Optimization Objective of Old and Young Adults during Overground Walking

by

Monica Orlandi

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Department:	Biomechanical Engineering
Supervisor:	Dr. ir. E. van der Kruk

TU Delft

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M. Orlandi (5706386)
Supervisor:
Dr. Ir. E. van der Kruk

Abstract

This study investigates different strategies that the central nervous system might adopt to solve the motor redundancy problem in young and older adults, focusing on metabolic costs, head accelerations, and gait adjustments during overground walking at various speeds. The study addresses gaps in previous research, which primarily focused on younger participants and treadmill-based trials, potentially overlooking natural gait patterns. Ten younger adults (aged 23–28) and five older adults (aged 69–77) completed eight overground walking trials at different speeds, including their preferred walking speed (PWS), predetermined speeds constant for each subject, and variations of their preferred walking speed.

Results showed that younger and older adults had similar preferred walking speeds and comparable metabolic costs when walking at their chosen pace, while younger adults exhibited higher metabolic costs at higher speeds. The PWS did not minimize the metabolic cost for either age group. At their PWS, younger adults both reduced head accelerations and maximized stability, whereas older adults prioritized stability over movement smoothness. Both groups primarily adjusted step frequency rather than step length to accommodate changes in walking speed. Additionally, no significant differences were found in maximum arm swing velocity between the two groups. These findings challenge previous assumptions about age-related differences in walking efficiency and suggest that stability may play a more critical role in gait optimization for older adults. Further research is necessary to uncover the mechanisms driving these adaptations and their impact on gait across the lifespan.

Keywords: ***Optimization, Elderly, Gait***

1 INTRODUCTION

The motor redundancy problem, introduced by Bernstein in 1967, describes the challenge the central nervous system (CNS) faces in selecting the optimal movement pattern for a task [1]. With more actuators than degrees of freedom at each joint, multiple muscles can produce the same movement, and different paths can be taken to achieve the same final position. To resolve this, the CNS likely selects a motor pattern that optimizes a criterion, often by minimizing or maximizing a cost function, though the exact nature of this function is unclear. The interest in this issue has grown significantly with the advent of musculoskeletal simulations of gait. Accurately solving the motor redundancy problem is essential for replicating real-life human movements [2]. Several theories have been proposed regarding the optimization criterion. The most widely supported hypothesis suggests that the CNS aims to minimize the metabolic cost of transport, thereby increasing movement efficiency. Numerous studies back this theory, emphasizing the importance of reducing energy demands for optimized movements [3–16]. Stability is another potential optimization criterion. Since the body's balance sensors are located in the ears, minimizing head accelerations during gait might be a strategy to enhance stability [17, 18], as consistent head accelerations have been observed under various walking conditions [19].

Another critical aspect of gait optimization is how the CNS adjusts the locomotion pattern to satisfy the optimization criterion, in particular the preferred walking speed (PWS) and step frequency [8–12, 14]. Recent studies emphasize how the preferred walking speeds or step frequencies optimize energy efficiency, while deviations from these self-selected values, whether increasing or decreasing, result in a higher metabolic cost [1, 10, 20]. Given the close relationship between step frequency, step length, and walking velocity, it is plausible that the CNS adjusts a combination of these factors to optimize walking patterns [3, 4, 7, 15, 17].

Determining the optimization criterion and method used by the CNS is particularly challenging because they may vary in response to external factors. For example, older individuals exhibit different locomotion patterns compared to younger adults, often walking at slower speeds with shorter steps

[21, 22]. Additionally, older adults typically incur higher metabolic costs for the same task, indicating changes in CNS and musculoskeletal system mechanisms [23, 24].
Several explanations have been explored for this phenomenon, including an increased focus on stability, increased muscle coactivation, and age-related changes in muscle mass and strength. It has been hypothesized that the elderly might prioritize stability due to the heightened risk of falls [5, 25–28]. However, they employ less efficient strategies to attenuate head accelerations during locomotion [18], such as muscle coactivation to increase limb stiffness. Leg muscle coactivation has largely been observed to increase with age during locomotion [24, 29–33]. Furthermore, the loss of muscle mass and strength necessitates recruiting more muscle fibers for longer durations to achieve the same strength [7, 23, 25, 30, 34, 35], potentially increasing fatigue [26]. Age-related brain functionality changes might also contribute to gait modifications, as studies have shown more intense brain activity in older individuals performing the same tasks as younger ones [36, 37].
As the global population ages [38], it is crucial to incorporate age-related adaptations into biomechanical studies. Understanding these adaptations will enhance our knowledge of human movement and provide strategies to improve mobility and stability in older adults.

The optimization criteria for walking have been examined in correlation with multiple factors, including walking speed, step frequency, head accelerations, and the influence of aging. However, studies on the metabolic cost of walking frequently focus solely on younger participants, typically 40 years old or younger [3, 10, 20], or fail to specify the ages of the subjects [12]. Both younger and older individuals should be included to investigate potential differences in gait and how the optimization criterion changes with age. Another common issue is the choice of walking speeds; researchers often concentrate only on the PWS [21, 28], or use subjective speeds, instructing participants to walk "as fast as they can" or to adopt a "fast" or "slow" pace [18, 28]. A more rigorous way of defining the walking speeds is necessary to have a better comparison between the age groups. To simplify experimental protocols or address logistical challenges, treadmills are frequently used in these studies [5, 7, 23, 24, 26, 29–31, 33, 34, 39–42]. However, the locomotion pattern on a treadmill differs significantly from the natural overground pattern. For example, a longer step length and a higher PWS are typically observed in overground walking [22, 43–51]. These changes in the gait might be adaptations of the CNS to the more unstable treadmill environment, given the lack of visual feedback and the forced constant walking velocity [16, 52]. It is therefore crucial that all participants walk overground to accurately examine natural locomotion patterns.
Finally, the literature suggests that the maximum velocity of the arm swing decreases for the elderly [53], and this phenomenon is linked to the lower gait velocity preferred by older adults [53, 54]. However, this hypothesis has not been thoroughly tested yet, as only two speeds were analyzed.

The purpose of this study was to investigate the following hypotheses:

a) At their preferred walking speed, the metabolic cost of the older adults is higher than that of the younger adults.

b) At the same fixed speeds, the metabolic cost of the older adults is higher than that of the younger adults.

c) The primary objective of locomotion is minimizing the cost of effort, hence the minimum metabolic cost aligns with the preferred walking speed.

d) Head accelerations are minimized at the preferred walking speed in both young and older adults.

e) When different velocities are imposed, younger adults maintain their preferred step frequency and adjust step length to keep pace.

f) When different velocities are imposed, older adults will increase step frequency and maintain the same step length to keep pace.

g) When the same velocity is imposed, older adults have the same maximum arm swing velocity as younger adults.

2 METHODS

All data analyses were performed using MATLAB R2022b (The MathWorks) [55].

2.1 Subjects

A total of 15 participants were enrolled in the study, including 5 older adults (OA) aged 69 to 77 years (3F, 2M) and 10 younger adults (YA) aged 23 to 28 years (5F, 5M). Exclusion criteria were chronic heart disease, diabetes, prior lower limb surgeries or prostheses, neuromuscular injuries, recent falls within the past 6 months, or participation in specialized strength or endurance training. The inclusion criteria required participants to be capable of performing their daily activities without assistance and walking independently. Written consent was obtained from all participants [56]. A summary of the demographic and anthropometric data of the subjects is presented in Table 4 in Appendix B.

2.2 Protocol

Preparation
All participants were advised to observe a fasting period of 3 hours, refrain from alcohol and nicotine consumption for 2 hours, and coffee for 4 hours before the experiment, in line with the protocol formulated by Compher et al. [57].

Beginning of the session
At the beginning of the session, participants' height and weight were measured. Their resting metabolic rate was then measured while they stood unsupported and silently for 7 minutes with the COSMED K5 metabolic mask [58]. The first 4 minutes were discarded as an adjustment period, and the oxygen consumption rate from the remaining 3 minutes was used to calculate the Net Cost of Walking (NCoW) for the subsequent trials, as indicated in the protocol followed by [6, 59].

Trials
Following the resting metabolic rate assessment, participants completed eight 6-minute overground walking trials on an outdoor flat and straight pavement. The first trial was conducted at each participant's preferred walking speed, followed by trials at predetermined speeds of 0.8 m/s, 1.2 m/s, and 1.6 m/s, as well as personalized speeds of PWS $\pm$ 5% and PWS $\pm$ 10%. The trial order was semi-randomized to minimize fatigue, with rest periods of 5 minutes provided between trials for hydration and rest. During the breaks, participants were invited to sit down on a chair. To regulate the walking speed during the trials, participants were instructed to match the pace set by a pacing cart, carried by a researcher, described in section 2.4. To avoid influencing the step frequency and length, the researcher consistently walked behind the participants. The participants kept a fixed distance from a visual reference point on the pacing cart (see Figure 1).

2.3 Sensors

Before the start of the trials, each participant was equipped with a COSMED K5 mask covering their mouths and noses to measure oxygen (O_2) consumption and carbon dioxide (CO_2) production rates. Additionally, three IMU sensors were placed on the forehead, right tibia, and right forearm to respectively assess head accelerations, step frequency, and arm swing velocity.

2.3.1 COSMED K5

The COSMED K5 is a wearable metabolic system designed to analyze gas exchange rates outside of the laboratory with a breath-by-breath analysis [58]. It consists of a lightweight device that can be worn as a backpack and carried around, along with a silicone mask that covers the mouth and nose (Figure 9 in Appendix A). The mask was checked to ensure a proper fit for each participant. The COSMED K5 was calibrated before each experimental session following the manufacturer's guidelines.

2.3.2 Cometa MiniX

The IMUs used in this study were the Cometa MiniX sensors (Figure 10 in Appendix A) [60]. The sensors were attached with double sided tape to the center of the forehead, the middle of the right dorsal forearm, and the right tibia (Figure 11 in Appendix A). For subject 3 in the OAs group, the leg IMU was placed on the left side due to a skin condition. The sampling rate was 500 Hz.

2.4 Pacing Cart

The pacing cart is shown in Figure 1. It was constructed using the Zozen collapsible measuring wheel (Figure 12 in Appendix A). To measure the instantaneous speed, the N317 Retoo Bike Computer (Figure 13 in Appendix A) was mounted on the cart. This model was selected for its significant digits, which provide higher speed precision. During the self-selected speed trial, the measuring wheel measured the distance walked. Since the duration of the trial is fixed, it was possible to calculate the average walking speed. During the other trials, the bike computer measured the instantaneous speed by counting the rotation per minute of the wheel, allowing the researcher to check that the correct velocity was kept continuously. A PVC pipe was mounted on the measuring wheel to provide a reference point the subjects could look at to check their speed without looking at the researcher.

