Cureus
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Open Access Original Article

Changes in Iliopsoas Muscle Activity and Hip/Pelvic Kinematics With Variations in Step Length and Cadence at a Fixed Walking Speed

Review began 04/30/2025
Review ended 05/13/2025
Published 05/24/2025

DOI: 10.7759/cureus.84737

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Abstract

Background: Older adults often compensate for reduced mobility by adopting shorter steps and a faster cadence, which may alter muscle recruitment and joint kinematics. The iliopsoas, a key muscle for leg swing and posture control, is particularly affected. Understanding its role across different gait patterns may help improve gait training and fall prevention.

Purpose: To investigate how step length and cadence influence iliopsoas muscle activation and hip/pelvic kinematics during treadmill walking at a constant speed.

Methods: Ten healthy young men walked on a treadmill at 5 km/h under three conditions: (1) Step-Length Increased Walking (SL), longer steps with lower cadence; (2) Normal Walking (N), self-selected gait; and (3) Pitch Increased Walking (P), shorter steps with higher cadence. Iliopsoas activity was measured using surface electromyography, and hip/pelvic motion was captured using 3D motion analysis. The gait cycles were divided into four functional phases for phase-specific analysis. Nonparametric tests were used to compare muscle activity across conditions.

Results: Iliopsoas activity was significantly higher during the early (Phase 1) and late stance (Phase 2) in both the SL and P conditions than in the N condition. During the late swing (Phase 4) period, the SL condition elicited the greatest iliopsoas activation accompanied by increased hip flexion and pelvic tilt. This phase showed the strongest effect size among all phases, highlighting the critical role of iliopsoas during terminal swing. P walking also showed elevated iliopsoas activation during the stance, likely due to the faster step cycle.

Conclusion: The findings suggest that step length and cadence influence the timing and magnitude of iliopsoas activation. Longer steps increase activation during the late swing phase, while a higher cadence enhances activation during stance. Phase-specific training targeting iliopsoas strength and hip flexion mobility may help improve gait performance.

Categories: Physical Medicine & Rehabilitation
Keywords: cadence, gait kinematics, hip flexion, iliopsoas activation, pelvic tilt, step length, surface emg, walking speed

Introduction

Walking is a fundamental form of human mobility, and preserving this ability is essential to maintain independence and quality of life in older adults. A decline in walking function, characterized by reduced walking speed, shorter step length, and increased instability, is a common age-related change and is associated with limitations in daily activities, reduced mobility, cognitive impairment, increased fall risk, and lower survival rates [1,2].

Age-related reductions in muscle strength, joint range of motion, and balance contribute to typical gait changes in the elderly, including slower walking speed, shorter step length, longer double-support duration, increased step width, and higher cadence (step frequency) [3-5]. These changes not only increase fall risk but also compromise functional autonomy, underscoring the importance of understanding the underlying mechanisms and developing effective interventions [2].

Walking speed is the product of the step length and cadence [6]. In young adults, increased speed is primarily achieved by lengthening steps. However, older adults often have a reduced capacity to do so and instead compensate by increasing cadence [5,7,8]. This adaptation may enhance stability and reduce fall risk [7].

The hip flexor muscles, particularly the iliopsoas, play a critical role in initiating leg swings and controlling step length [9-11]. In addition to its role in hip flexion, the iliopsoas also contributes to lumbar and pelvic stabilization and postural control [12,13]. Although the iliopsoas is known to be important for leg swing and posture, how its activation responds to specific gait modifications, particularly changes in step length and cadence, remains incompletely understood.

Although many studies have examined the effects of walking speed, few have independently manipulated step length and cadence [9,14]. Since the gait in the elderly is often characterized by both a shorter step length and higher cadence [5,7,8], clarifying how these factors affect iliopsoas activation is vital for designing targeted fall prevention and gait rehabilitation strategies.

This study aimed to simulate the gait characteristics commonly observed in older adults, namely decreased step length and increased cadence, and to investigate their effects on iliopsoas electromyographic (EMG) activity and hip/pelvic kinematics. Three walking conditions were evaluated at a constant speed of 5 km/h: (1) Step-Length Increased Walking (SL): Long step length, low cadence; (2) Normal Walking (N): Self-selected step length and cadence; (3) Pitch Increased Walking (P): Short step length, high cadence. Under each condition, surface EMG data of the iliopsoas and hip/pelvic kinematics were recorded. The objective was to determine how gait pattern modifications affect iliopsoas activation and joint movements. These three conditions were chosen to reflect gait characteristics frequently observed in older adults. The SL condition represents a gait strategy emphasizing step length, often used in rehabilitation. The P condition simulates a compensatory gait with short steps and high cadence commonly seen in elderly individuals. The N condition serves as a natural baseline. This setup allows us to isolate the effects of step length and cadence while keeping speed constant.

