two different variables would lead to similar outcomes. Both showed consistent patterns across conditions and yielded similar relationships with metabolic cost. After a correlation between postural variability and net metabolic cost was established, and since the correlation was not confounded by vestibular stimulation, reflexive muscle activity or the origin of variability, a comparison with the study of Houdijk et al. [10] was made (Figure 9). The comparison showed that the slope coefficient of CoP velocity vs gross metabolic cost was larger *(0.051 vs 0.007)* in the study of Houdijk et al. [10], suggesting that their correlation was confounded by the changes in biomechanics.

To determine whether the increased metabolic cost was driven by vestibular stimulation or reflexive muscle activity, several comparisons were performed. First, coherence analysis confirmed the absence of mechanical or muscular responses evoked by EVS during supported standing (Figure 5), which was in line with the study of Fitzpatrick et al. [14]. Additionally, no significant difference in metabolic cost was found between externally supported standing with and without EVS (Figure 6). This indicates that vestibular stimulation itself does not carry a measurable metabolic cost, as the stimulation was the only difference between both conditions that could explain a difference. The outcome was in line with studies on vestibular neurons, which showed that the average activity (and thus metabolic cost) of both vestibular organs remain similar due to its binaural mechanism [30]; an increase in activity on one side is compensated with a decrease in activity on the other side, causing vestibular stimulation to have no increased metabolic cost [31]. Second, as expected through previous studies [12], [19], coherence was found between high-frequency EVS and (i) muscular and (ii) mechanical responses (Figure 5), without increasing variability (Figure 7). However, these EVS-evoked reflexes did not increase metabolic cost (Figure 7), suggesting that reflexive muscle activity alone, when not accompanied by increased sway, does not elevate metabolic cost during standing.

To further investigate whether the origin of variability influences metabolic cost, the condition with the most reflexively induced sway (Low-freq 3mA) was compared to Self movement, in which participants aimed to mimic the previous magnitude and frequency of postural sway. No significant differences were found in either variability or metabolic cost between these conditions (Figure 8), suggesting that the metabolic cost of movement during standing primarily depends on the magnitude of variability. However, it should be noted that mimicking the sway, induced by EVS, proved to be difficult for several participants, leading to a large spread in variability during this condition. This spread caused the variability of the self-initiated movement condition to be not normally distributed, as tested with the Kolmogorov-Smirnov test. As a consequence, statistical power is reduced and the strength of conclusions that can be drawn from this comparison is limited. Future studies could improve this by providing live feedback of variability to match other conditions.

Several other methodological choices should be considered while interpreting the results of this study. First, it was checked whether trial order during the experiment played a role in the metabolic measurements. Factors like fatigue during prolonged standing or excitement in the beginning of the experiment could lead to an increase or decrease in metabolic cost over time. A potential trial order effect on the data was suspected, since 7 out of 9 trials with a negative net metabolic cost, as seen in Figure 3 and 4, took place before the immobile condition (on which net metabolic cost was based). Although these observations suggest an increasing metabolic cost over time, linear regression showed no time dependency over all trials as shown in Appendix C.

Since trial order did not explain unexpected metabolic outcomes, it was investigated whether variability in metabolic data was caused by not reaching steady-state after the first 60 seconds. For every condition separately, a paired samples t-test was executed between the second and third 60 seconds. Since no difference was found for all conditions $(p > 0.05)$, it was assumed that steady-state was reached within 60 seconds. Therefore, the observed variability in metabolic cost likely originated from fluctuations inherent to breath-to-breath data and was minimized by averaging the data over a 2-minute time window.

An outcome not in line with the hypothesis and pilot studies was that no significant difference in variability was observed between Low-freq 1.5mA and 3mA, while a difference was observed between Baseline and Low-freq 1.5mA. This suggests that the relationship between EVS peak amplitude and postural variability is not linear within the tested range. In two studies of Day et al. [32], [33] the relationship between current strength and variability was nonlinear: the largest increase in variability was observed at low current strengths, while the increase in variability flattened at higher current strengths. This might explain why a difference in variability was not found between the 1.5 mA and 3 mA conditions.

While the current study has demonstrated that metabolic costs increase with postural variability during free-standing, there are task-dependent scenarios where an opposing relationship can be observed. In an additional study of Houdijk et al. [34], it was shown that the metabolic cost can in fact increase when participants are instructed to minimize their postural variability within a certain boundary of CoP motion. These counterintuitive results suggest that more effort is needed to reduce the CoP variations to a level that is below the variability of quiet standing. Together with the findings in our study, this indicates that humans select a preferred postural variability during quiet standing that is close to the metabolic minimum.