2.5 Data Pre-Processing

The COSMED system recorded the trials separately, whereas the MiniX sensors recorded uninterrupted for the whole duration of the session. Moreover, the IMU data quality was greatly affected by high-frequency noise, whereas the metabolic system provided breath-by-breath measurements, ensuring high precision without noise interference. However, when the subjects spoke during a trial, the measurements displayed a sudden, pronounced peak.

To address these issues, the IMU data of the lower leg were filtered using a low-pass, zero-lag, second-order Butterworth filter with a 3 Hz cutoff frequency [54]. Next, the continuous IMU recordings were segmented into individual trials. The lower leg sensor data were used for the segmentation given that the participants sat down between trials, which provided clear markers for identifying the trial start and end points. Using the known trial sequence, each burst of lower leg acceleration was matched to the corresponding walking speed. The timestamps identified for each trial were then synchronized across all IMU recordings.

The COSMED data were filtered to remove the spikes associated with speech. These peaks were identified as outliers and removed from the signal.

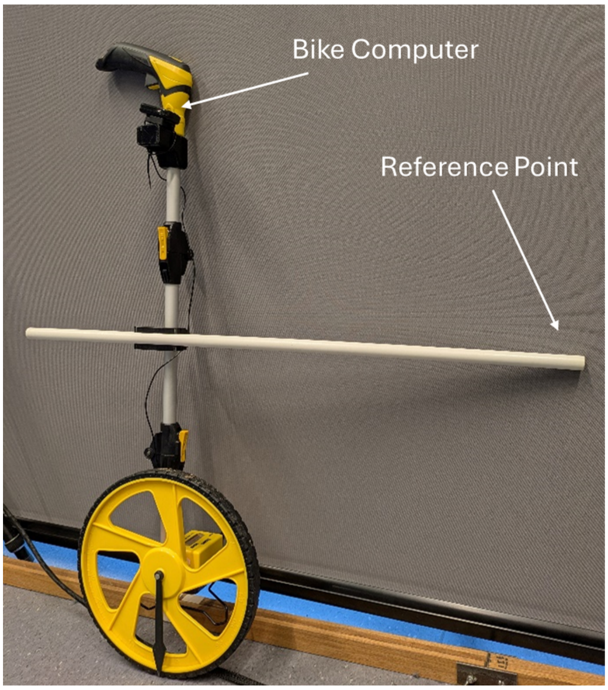

Figure 1: Pacing Cart used to measure and impose the walking speeds. It comprises a measuring wheel, a bike computer, and a horizontal plastic pipe as a reference point.

2.6 Description of Variables

2.6.1 Metabolic Cost

Oxygen Consumption and Carbon Dioxide Production Rates $[\frac{L}{kg \cdot s}]$: these measures were recorded by the COSMED system with each breath. They indicate the amount of gas, either O_2 or CO_2, present in the breath in a period of time [61]. For the analysis, only the data from the last three minutes of each trial were used, as the initial portion was excluded as a transitional period. All values were normalized to body weight. Additionally, the respiratory exchange ratio was calculated for each trial as the ratio of CO_2 production rate to O_2 consumption rate.

Gross and Net Cost of Walking $[\frac{J}{kg \cdot m}]$: the Cost of Walking (CoW) is defined as the metabolic energy expended per kilogram of body mass per meter traveled [26]. The net value is determined by subtracting the energy expenditure during quiet standing from the total energy used. These variables were calculated following the methodology outlined by Das Gupta et al. [26]:

$$GCoW = \frac{(15962 + 5155 \cdot RER) \cdot \dot{V}O_2}{walking\ speed} \quad (1)$$

$$NCoW = \frac{(15962 + 5155 \cdot RER) \cdot \Delta\dot{V}O_2}{walking\ speed} \quad (2)$$

where RER stands for Respiratory Exchange Ratio $[-]$, $\dot{V}O_2$ denotes the oxygen consumption rate, and walking speed is defined as the average speed over the entire walking trial, calculated as the ratio of the distance covered to the trial duration $[m/s]$. The mean values of the GCoW and of the NCoW were computed for each trial.

2.6.2 Head Accelerations

Root Mean Square: after removing the outliers in the vertical (V), anterior-posterior (A-P), and medio-lateral (M-L) directions, the Root Mean Square (RMS) of the head accelerations was calculated over the whole trial using the formula:

$$a_{RMS,i} = \sqrt{\frac{1}{N} \sum_{n=1}^{N} |a_{n,i}|^2} \quad (3)$$

where $a_{n,i}$ is the head acceleration's vector in the i-direction at the time n, and N is the total length of the vector. The RMS measures the overall magnitude of acceleration fluctuations in each direction, so this measure is used to quantify the smoothness of the gait.

Harmonic Ratio: The Harmonic Ratio (HR) quantifies gait stability by measuring the symmetry and rhythmicity of walking, assessing the regularity of oscillations in acceleration signals [17]. It is calculated by segmenting acceleration signals into individual strides, identified by peaks in vertical acceleration. For each segment, the first 10 harmonics were derived from the discrete Fourier transform of the acceleration signals. These harmonics, corresponding to multiples of the fundamental frequency, were categorized as odd (alternating patterns) and even (repeating patterns). Even harmonics complete an even number of cycles per stride, thus reflecting stable, symmetrical gait patterns, while odd harmonics indicate asymmetry and potential instability.

The stride-based HRs for the vertical and anterior-posterior directions were computed as the ratio of the sum of even harmonic magnitudes to the sum of odd harmonic magnitudes [17, 19]:

$$HR_{stride} = \frac{\Sigma\ even\ harmonics}{\Sigma\ odd\ harmonics} \quad (4)$$

For the medio-lateral accelerations, the ratio was reversed as they are limb-dependent and monophasic in each stride, resulting in higher amplitudes for odd harmonics compared to even harmonics [19]. The HR for each stride was averaged within each trial to produce a single value per walking speed.

$$HR_{trial} = \frac{1}{N} * \sum_{i=1}^{N} HR_{stride_i} \quad (5)$$

This method emphasizes stride-to-stride variability in harmonic content, providing insight into gait consistency across the trial. Averaging stride-based HRs captures minor fluctuations, offering a more representative measure of overall gait regularity and symmetry. This is particularly useful for populations with higher gait variability, such as older adults or individuals with neurological impairments, as it reflects subtle irregularities that may impact stability.

2.6.3 Step Frequency

Mean Step Frequency [Hz]: the mean step frequency for each trial was determined by analyzing the frequency spectrum of the lower leg anterior-posterior acceleration. Since the IMU was attached to a single leg, the identified peak corresponded to the stride frequency, which was then doubled to obtain the step frequency. Limb symmetry was assumed.

2.6.4 Arm Swing Velocity

Maximum Angular Velocity [°/s]: for each trial, the angular velocity of the arm in the shoulder and elbow's flexion-extension direction was divided into individual strides, based on the step frequency analysis. The peak angular velocity was determined for each stride, and the average maximum swing velocity was computed across the entire trial. Limb symmetry was assumed.

2.7 Statistical Analysis

The Shapiro-Wilk test was used to assess the normality of the data. Levene's test was applied to evaluate homoscedasticity. For comparing differences between groups, a Student's t-test was performed if the assumptions were met; otherwise, the Wilcoxon rank-sum test was used. Boxplots are presented to provide a visual representation of the results.

To test hypothesis c, a quadratic curve was fitted to each individual's data to determine whether the preferred walking speed corresponded to the minimum metabolic cost of walking. This approach was based on the established quadratic relationship between walking speed and metabolic cost, as documented in previous studies [10, 23, 24, 34]. The difference between the speed that minimizes the quadratic curve, referred to as the optimal speed, and the PWS was calculated (see Figure 20 in Appendix C) and normalized by the PWS to facilitate inter-subject comparisons. The absolute value of the normalized speed was used, as the focus was on identifying any difference between the optimal speed and the PWS, regardless of whether the optimal speed was higher or lower. A one-sample t-test was performed to determine if the mean of the distances across subjects was statistically significantly different from zero. A similar approach was followed for the head accelerations to test hypothesis d. The only difference was that the recorded data were used instead of their quadratic model to calculate the optimal speed, as no assumption was found on the relationship type between head accelerations and speed in the literature. For the RMS of the head accelerations, the optimal speed was defined as the speed corresponding to the minimum RMS, whereas for the HR, it was defined as the speed corresponding to the maximum HR.

To test for statistically significant differences between trials, a repeated measure ANOVA was run. In case the test assumptions were not met, its correspondent non-parametric test was chosen, the Friedman test. For the post-hoc analysis, Tukey's honestry significant difference was calculated as a multiple pairwise comparison.

3 RESULTS

An overview of the available data per subject is reported in Table 6 in Appendix B. Subject 4 from the YA group was excluded from the metabolic cost analysis due to excessive talking, which likely disrupted the mask's seal during the trials, compromising the accuracy of gas exchange measurements. Due to connectivity issues between the IMUs and the receiver, as well as memory limitations of individual units, acceleration data for YA subjects 3, 4, 5, 7, 8, and 9 were lost. Therefore, only the remaining four young subjects were included in the relative analyses.

3.1 Participant Selection

As expected, there was a significant difference in age between the two groups ($p<0.001$), while height ($p = 0.894$), weight ($p = 0.407$), and MBI ($p = 0.390$) did not differ significantly between groups (Figure 14 in Appendix B). Therefore, the two groups were equivalent from an anthropometric point of view.

Table 1: Relative Error of walking speed per trial for each subject. PWS represents the Preferred Walking Speed. The mean was calculated using the absolute values of the relative errors. The errors exceeding 5% are highlighted in orange. Cells with a red background indicate trials that were not conducted.