Materials And Methods

Participants

Ten healthy adult men participated in this study (mean age: 25.0 ± 3.9 years; mean weight: 64.7 ± 3.7 kg; mean height: 175.7 ± 5.3 cm). All participants received both verbal and written explanations of the study and provided written informed consent. The study was approved by the Kanazawa Orthopaedic Sports Medicine Clinic Ethics Committee (approval number: Kanazawa-OSMC-2022-008) and conducted in accordance with the Declaration of Helsinki. Participants with a history of orthopedic, neurological, or cardiovascular conditions that could affect gait or muscle activation were excluded. All patients reported being physically active and free of lower limb injuries in the past six months.

Experimental protocol

The three walking conditions (SL, N, and P) were designed to simulate clinically relevant gait patterns. Specifically, SL represents a gait with enhanced step length, P represents a high-cadence gait typically used to compensate for reduced mobility, and N represents a baseline of self-selected walking. This experimental design enables the independent analysis of step length and cadence effects on muscle activity and joint kinematics under a fixed speed. Participants walked barefoot on a treadmill (AUTORUNNER; Minato, Osaka, Japan) under three gait conditions: (1) SL: step length ~1.0 m; cadence: 80 steps/min; (2) N: self-selected step length and cadence; (3) P: step length ~0.4 m; cadence: 190 steps/min. A metronome was used to standardize cadence in SL and P (80 and 190 beats/min, respectively). No external cues were provided for N. The treadmill speed was fixed at 5 km/h for all trials. Surface EMG data of the iliopsoas and hip/pelvic kinematics were simultaneously recorded.

Walking procedure

Prior to data collection, participants completed a familiarization session at least one week in advance. On the test day, a standardized warm-up (two minutes per condition: SL, N, and P) was conducted. The three walking conditions were performed in a randomized order. Each trial recorded at least five gait cycles (approximately 10 seconds per trial). A three-minute rest was provided between trials to prevent fatigue. All sessions were conducted in a quiet, temperature-controlled lab (22°C) at the same time of day for each participant. A single examiner performed all procedures to avoid interrater variability. Furthermore, to enhance measurement reliability, all experimental procedures, including electrode placement, ultrasound verification, and motion capture setup, were performed by the same experienced examiner. This approach was intended to reduce inter-rater variability and ensure consistency across trials.

Motion capture and kinematic analysis

Reflective markers were placed on 14 anatomical landmarks: the bilateral anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), greater trochanters, lateral femoral epicondyles, lateral malleoli, calcanei, and fifth metatarsal bases. Marker trajectories were recorded using a four-camera 3D motion capture system (UM-CAT; Unimec, Tokyo, Japan) at 200 Hz. Data were processed using Kine Analyzer software (Kissei Comtec, Nagano, Japan). A second-order low-pass Butterworth filter (cut-off: 8 Hz) was applied. The global coordinate system was defined relative to the treadmill as follows: X-axis (anteroposterior), Y-axis (mediolateral), and Z-axis (vertical).

A gait cycle was defined as the time from right foot contact to the subsequent right foot contact. For analysis, three consecutive gait cycles were averaged for each condition (SL, N, and P). Step length was measured as the distance between successive contralateral foot contacts, and cadence was calculated in steps per minute. Hip flexion and extension angles were calculated from the vector connecting the right ASIS to the PSIS (pelvis) and the vector from the right greater trochanter to the lateral femoral epicondyle (thigh). The pelvic anterior tilt angle was computed as the angle between the pelvic plane (defined by the bilateral ASIS and PSIS) and the vertical plane. Angular velocity was derived by differentiating the joint angle time series. Angular acceleration was then calculated as the second derivative of the joint angle time series, representing the rate of change of angular velocity over time. Although test-retest procedures were conducted during pilot sessions to confirm consistency in motion capture and EMG data collection, these results were not recorded as formal data. However, all measurements were performed by the same experienced examiner using standardized protocols, which we believe helped maintain procedural reliability across participants.