5. Conclusion

This study showed that maintaining standing balance required a measurable amount of metabolic cost and that this cost increased with postural variability. By manipulating postural sway using EVS without altering the biomechanics of standing, the metabolic cost of variability itself could be isolated. Free-standing resulted in an approximately 8% higher metabolic cost compared to externally supported standing, reflecting the energetic cost of active balance control. Increased metabolic cost was only observed when postural variability increased, as vestibular stimulation and reflexive muscle activity alone did not elevate metabolic cost in the absence of increased sway. Furthermore, movements with comparable variability showed similar metabolic costs, independent of whether the movements were reflexively induced or self-initiated. Together, a positive relationship between postural variability and metabolic cost was established that was not confounded by the effects of EVS or biomechanics, indicating that humans adopt a preferred level of postural sway that is close to the metabolic minimum.

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Appendix A – material

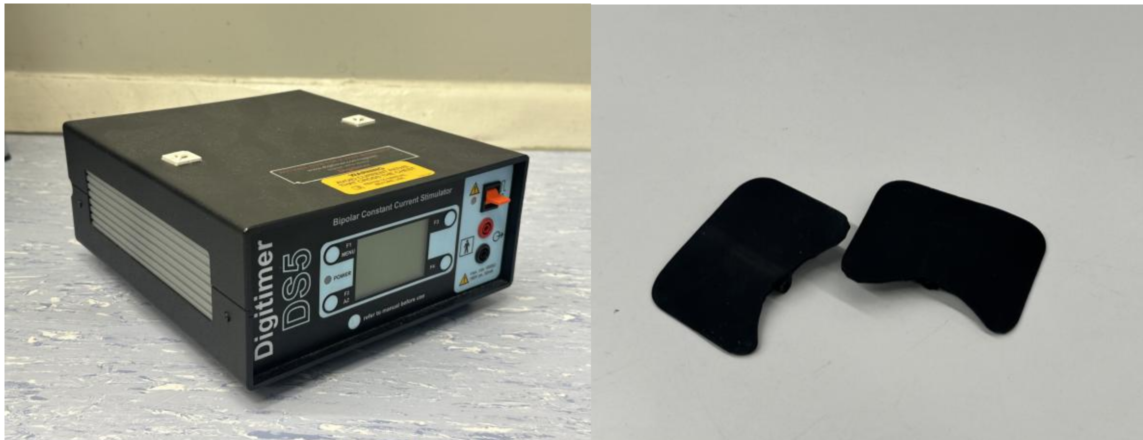

Figure 10: constant current stimulator (DS5, Digitimer, UK) from which the vestibular stimulation was delivered (left). The electrodes which were placed on the mastoid bones behind the ears (right).

Figure 11: metabolic mask (V-982185, Vyaire Medical, USA) where flow and gas measurements are captured and which is connected to the wearable metabolic device (SBx/CPX, CareFusion, USA) (left). The wearable metabolic device was connected to the indirect calorimetry system (MS-CPX, VIASYS, Germany) (right).

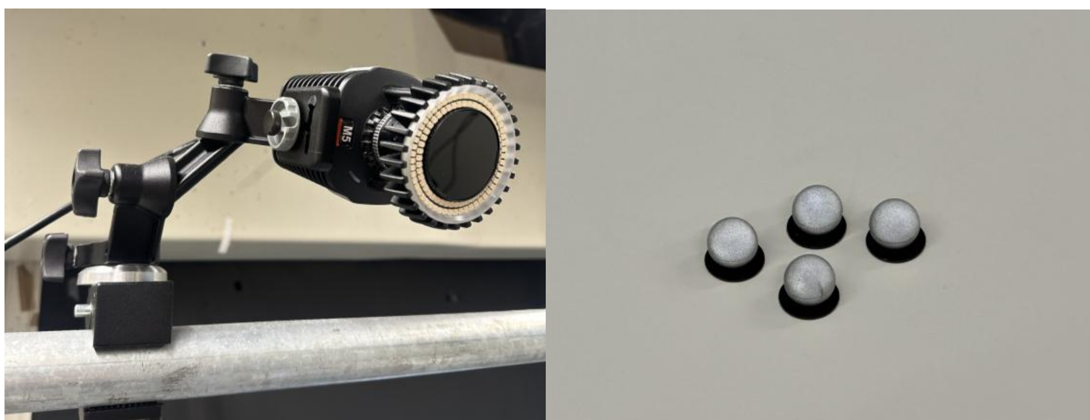

Figure 12: one of the eight motion capture cameras (4x Miqus M3 and 4x Miqus M5, Qualisys AB, Göteborg, Sweden) which surrounded the participants (left). Four of the sixteen motion capture markers that were placed on the anatomical landmarks (right).