	Error Table						
Subject	**PWS + 5%**	**PWS - 5%**	**PWS + 10**	**PWS - 10**	**0.8**	**1.2**	**1.6**
1 young	0%	1%	0%	2%	1%	-2%	-3%
2 young	-2%	0%	0%	-1%	-1%	-2%	-3%
3 young	-1%	-1%	1%	-1%	0%	-1%	-1%
4 young	-4%	0%	1%	1%	-1%	3%	-3%
5 young	-3%	2%	2%	1%	18%	0%	-4%
6 young	-2%	0%	4%				1%
7 young	-2%	-1%	-2%	-1%	13%	-3%	-4%
8 young	-1%	-2%	-2%	-4%	5%	0%	3%
9 young	1%	-1%	0%	0%	9%	1%	4%
10 young	0%	-2%	1%	-2%	10%	4%	2%
1 old	-4%	2%	-1%	0%	4%	-1%	-4%
2 old	-3%	-1%	-3%	2%	5%	3%	-1%
3 old	-3%	1%	2%	1%	1%	-2%	
4 old	-2%	-1%	1%	-1%	-1%	5%	-1%
5 old	0%	0%	2%	0%	21%	10%	-2%
Mean	**-2%**	**0%**	**0%**	**0%**	**6%**	**1%**	**-1%**
Mean (abs)	**2%**	**1%**	**1%**	**1%**	**6%**	**3%**	**3%**

3.2 Walking Speed per Trial

During the session with subject 6 of the YAs, it started to rain; therefore, the trials at PWS + 10%, at 0.8 m/s, and at 1.2 m/s were not performed. Subject 3 of the OAs was not physically able to complete the trial at 1.6 m/s.

Keeping a constant speed during the 6 minutes of each trial was challenging for all subjects. Maintaining a speed of 0.8 m/s proved to be the most challenging for participants, as indicated by the high relative errors (Table 1). Seven out of fifteen subjects had a relative error greater or equal to 5%, with an average of 6%. This speed was significantly slower than each participant's PWS ($p < 0.01$), making it difficult for them to adjust to such a reduced pace.

3.3 Preferred Walking Speed

The preferred walking speeds of younger and older adults showed no statistically significant difference in means ($p = 0.598$), as shown in Figure 2.

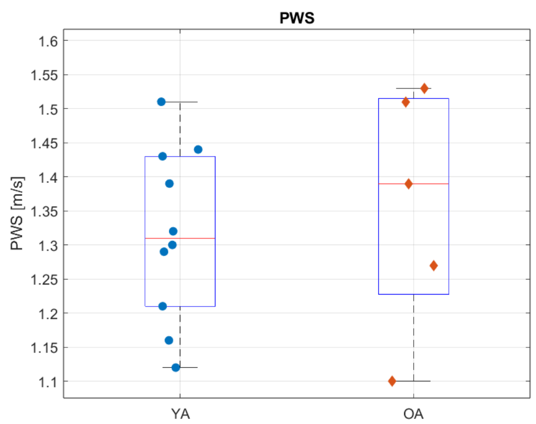

Figure 2: Boxplot of the Preferred Walking Speed (PWS) of the young (blue dots) and old (orange diamonds) subjects.

3.4 Metabolic Cost of Walking

The mean VO_2 consumption during the resting period did not show statistically significant differences between YAs and OAs ($p = 0.2030$) (Figure 15 in Appendix C).

In the analysis of the metabolic cost of walking, considerable variability was observed between subjects (Figures 16 and 17 in Appendix C).

3.4.1 Hypothesis a

At their preferred walking speed, the metabolic cost of the older adults is higher than that of the younger adults.

When comparing the Gross and Net Cost of Walking between young and old adults, no statistically significant differences were observed ($p = 0.118$ and $p = 0.181$, respectively). Therefore, hypothesis a is rejected. As shown in Figure 3, the 95% confidence intervals for the means of the two populations overlap.

A power analysis revealed that the minimum sample size to find meaningful differences between the metabolic cost of the two age groups at the PWS was $n = 10$ for the GCoW and $n = 23$ for the NCoW.

3.4.2 Hypothesis b

At the same fixed speed, the metabolic cost of the older adults is higher than that of the younger adults.

As shown in Figure 4, the older adults exhibited greater variability between subjects, as indicated by the wider confidence intervals for the means.

No statistically significant differences were found in the Gross or Net Cost of Walking at 0.8 m/s between young and older adults ($p = 0.222$ and $p = 0.352$, respectively). At 1.2 m/s, older adults exhibited a significantly lower GCoW compared to younger adults ($p = 0.020$), though no difference was observed in the NCoW ($p = 0.062$). At 1.6 m/s, older adults demonstrated significantly lower GCoW and NCoW than younger adults ($p = 0.001$ and $p = 0.012$, respectively).

This suggests that as walking speed increases, younger adults experience a greater rise in metabolic cost compared to older adults.

A power analysis showed the minimum sample sizes needed to detect differences in metabolic cost between age groups: at 0.8 m/s, $n = 7$ (GCoW) and $n = 47$ (NCoW); at 1.2 m/s, $n = 6$ (GCoW) and $n = 13$ (NCoW); and at 1.6 m/s, $n = 4$ (GCoW) and $n = 7$ (NCoW).

3.4.3 Hypothesis c

The primary objective of locomotion is minimizing the cost of effort, hence the minimum metabolic cost aligns with the preferred walking speed.

For the GCoW, the absolute value of optimal walking speed for young and older adults was 16% more than the PWS ($p = 0.005$ and $p = 0.028$, respectively) (Figure 5).

The NCoW displayed similar results. For the younger adults, the absolute value of optimal walking speed was 31% higher than the PWS ($p = 0.004$). For older adults, the absolute value of optimal walking speed was 35% higher than the PWS ($p = 0.009$).

Regarding differences between groups, no statistical difference was found in the results between YAs and OAs for the GCoW ($p = 0.940$) or the NCoW

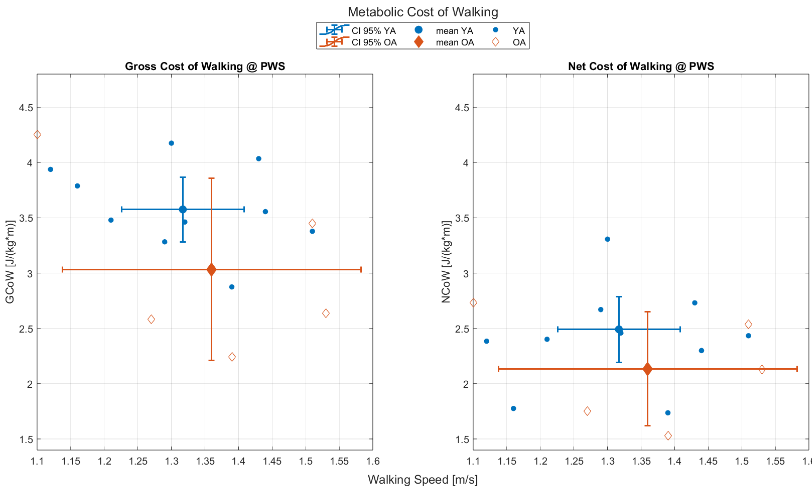

Figure 3: Mean Gross (left) and Net (right) Cost of Walking during the Preferred Walking Speed (PWS) trial. The blue and orange error bars represent the 95% confidence intervals of the young adults (YA) and old adults (OA)' means, respectively.

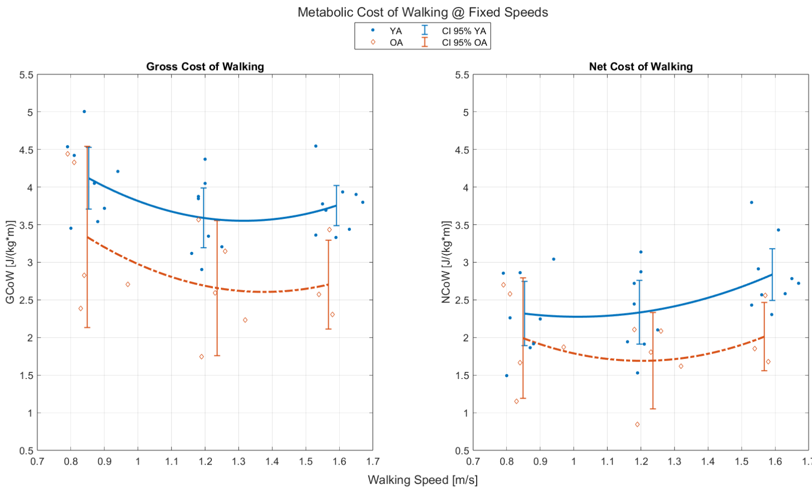

Figure 4: Comparison of the Metabolic Cost of Walking (CoW) between younger (YA) and older adults (OA) across fixed walking speeds, represented with the solid blue line and the dash-dotted orange line, respectively. The plots display the Gross CoW (left) and Net CoW (right) fitted with second-order polynomial curves, based on trials conducted at walking speeds of 0.8 m/s, 1.2 m/s, and 1.6 m/s. Error bars represent the 95% confidence intervals of the mean for each group at these fixed speeds.

($p = 0.606$). Similarly, there were no differences between GCoW and NCoW for the YAs ($p = 0.066$) and for the OAs ($p = 0.057$).

These findings prove that both the GCoW and the NCoW were not minimized at the PWS for young or older adults.

The power analysis indicated a minimum sample size of $n = 7$ (GCoW) and $n = 5$ (NCoW) for YAs, and $n = 6$ (GCoW) and $n = 4$ (NCoW) for OAs.

3.5 Head Accelerations

Figures 18 and 19 in Appendix C display the RMS of the head accelerations and harmonic ratio per subject, respectively.

3.5.1 Hypothesis d

Head accelerations are minimized at the preferred walking speed in both young and older adults.

For the YAs, the absolute value of optimal walking speed for the RMS in the V direction resulted 18% more than the PWS ($p = 0.045$) (Figure 6, Table 2a). For the other directions, there was no statistically significant difference between the optimal speed and the PWS ($p = 1$ for both A-P and M-L).

In contrast, for the OAs, the PWS did not align with the optimal speed of the RMS in any direction ($p < 0.05$), indicating that the PWS did not minimize head accelerations for older adults.