Surface EMG

Surface EMG signals of the right iliopsoas were collected using active electrodes (10 × 10 mm; 10 mm inter-electrode distance) and a 16-bit amplifier (MQ8/16; Kissei Comtec, Nagano, Japan). The electrodes were placed 3-5 cm distal to the ASIS and aligned with the muscle fibers. The location was confirmed using ultrasound imaging (LOGIQ P5; GE Healthcare, Chicago, USA) of the lower inguinal region [15,16]. Although the iliopsoas is a deep muscle, we used a validated surface EMG technique based on our previous studies [15,16], which identified a specific electrode placement region in the inguinal area that minimizes crosstalk and enables reliable detection of iliopsoas activity. The reference electrode was attached to the right patella. The skin was shaved and cleaned with alcohol before electrode placement. Standardized electrode placement protocols were used to minimize crosstalk between the surrounding muscles. All placements and ultrasound verifications were performed by the same examiner.

EMG signals were recorded across three continuous gait cycles per condition (SL, N, and P). The signals were full-wave rectified and filtered using a 10-1000 Hz band-pass filter, as recommended by Andersson et al. [14]. Root mean square (RMS) values were calculated at 5% intervals across the normalized gait cycle (0-100%).

Following the walking trials, a maximum voluntary isometric contraction (MVIC) test was conducted for signal normalization. The participants lay in a supine position (hip at 0°, knee at 90°) and exerted maximal hip flexion force for five seconds. The peak RMS values from three trials were used for normalization. The participants received strong verbal encouragement and were monitored for compensatory trunk movements.

Gait phase segmentation

The right gait cycle was segmented into four functional phases based on foot contact and hip angular velocity: (1) Phase 1 (early stance), from right foot contact to zero hip extension angular velocity; (2) Phase 2 (late stance), from zero extension velocity to right toe-off; (3) Phase 3 (early swing), from toe-off to the first zero crossing of the hip flexion angular velocity; (4) Phase 4 (late swing), from this point to the next right foot contact. These divisions reflect the typical activation pattern of the iliopsoas, which is active during the pre-swing and throughout the swing phase, particularly during initial swing (hip flexion) and terminal swing (hip extension) [14,17]. This segmentation allowed for phase-specific EMG analysis across the SL, N, and P conditions.

Statistical analysis

Data are reported as mean ± standard deviation (SD). The Shapiro-Wilk test was used to assess the normality of the EMG data. As the data were non-normally distributed ($p < 0.05$), the Friedman test was used to compare iliopsoas activity across gait conditions (N, SL, and P) within each gait phase. When significant, post hoc pairwise comparisons were performed using the Dunn-Bonferroni method. Kendall's W was calculated to estimate effect size. Statistical significance was set at $p < 0.05$. All analyses were conducted using IBM SPSS Statistics for Windows, Version 27.0 (Released 2020; IBM Corp., Armonk, NY, USA).

Based on the observed effect sizes from the Friedman test (Kendall's W), a priori power analysis was performed using G*Power (v3.1) to estimate the minimum required sample size. For the largest effect observed in Phase 4 ($W = 0.91$), a sample size of six participants would be sufficient to achieve 80% power at $\alpha = 0.05$. For moderate effects observed in Phases 1 ($W = 0.63$) and 2 ($W = 0.49$), the required sample sizes were estimated at 9 and 12, respectively. These results support the adequacy of the current sample size ($n = 10$) for detecting moderate to strong effects across the key gait phases in N, SL, and P.

Results

EMG analysis

The normality of the EMG data was assessed using the Shapiro-Wilk test. Temporal boundaries for each gait phase were calculated as a percentage of the gait cycle in the N, P, and SL conditions. Specifically, for Phase

1, durations were as follows: N: 0-49.7 ± 2.3%, P: 0-46.8 ± 1.6%, SL: 0-50.4 ± 2.5%; for Phase 2, N: 49.7 ± 2.3%-59.3 ± 1.6%, P: 46.8 ± 1.6%-60.8 ± 1.6%, SL: 50.4 ± 2.5%-55.4 ± 1.8%; for Phase 3, N: 59.3 ± 1.6%-83.8 ± 2.5%, P: 60.8 ± 1.6%-86.5 ± 1.2%, SL: 55.4 ± 1.8%-80.5 ± 3.0%; and for Phase 4, N: 83.8 ± 2.5%-100%, P: 86.5 ± 1.2%-100%, SL: 80.5 ± 3.0%-100%.