Figure 13: Force plate amplifier (Optima OPT-SC, AMTI, MA, US) (left) and force plate (BMS 400600HF-1K, AMTI, MA, US) (right). The horizontal tape was used to outline the participants in anteroposterior direction.

Figure 14: pre-amplifier (NL844, Digitimer) (left) and filter and isolator (NL135 filter, NL820 isolator, Digitimer) (right) with which EMG measurements were taken.

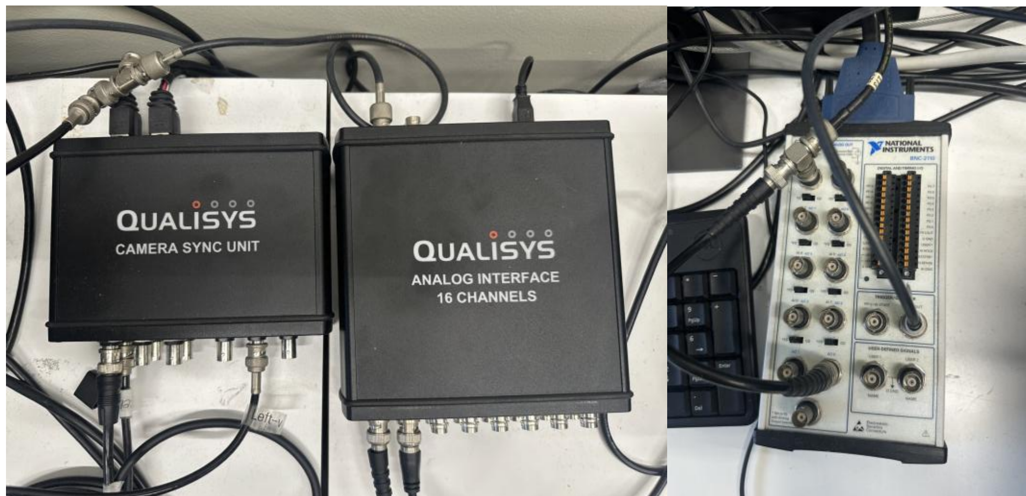

Figure 15: Qualisys sync box (410850, Qualisys AB, Göteborg, Sweden) and Qualisys analog measurement board (230597, Qualisys AB, Göteborg, Sweden) which were used to sync cameras and other measurement equipment (left). D/A board (National Instruments, BCN-2110, TX, US) which was used to send the EVS signal and trigger the measurements of the other equipment (right).

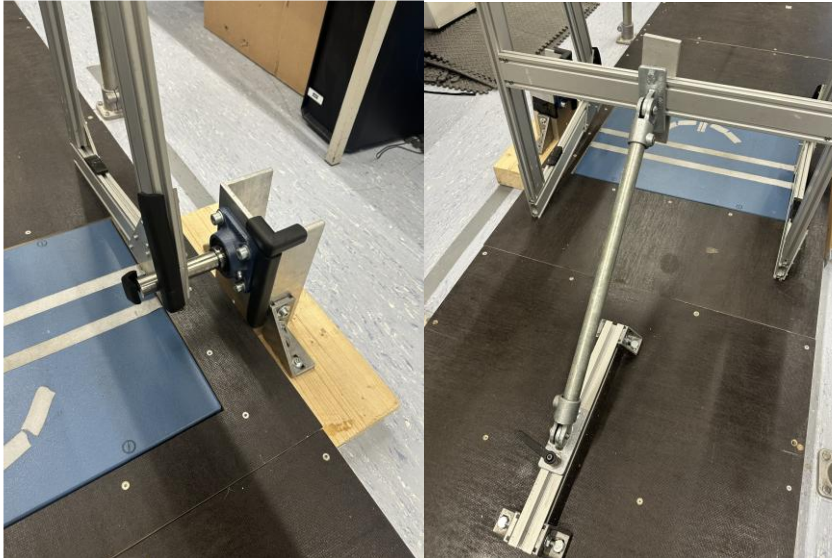

Figure 16: custom-made backboard which was adjustable with the help of a hinge (left). The backboard could be fixed at any given angle by tightening the screw on the glider (right).