Among the YAs, no significant difference was found between the optimal speed of the HR and the PWS in any direction ($p \geq 0.5$) (Figure 6, Table 2b). In the V direction, the p-value was equal to the significance threshold (α), likely due to half of the subjects having their PWS exactly match the optimal speed, resulting in the statistical test being applied to only two values as the Wilcoxon signed-rank excludes zero from its analysis, thus reducing

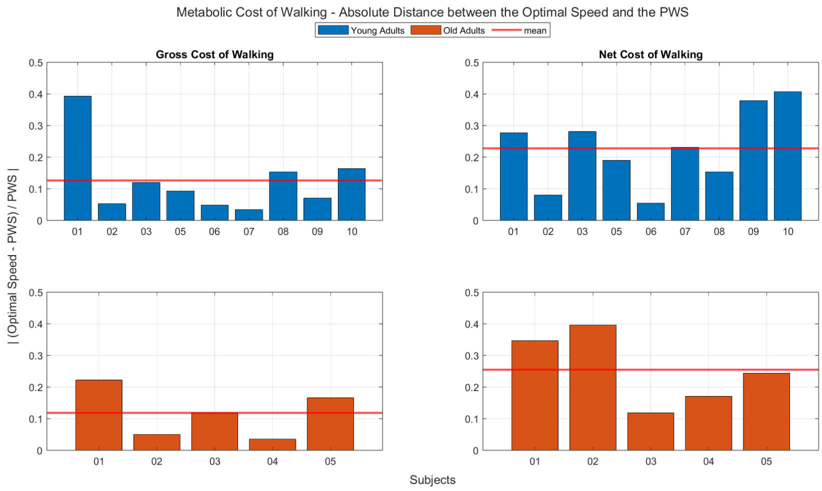

Figure 5: Bar plot of the absolute values of the distance between the optimal walking speed, determined as the speed that minimizes the quadratic fit of the metabolic cost of walking (CoW), and the preferred walking speed (PWS) for each subject, normalized by PWS. Results for Gross CoW are displayed on the left, and those for Net CoW on the right. The top row represents data from young adults, while the bottom row shows data from older adults. The mean distance is marked in red.

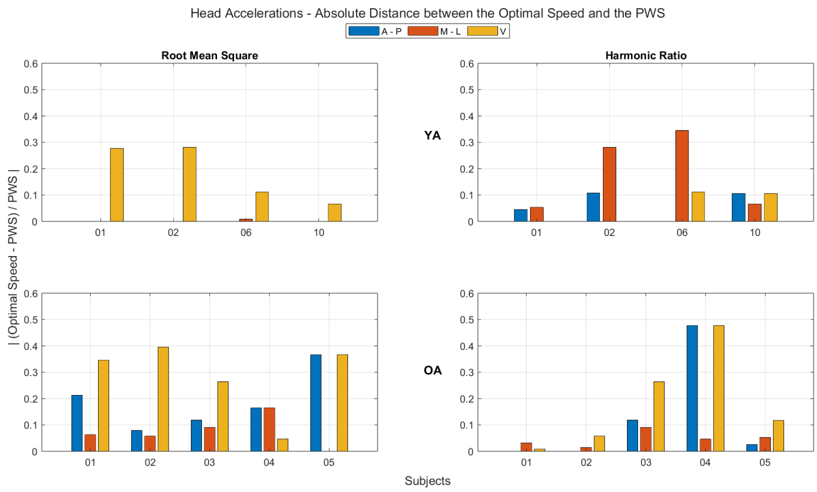

Figure 6: Bar plot with the absolute values of the distance between the optimal walking speed, determined as the speed that minimizes the Root Mean Square (RMS) or the Harmonic Ratio (HR) of the head accelerations, and the preferred walking speed (PWS) for each subject, normalized by PWS. The values relative to the RMS are shown on the left, whereas those relative to the HR are on the right. Results for the young adults (YA) are displayed on the top row and those for the old adults (OA) are on the bottom. The blue bars represent data from the anterior-posterior (A-P) direction, the orange ones are those relative to the medio-lateral (M-L) direction, while the yellow ones show data from the vertical (V) direction.

confidence in the results.

For the OAs, the PWS aligned with the optimal speed in the A-P and V directions. However, in the M-L direction, the absolute value of optimal walking speed resulted almost 5% higher than the PWS ($p = 0.021$).

The power analysis indicated that, for the RMS of head accelerations, the minimum sample size required to detect a significant difference was $n = 34$ (M-L) and $n = 6$ (V) for YAs, and $n = 5$ (A-P), $n = 8$ (M-L), and $n = 5$ (V) for OAs. For the HRs, the required sample sizes were $n = 8$ (A-P and M-L) and $n = 13$ (V) for YAs, and $n = 23$ (A - P), $n = 6$ (M-L), and $n = 12$ (V) for OAs.

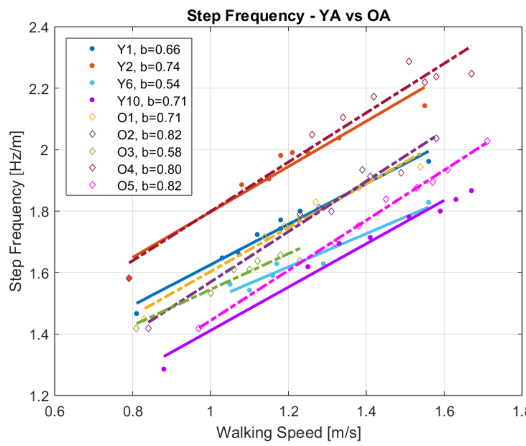

Figure 7: Mean step frequency per walking speed with corresponding linear regression lines. Slope coefficients of the regressions are provided in the legend. Young Adults (YA) are represented by dots and solid lines, while Old Adults (OA) are depicted with open diamonds and dotted lines.

3.6 Step Frequency

There is no statistically significant difference in the slope coefficients of the step frequency's linear regressions between YAs and OAs ($p = 0.238$) (Figure 7).

No statistically significant differences were found between the step frequencies of YAs and OAs ($p = 0.2260$) (Figure 24 in Appendix C), and between their step lengths ($p = 0.7714$).

3.6.1 Hypothesis e

When different velocities are imposed, younger adults maintain their preferred step frequency and adjust step length to keep pace.

The analysis of step frequency showed significant variation across trials for young adults ($p = 0.002$), with notable differences between the slowest (0.8 m/s) and fastest (1.6 m/s) trials (Figure 25a in Appendix C, Table 3a).

Similarly, step length also varied significantly across speeds ($p = 0.002$), aligning with changes in step frequency (Figure 26a in Appendix C, Table 3a).

A comparison of intra-subject variances revealed that step frequency (mean variance: 0.023 Hz) was more adaptable than step length (mean variance: 0.005 m; $p = 0.033$). These findings suggest that young adults primarily adjusted their step frequency, especially at speeds further from their PWS, where changes were more pronounced.

A power analysis of the step frequency revealed that a minimum of 9 subjects were required to detect statistically significant differences.

3.6.2 Hypothesis f

When different velocities are imposed, older adults will increase step frequency and maintain the same step length.

The step frequency analysis for older adults revealed significant differences across trials ($p < 0.001$), indicating variability in frequency across walking speeds (Figure 25b in Appendix C, Table 3b). Initially, a significant change in frequency was only detected at 0.8 m/s. Repeating the analysis without this trial revealed additional differences: step frequency at PWS+10% differed from PWS-10% ($p = 0.035$), and trials at 1.2 m/s and 1.6 m/s also differed ($p = 0.031$).

Step length also varied significantly ($p < 0.001$) (Figure 26b in Appendix C, Table 3b). Intra-subject variance showed a preference for adjusting frequency (mean variance: 0.030 Hz) over length (mean variance: 0.004 m; $p = 0.022$), similar to younger adults.

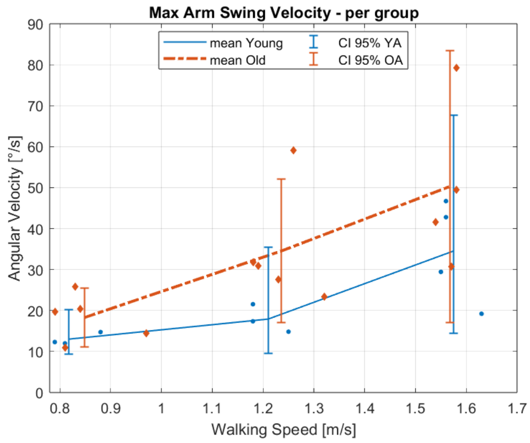

Figure 8: Comparison of the mean maximum arm swing velocity between younger (solid blue line) and older adults (dash-dotted orange line) across fixed walking speeds. The plots display the mean of the maximum arm swing velocity based on trials conducted at walking speeds of 0.8 m/s, 1.2 m/s, and 1.6 m/s. Error bars represent the 95% confidence intervals of the mean for each group at these fixed speeds.

A power analysis of the step frequency revealed that a minimum of 70 subjects were required to detect statistically significant differences.

3.7 Arm Swing Velocity

The maximum arm swing velocity at the PWS did not show statistically significant differences between YAs and OAs ($p = 0.064$) (Figure 27 in Appendix C).

Table 2: Results for the RMS and HR of the head accelerations. The mean distance, in absolute value, between the optimal speed and the PWS, normalized by PWS, its 95% confidence interval, and its p-value are reported for each direction, for both young (YA) and old (OA) adults.

(a) RMS

Axis	Mean [95% CI]	p-value
YA		
A - P	0 [0, 0]	1
M - L	0.002 [0, 0.007]	1
V	0.184 [0.007, 0.361]	0.045
OA		
A - P	0.188 [0.050, 0.327]	0.019
M - L	0.075 [0.001, 0.150]	0.049
V	0.284 [0.108, 0.459]	0.011

(b) HR

Axis	Mean [95% CI]	p-value
YA		
A - P	0.065 [0.022, 0.107]	0.250
M - L	0.186 [-0.050, 0.423]	0.087
V	0.055 [0, 0.109]	0.500
OA		
A - P	0.124 [-0.128, 0.376]	0.243
M - L	0.047 [0.012, 0.083]	0.021
V	0.185 [-0.051, 0.420]	0.095

3.7.1 Hypothesis g

When the same velocity is imposed, older adults have the same maximum arm swing velocity as younger adults.

No statistically significant differences were observed in the maximum arm swing velocities between young and older adults for any trials with a fixed walking speed (Figure 8). The p-values for the trials at 0.8 m/s, 1.2 m/s, and 1.6 m/s were 0.393, 0.099, and 0.244, respectively. However, it is interesting that the confidence intervals of the means widen as the walking speed increases, as it indicates that the inter-subject differences amplify as the walking speed increases.

When young and old adults are required to walk at the same speed, they tend to exhibit similar maximum arm swing velocities.

The power analysis indicated that the minimum sample size required to detect a significant difference between age groups was n = 3 at both 0.8 m/s and 1.2 m/s, and n = 12 at 1.6 m/s.

4 DISCUSSION

This study aimed to examine how the CNS selects the optimal gait pattern, focusing on the cost function used to address motor redundancy. Key findings revealed that the preferred walking speed did not minimize metabolic cost for either age group, but it maximized harmonic ratios of head acceleration in both. Younger adults also minimized head acceleration amplitudes at their PWS, while older adults did not.

The results of hypotheses a and b suggest that elderly adults have a similar or lower metabolic cost of walking than younger adults when walking at the same speed or at their PWS. This finding directly contradicts numerous previous studies, which have consistently reported that older adults have a higher metabolic cost of walking compared to younger adults [5, 7, 27, 29–31, 34, 39–41, 59, 62–65]. This discrepancy may stem from a fundamental methodological difference: while these prior studies assessed participants walking on a treadmill, the present study evaluated participants walking overground. One possible explanation is that older adults exhibit a higher metabolic cost of walking (CoW) on a treadmill compared to overground, while younger adults do not [26, 66, 67]. This discrepancy may be caused by differing familiarization requirements or distinct neuromuscular adaptations to treadmill walking [23, 66].

Only three studies have employed a similar overground approach. Das Gupta et al. compared the metabolic cost of walking for older and younger adults overground at their PWS and also found no statistically significant difference in either PWS or NCoW [26]. Waters et al. conducted two studies in 1983 and 1988, with mixed results: in the first study, they reported that the net oxygen cost per meter walked $[ml/(kg*m)]$ at the PWS was higher for older adults than younger ones, though they attributed this to the lower PWS of the elderly [68]; however in the present study the two age groups walked at the same PWS. In their second study, testing three speeds (PWS, "slow", and "fast"), Waters et al. found that older adults (60–80) had a higher oxygen cost than adults (20–59) at PWS, no significant difference at the fast pace, and a lower cost at the slow pace [69]. However, differing speeds between age groups complicate comparisons. Matching velocities between the older adults and the adults may have aligned their findings with the present study.

A possible explanation for the lower metabolic cost in older adults at higher speeds shown in this study is a shift from fast-twitch (Type II) to energy-efficient, fatigue-resistant slow-twitch (Type I) muscle fibers, typically observed in older adults [70, 71]. Additionally, most older subjects were Dutch, known for higher fitness levels due to cycling prevalence, compared to younger subjects from southern and eastern Europe [72].

Table 3: Statistically significant differences in the mean step frequency (SF) and step length (SL) comparison between the different trials for young and old adults.

(a) Young Adults

Trial	Trial	p - value
SF		
0.8 m/s	PWS + 10%	0.033
0.8 m/s	1.6 m/s	0.019
PWS – 10%	PWS + 10%	0.037
PWS – 10%	1.6 m/s	0.008
PWS – 5%	1.6 m/s	0.028
SL		
0.8 m/s	PWS + 10%	0.033
0.8 m/s	1.6 m/s	0.019
PWS – 10%	PWS + 10%	0.017
PWS – 10%	1.6 m/s	0.005
PWS – 5%	1.6 m/s	0.052

(b) Old Adults

Trial	Trial	p - value
SF		
0.8 m/s	1.6 m/s	0.013
0.8 m/s	PWS + 10%	0.002
0.8 m/s	PWS + 5%	0.027
0.8 m/s	1.2 m/s	0.010
SL		
0.8 m/s	1.6 m/s	0.007
0.8 m/s	PWS + 10%	¡0.001
0.8 m/s	PWS + 5%	0.050
0.8 m/s	1.2 m/s	0.008
PWS – 10%	PWS + 10%	0.022
1.2 m/s	PWS + 10%	0.050

Interestingly, resting oxygen consumption rates between younger and older adults showed no statistically significant difference. Given that the trials with the older adults were conducted in cooler temperatures, which can increase resting metabolic rates as the body maintains a constant internal temperature [73], this finding suggests two possible interpretations: either that the age-related adaptations primarily affect walking efficiency, and thus the NCoW, or that the GCoW for older adults would have been lower if both age groups had been tested under similar weather conditions. Future research should explore this in more detail. For the current study, the NCoW results appear to provide the most meaningful insights for comparing metabolic costs between younger and older adults.

In younger and older adults, both the gross and net CoW were not minimized at the preferred walking speed, as indicated by the findings relative to the hypothesis c. These results are in contrast with previous studies [10, 20]. Two possible explanations could account for these findings. First, it's possible that subjects overestimated their PWS. Because the PWS trial was conducted first, participants may not have been fully acclimated to the experimental setting, or they may have been nervous, causing them to walk faster than their usual preferred pace. Second, minimizing the cost of walking may not be the main cost function chosen by the CNS to resolve the functional redundancy problem, but other factors might have a greater influence. This would imply that the CNS might optimize another parameter, leading to a choice of PWS that is not fully energy-efficient.

The results regarding head accelerations in hypothesis d indicate that young adults tend to minimize their head movements in the AP and ML directions while maximizing HRs in all directions at their PWS. These findings align with Latt et al., who observed that HRs peak at PWS in the V and AP directions [17], as well as with other studies showing that HRs are maximized at PWS across all directions [18, 19].
In contrast, older adults tended to choose a PWS that did not minimize the RMS of head accelerations. However, the HRs in the A-P and V directions indicate a more stable gait at their PWS, suggesting a prioritization of stability over movement smoothness.
These differences in RMS and HR outcomes can be explained by the distinct aspects of gait captured by each metric. RMS reflects the overall magnitude of acceleration fluctuations in each direction, where lower RMS values imply smoother or less intense movement, potentially aligning with energetically efficient speeds. However, RMS does not capture step-to-step consistency. In contrast, HR is a frequency-based measure that indicates the symmetry and rhythmicity of accelerations, with higher HR values representing a more regular, rhythmic, and stable gait.
For young adults, the alignment of RMS and HR results at the PWS suggests that their PWS naturally balances low-magnitude movement with a rhythmic, stable gait. This may be due to their generally better motor control and postural stability, allowing both RMS and HR to align closely with the PWS. For older adults, the mismatch between RMS and HR suggests that their PWS may not minimize movement intensity, as indicated by the RMS. Instead, it likely represents the speed that optimizes stability and gait symmetry, potentially as a compensatory mechanism for diminished balance control [25, 74]. In the M-L direction, older adults exhibited reduced balance at their PWS. While AP balance relies more on passive mechanisms, ML balance demands greater sensory input and environmental adaptability, enabling flexible adjustments to external changes rather than rigid control [17]. Declining vestibular function and increased neural noise in the central nervous system [25, 74] may further impair older adults' ability to maintain ML balance.

The step analysis of hypotheses e and f indicates that both younger and older adults adopted similar strategies for modulating walking speed, adapting both step frequency and step length as speed varied. This adaptation by the CNS aligns with findings from Ahuja et al. [16] and reveals that neither frequency nor length remained constant across different walking speeds. The degree of adjustment was more pronounced as the speed deviated further from the PWS. Both age groups showed a preference for modifying step frequency over step length, as evidenced by the higher intra-subject variance in frequency. In this study, an increase in speed was generally accompanied by an increase in step frequency and a decrease in step length for both groups. This pattern contrasts with Mirelman et al., who found that younger and older adults increased step length to walk faster [54], possibly due to the shorter 20-meter walking distance in their study, which may have influenced energy strategies differently compared to the 6-minute walks in the present study.
No differences were observed between younger and older adults in how they adjusted step frequency, which is consistent with findings by Fan et al. for PWS and slower speeds [75] and by Mazzà et al. for PWS [18]. However, Mazzà et al. noted that at higher speeds, older adults tended to vary their cadence more than younger adults. In terms of step length, this study found no group differences, which contrasts with prior studies where younger adults displayed longer steps than older adults [18, 75]. These discrepancies may be attributed to differences in experimental conditions, as previous studies involved shorter walking distances of less than 12 meters. Additionally, younger adults in those studies walked at faster speeds than older

adults, possibly because older adults accelerated more slowly and, within the short recorded distance, did not have enough time to reach their most comfortable speed.
Had the trial speeds been more distinct from one another, further differences might have emerged between the step frequencies and step lengths of the trials. For example, the trials at PWS with ±5% adjustments were quite close in speed. Furthermore, the high relative error in mean velocities for the 0.8 m/s and 1.6 m/s trials suggests that more accurate speed control might have yielded even clearer distinctions.
Since this study recorded only one step frequency per walking speed, it does not clarify whether the CNS also optimizes the frequency to decrease the metabolic cost or increase stability. Further research is needed to understand the CNS's criteria for achieving optimal walking conditions.

Finally, no differences were observed between young and older adults in arm swing velocity at the same speeds or at their preferred walking speeds, as shown by the results of hypothesis g. This finding confirms that previously reported differences may be attributed to variations in walking speed rather than age-based differences in arm movement [53, 54].

4.1 Limitations

Several limitations should be considered when interpreting the findings of this study. First, the sample size was limited, which may affect the generalizability of the results. The power analyses performed on all the statistical tests show that a larger sample size is needed to increase the reliability of the results. The elderly recruited in this study all showed a high general level of health. It is not to be excluded that older or less fit individuals might show different results. Additionally, participant nationality may have introduced bias, as the older subjects were predominantly Dutch, while the younger participants were primarily from southern and eastern European countries. The Dutch population is known for its high prevalence of cycling and generally superior fitness levels compared to other nationalities [72]. The order of the trials may have also influenced the results, as the PWS was always presented first, potentially causing nervousness or uncertainty among the participants, which could have led to a higher metabolic cost. Furthermore, the elderly tended to talk more during the trials and this could have caused some leakage from the mask, leading to inaccurate measurements. This hypothesis was tested by comparing the slope coefficients of the linear regressions for VO_2 and VCO_2 between YAs and OAs. No statistically significant differences were found ($p = 0.420$ and $p = 0.683$, respectively), confirming that VO_2 and VCO_2 did not progressively decrease due to a mask leakage. Environmental factors such as varying temperatures and weather conditions between testing days could have also impacted the results. Lastly, the pacing cart used to control walking speeds had limited precision, which might have affected the accuracy of the average walking speed across trials.

5 CONCLUSION

This study investigated strategies to solve the redundancy problem in younger and older adults, focusing on metabolic cost, head accelerations, and gait adjustments at different speeds.
Findings revealed that both age groups showed similar metabolic costs when walking overground at fixed speeds or their preferred walking speed, challenging previous treadmill-based studies that often report higher costs for older adults. Moreover, the metabolic cost may not be the optimization criterion of walking for either younger or older adults.
The analysis of head accelerations indicated that while younger adults managed to balance low-magnitude and rhythmic movement at their PWS, older adults exhibited a less fluid and more jerky gait, suggesting a shift toward prioritizing stability over movement intensity. These results suggest that stability might play a more significant role in selecting the optimal locomotion pattern at the preferred walking speed than the metabolic cost of walking.
Both groups preferred adjusting step frequency over step length to manage different walking speeds, but further research is necessary to fully comprehend how optimal gait is achieved.