Appendix B – participant information

Table 3: Age and anthropometric data per participant and the overall means

Participant	Age	Sex	Length	Weight	Ankle height
1	24	M	1.92	70.9	8.8
2	22	M	1.87	89.2	9.4
3	23	F	1.70	74.6	7.6
4	30	M	1.93	89.3	8.9
5	29	M	1.87	76.9	8.7
6	33	M	1.86	98.5	9.2
7	19	F	1.76	60.7	8.5
8	24	M	1.85	83.8	9.2
9	24	F	1.80	69.8	7.5
10	23	F	1.56	53.0	6.7
11	23	M	1.87	88.4	7.5
12	25	M	1.84	78.8	8.6
13	32	F	1.76	67.6	8.7
14	29	M	1.92	82.1	9.0
15	28	F	1.68	66.1	8.1
16	25	F	1.75	59.6	8.1
17	48	M	1.96	105.5	8.9
18	28	F	1.78	78.0	7.8
19	29	F	1.76	73.7	7.9
20	26	F	1.84	73.0	8.8
mean	27.2	10 F, 10 M	1.81	77.0	8.4

Appendix C – figures

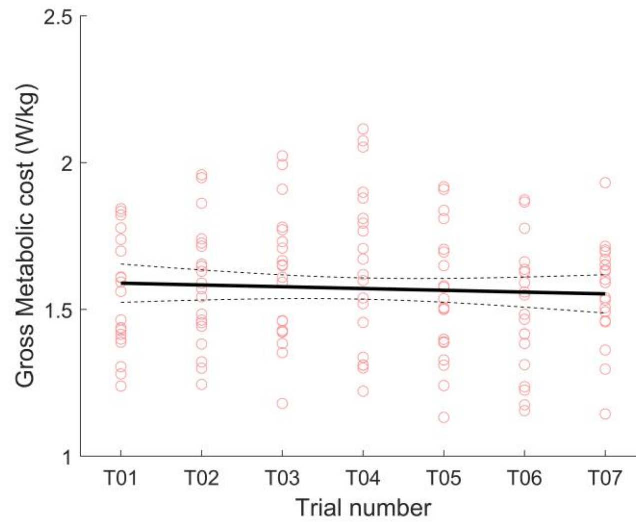

Figure 17: Regression line plot of trial number vs. gross metabolic cost. No trial order effect was found (p = 0.509)

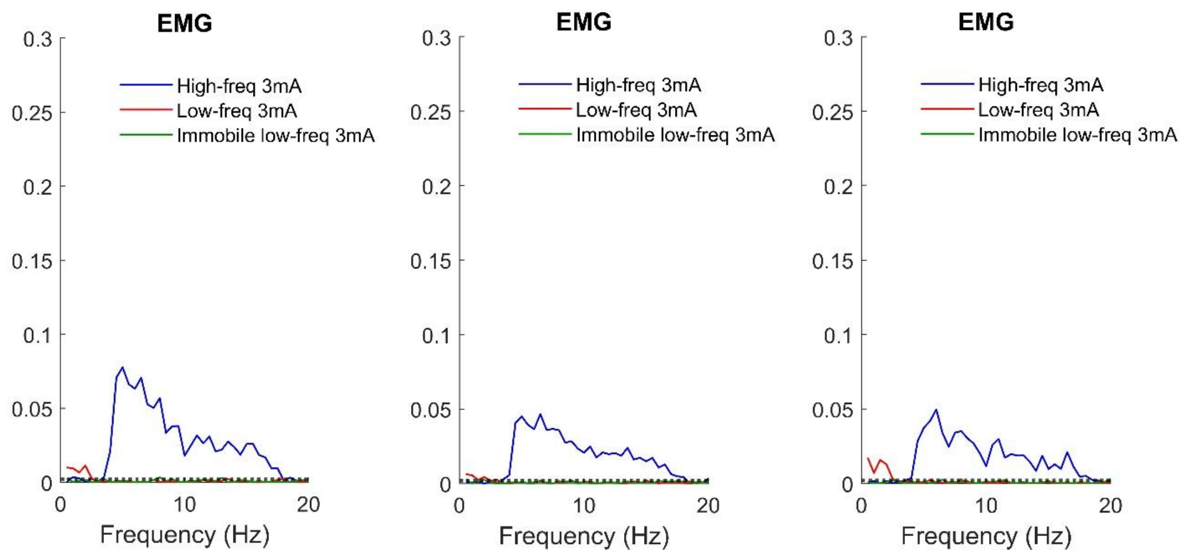

Figure 18: Coherence between EVS signal and EMG for the left gastrocnemius medialis (left), left soleus (middle) and right soleus (right).

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