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Appendices

A Materials

Figure 9: COSMED K5, adapted from [58]

Figure 10: Cometa MiniX

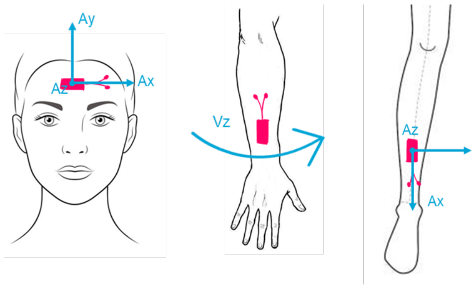


Figure 11: Position of the IMUs

Figure 12: Measuring wheel

Figure 13: N317 Retoo Bike Computer, adapted from [76]

B Subjects

Table 4: Demographic and anthropometric data of the subjects

Sub	Sex	Age [years]	Mass [kg]	Height [cm]	Nat
YA					
1	M	23	71	178	ES
2	F	24	60	165	ES
3	F	26	71	170	ES
4	M	23	86	189	NL
5	F	27	80	160	PL
6	F	25	67	153	IN
7	F	25	52	154	ES
8	M	28	80	187	NL
9	M	25	83	172	ES
10	M	27	88	183	NL
mean:	5F, 5M	25.10	73.7	171.1	
OA					
1	F	77	59	168	NL
2	F	73	68	160	NL
3	M	77	66	180	NL
4	F	69	64	163	GB
5	M	76	85	180	NL
mean:	3F, 2M	74.4	68.4	170.2	

Table 5: Preferred Walking Speed per subject

Subject (YA)	PWS [m/s]	Subject (OA)	PWS [m/s]
1	1.12	1	1.27
2	1.21	2	1.39
3	1.39	3	1.10
4	1.16	4	1.51
5	1.30	5	1.53
6	1.29		
7	1.32		
8	1.43		
9	1.44		
10	1.51		
mean:	1.32	mean:	1.36

Table 6: Overview of the available data per subject and the minimum and maximum temperature of the session day [77]. * = incomplete; ** = excluded.

Subject	COSMED	IMU head	IMU arm	IMU leg	T
YA					
1	✓	✓	✓	✓	18 - 20 °C
2	✓	✓	✓	✓	21 - 23 °C
3	✓				19 - 20 °C
4	✓**				20 - 22 °C
5	✓				16 - 20 °C
6	✓*	✓*	✓*	✓*	20 - 22 °C
7	✓				18 - 20 °C
8	✓				26 - 28 °C
9	✓				19 - 22 °C
10	✓	✓	✓	✓	17 - 19 °C
OA					
1	✓	✓	✓	✓	11 - 12 °C
2	✓	✓	✓	✓	10 - 13 °C
3	✓*	✓*	✓*	✓*	13 - 15 °C
4	✓	✓	✓	✓	9 - 13 °C
5	✓	✓	✓	✓	13 - 15 °C

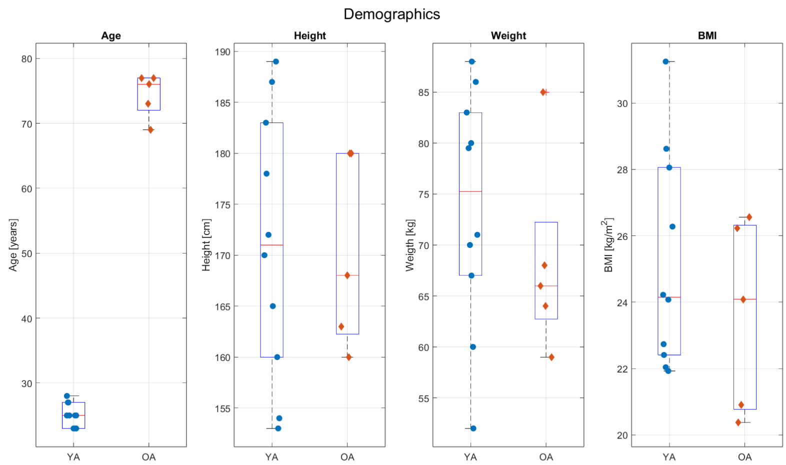


Figure 14: Boxplots of the age [years], height [cm], weight [kg], and BMI [kg/m^2] of the young (blue dots) and old (orange diamonds) participants.

C Graph

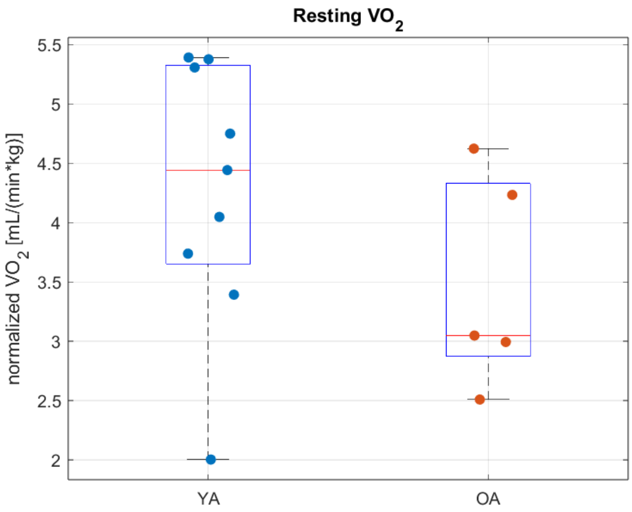


Figure 15: Boxplot of the resting oxygen consumption rate of the Young Adults (YA) and the Old Adults (OA).

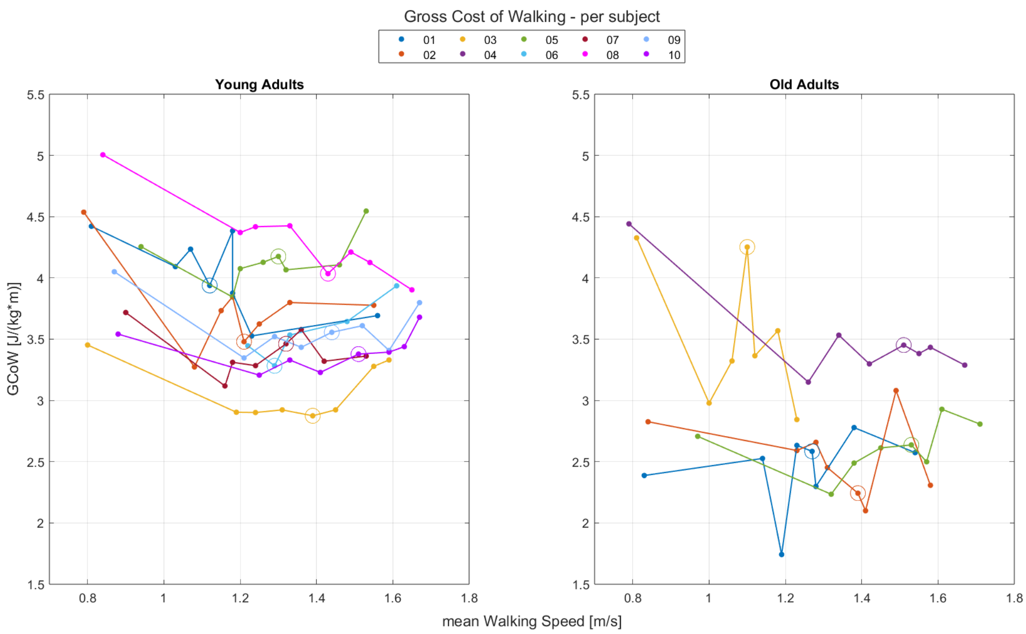


Figure 16: Mean Gross Cost of Walking (GCoW) per trial, for Young and Old Adults. The PWS trial is marked with a circle.

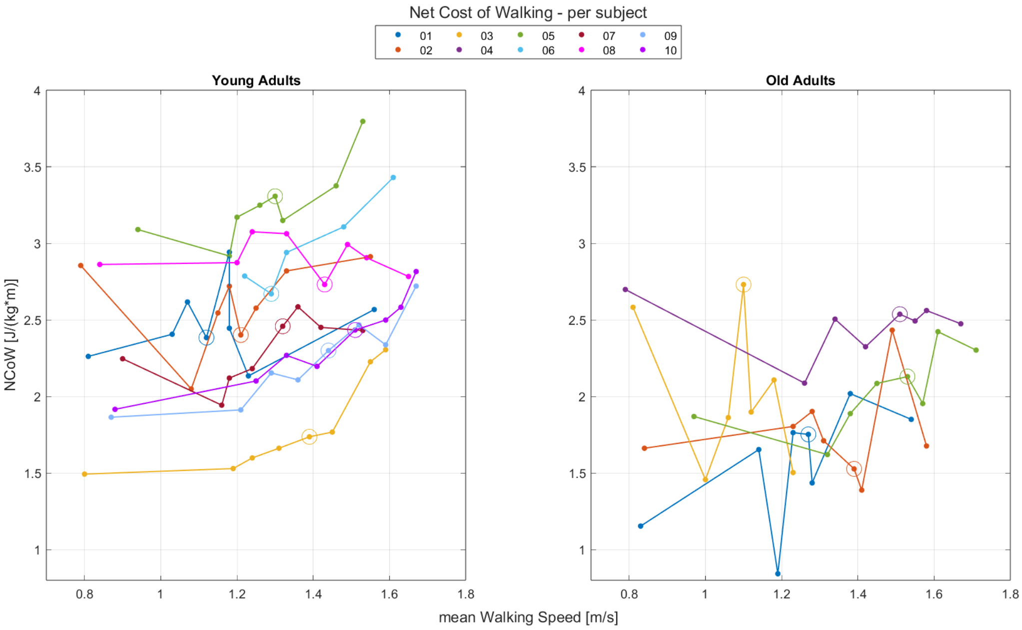


Figure 17: Mean Net Cost of Walking (NCoW) per trial, for Young and Old Adults. The PWS trial is marked with a circle.

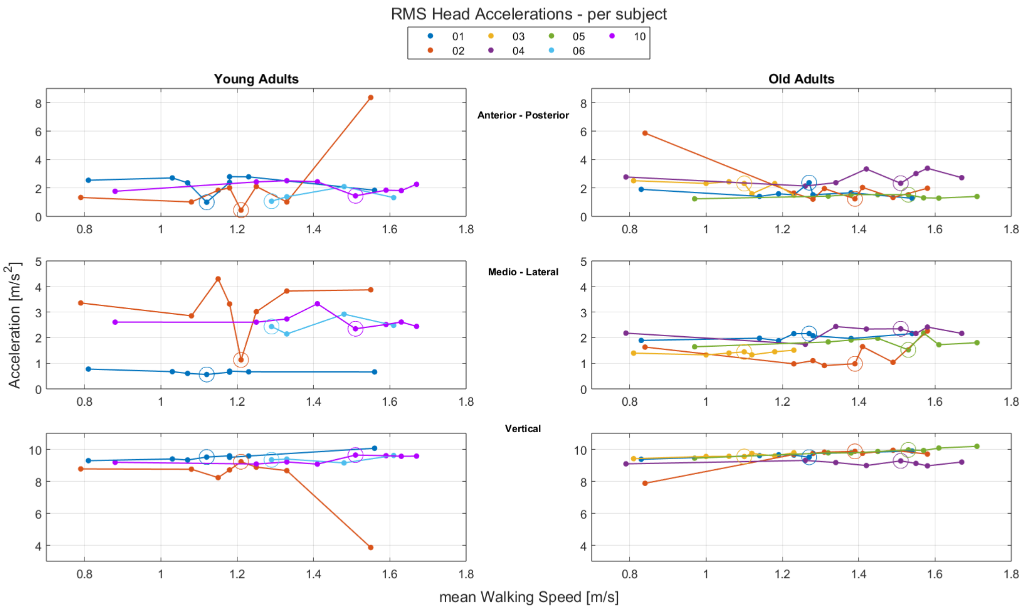


Figure 18: Root Mean Square (RMS) of head accelerations per trial, for Young and Old Adults. The PWS trial is marked with a circle.

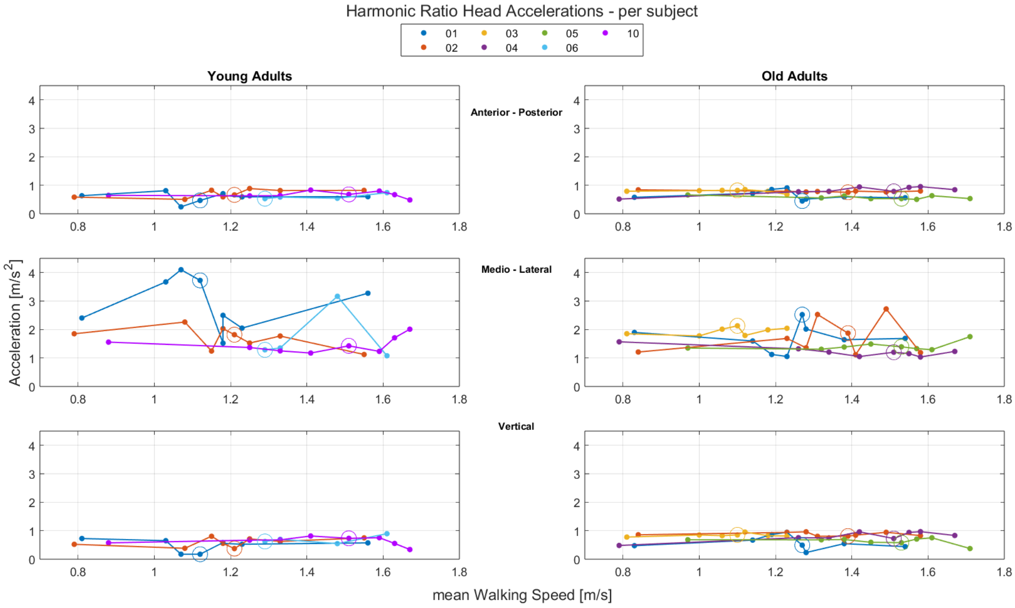
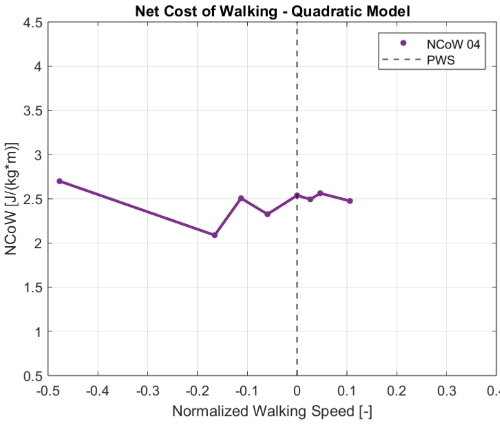
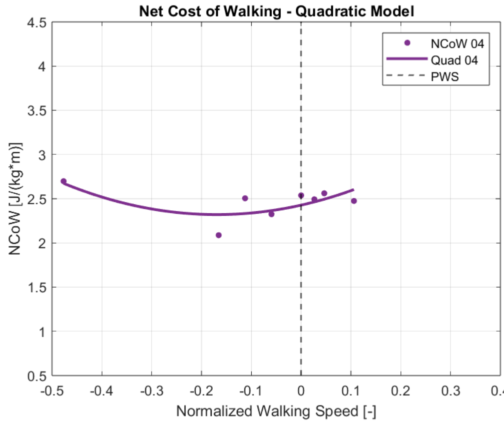


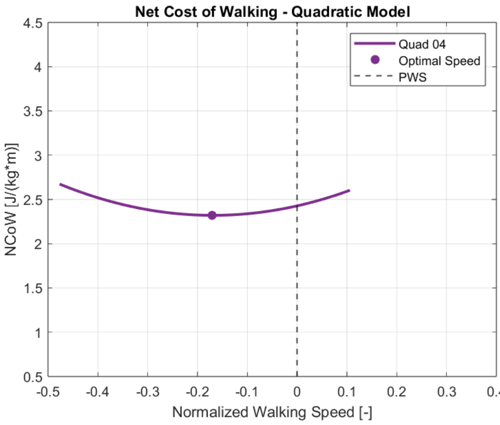
Figure 19: Mean Harmonic Ratio (HR) of head accelerations per trial, for Young and Old Adults. The PWS trial is marked with a circle.



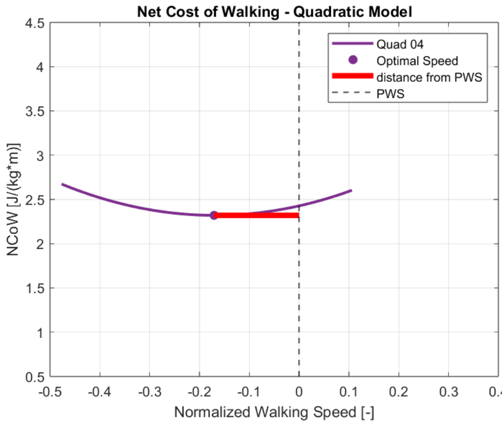
(a)



(b)



(c)



(d)

Figure 20: Quadratic model representing the relationship between Net Cost of Walking (NCoW) and normalized walking speed for Subject 4 in the older adult group. The x-axis is normalized to the participant's Preferred Walking Speed (PWS). 20a shows the NCoW, 20b shows the quadratic curve fitted to the data, 20c shows the optimal speed, which is the speed that minimizes the quadratic curve, and 20d shows the distance between the optimal speed and the PWS.

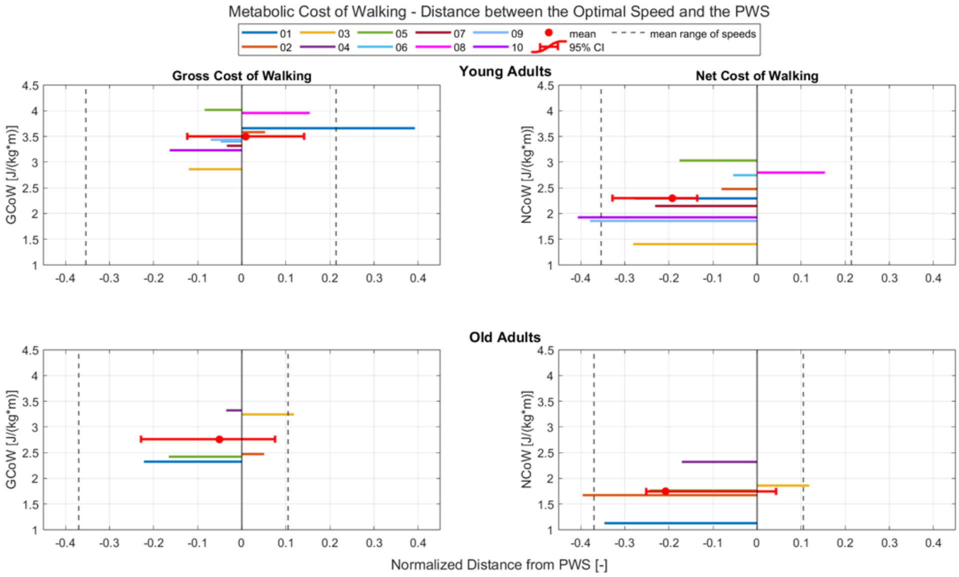


Figure 21: Distance between the optimal walking speed, determined as the speed that minimizes the Gross (left) and Net (right) Cost of Walking (CoW), and the preferred walking speed (PWS) for each subject, normalized by PWS. Results for young adults (YA) are displayed on the top row and those for old adults (OA) on the lower one. The mean distance and its 95% confidence interval are marked in red. The dotted black vertical lines represent the mean distance from the speed extremities (0.8 m/s and 1.6 m/s). A negative mean distance would indicate that the optimal speed is slower than the PWS.

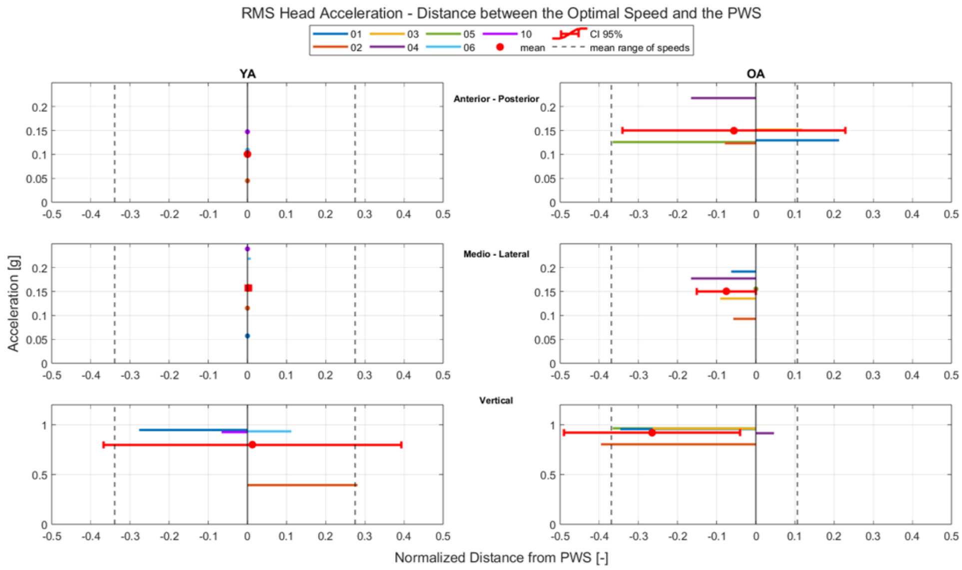


Figure 22: Distance between the optimal walking speed, determined as the speed that minimizes the Root Mean Square (RMS) of the head accelerations, and the preferred walking speed (PWS) for each subject, normalized by PWS. Results for young adults (YA) are displayed on the left and those for old adults (OA) on the right. The top row represents data from the anterior-posterior (A-P) direction, the middle one those relative to the medio-lateral (M-L) direction, while the bottom row shows data from the vertical (V) direction. The mean distance and its 95% confidence interval are marked in red. The dotted black vertical lines represent the mean distance from the speed extremities (0.8 m/s and 1.6 m/s). A negative mean distance would indicate that the optimal speed is slower than the PWS.

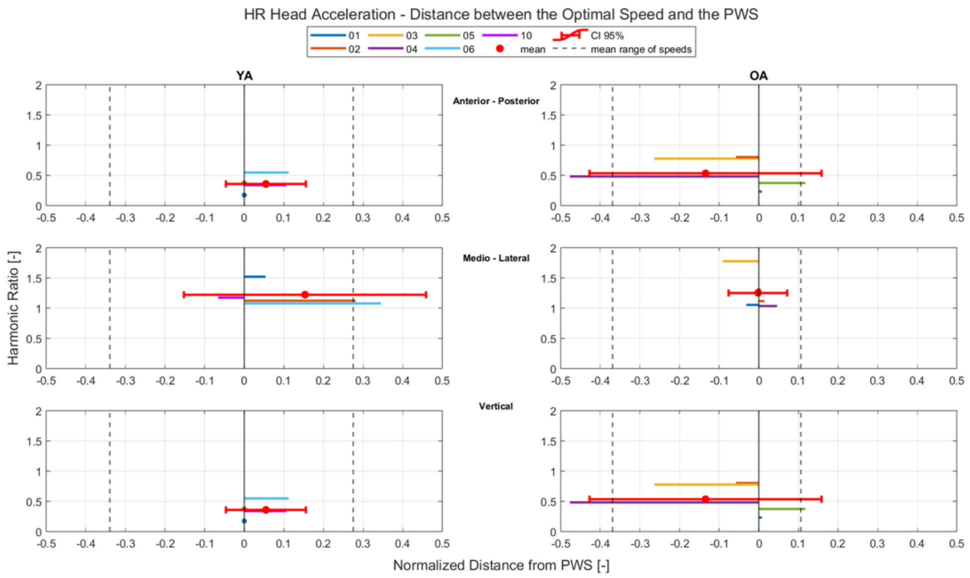


Figure 23: Distance between the optimal walking speed, determined as the speed that minimizes the Harmonic Ratio (HR) of the head accelerations, and the preferred walking speed (PWS) for each subject, normalized by PWS. Results for young adults (YA) are displayed on the left and those for old adults (OA) on the right. The top row represents data from the anterior-posterior (A-P) direction, the middle one those relative to the medio-lateral (M-L) direction, while the bottom row shows data from the vertical (V) direction. The mean distance and its 95% confidence interval are marked in red. The dotted black vertical lines represent the mean distance from the speed extremities (0.8 m/s and 1.6 m/s). A negative mean distance would indicate that the optimal speed is slower than the PWS.

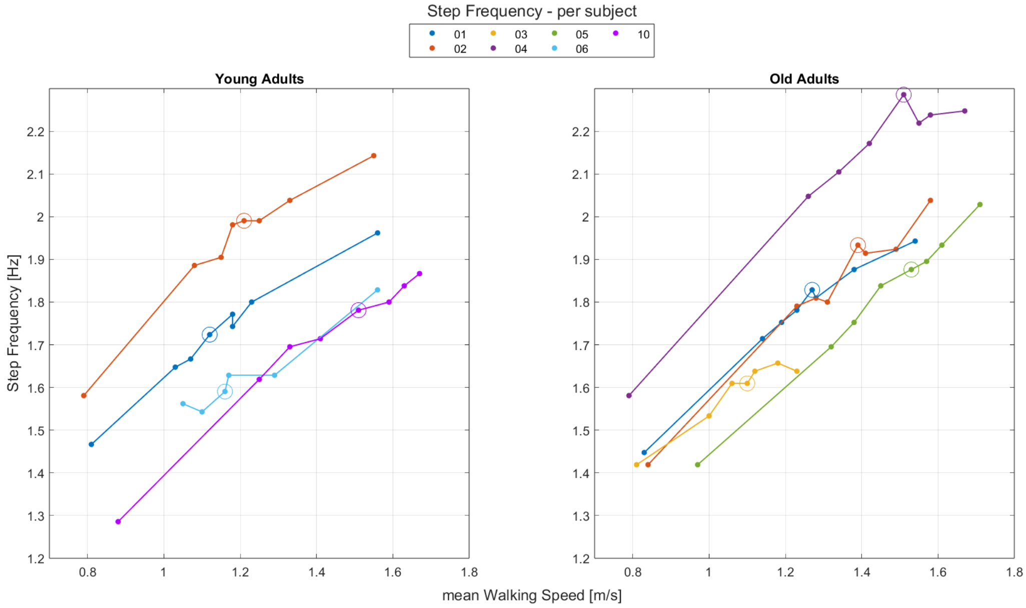
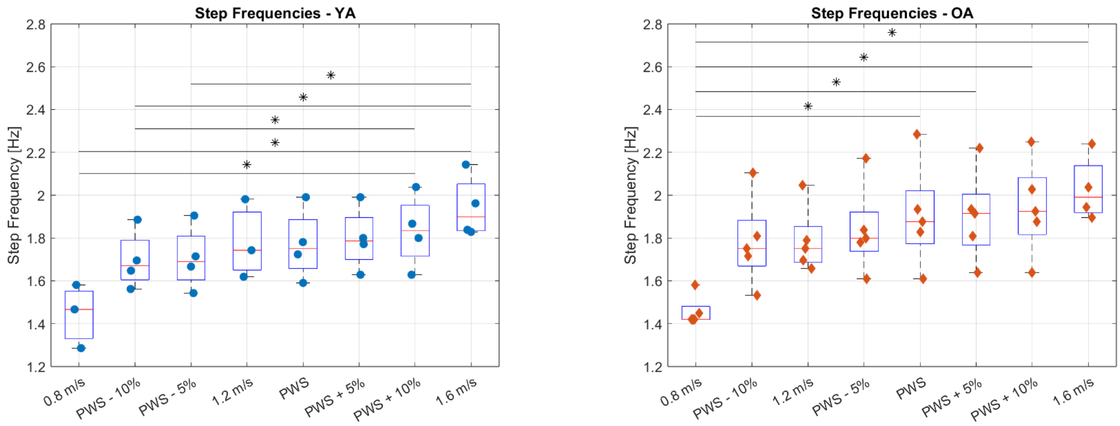
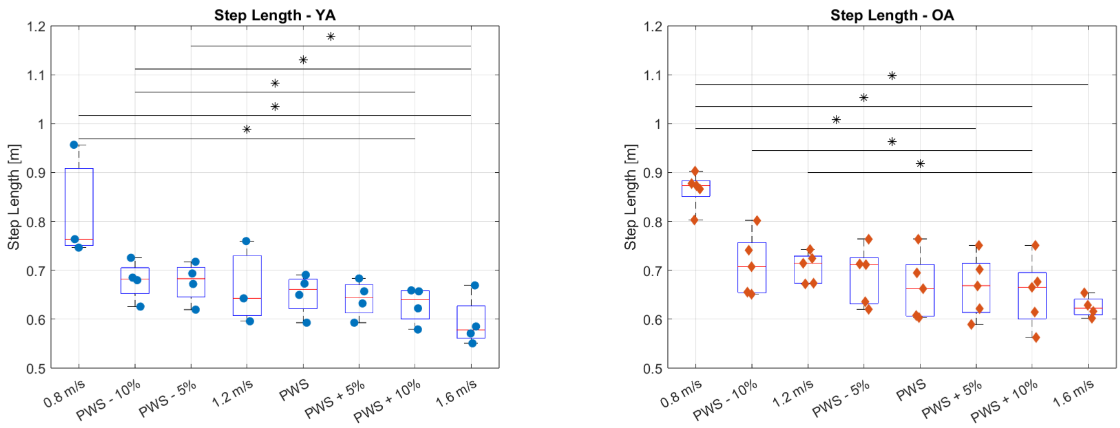


Figure 24: Mean Step Frequency per trial, for Young and Old Adults. The PWS trial is marked with a circle.



(a) YA (b) OA

Figure 25: Boxplots of step frequency across different walking speeds. The individual dots represent each subject's data for each speed condition. Horizontal bars above the boxplots denote statistically significant differences between specific conditions.



(a) YA (b) OA

Figure 26: Boxplots of step length across different walking speeds. The individual dots represent each subject's data for each speed condition. Horizontal bars above the boxplots denote statistically significant differences between specific conditions.

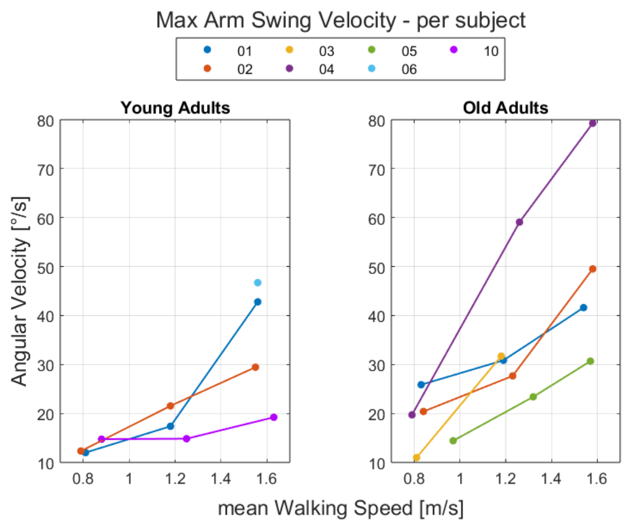


Figure 27: Mean Arm Swing per trial, for Young and Old Adults.

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