As all conditions violated the assumption of normality ($p < 0.05$), the Friedman test was used to compare iliopsoas activity across N, SL, and P in each of the four gait phases. Box plots were generated for each phase to visualize distributions and identify outliers (Figure 1, Table 1). No extreme outliers were found in any condition.

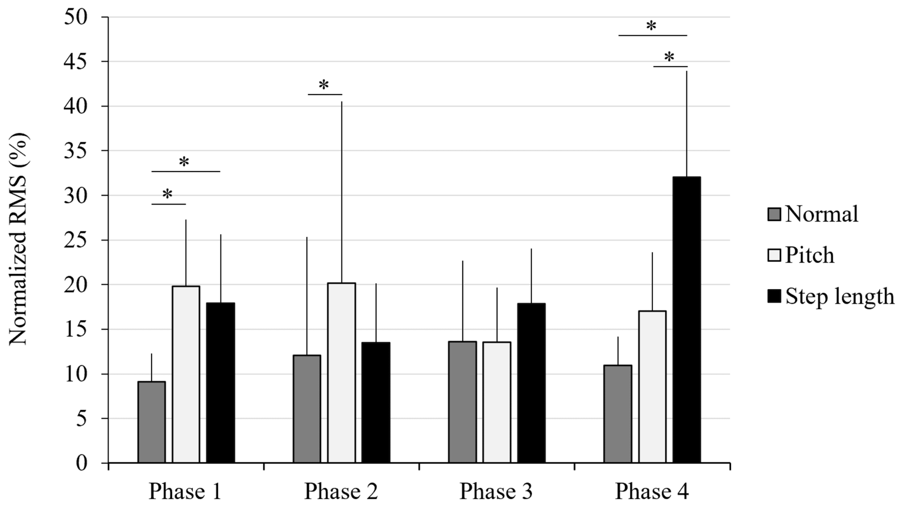

FIGURE 1: Gait phase-specific comparison of iliopsoas muscle activity across three walking conditions.

Mean normalized EMG activity of the iliopsoas muscle during four distinct gait phases (Phases 1–4) under three walking conditions (N, P, and SL). Asterisks denote statistically significant differences between conditions, as determined by the Friedman test followed by Dunn–Bonferroni post hoc analysis (*$p < 0.05$). Error bars indicate standard deviations. Significant differences in iliopsoas activity across conditions were found in Phase 1, Phase 2, and Phase 4, as shown in Table 1. No significant differences were observed in Phase 3 (early swing).

EMG: electromyographic; RMS: root mean square; SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

Gait phase	χ^2 (df = 2)	p-value	Kendall's W	Effect size interpretation
Phase 1	12.6	0.002**	0.63	Small–moderate
Phase 2	9.89	0.007**	0.49	Small
Phase 3	–	>0.05	–	Not significant
Phase 4	18.2	0.001***	0.91	Moderate–strong

TABLE 1: Summary of Friedman test results.

Summary of Friedman test results for iliopsoas activity across the three walking conditions during each gait phase. Significant differences were observed in Phases 1, 2, and 4. **$p < 0.01$, ***$p < 0.001$ indicate statistically significant predictors.

Effect sizes indicated small-to-moderate effects in Phases 1 and 2, while Phase 4 showed a relatively strong effect, where joint kinematics varied the most across conditions. According to the post hoc Dunn-Bonferroni results (Table 2), iliopsoas activity in Phase 1 was significantly higher in both P and SL compared to N. In Phase 2, P showed greater activity than N. In Phase 4, SL showed significantly higher activity than both N and P.

Gait phase	Comparison	p-value	Interpretation
Phase 1	P vs. N	0.002**	P > N
Phase 1	SL vs. N	0.022*	SL > N
Phase 2	P vs. N	0.008**	P > N
Phase 4	SL vs. N	<0.001***	SL > N
Phase 4	SL vs. P	0.042*	SL > P

TABLE 2: Post hoc Dunn–Bonferroni results.

Pairwise comparisons were performed using the Dunn–Bonferroni post hoc test. Significant differences ($p < 0.05$) were observed between Phases 1, 2, and 4. *$p < 0.05$, **$p < 0.01$, ***$p < 0.001$ indicate statistically significant predictors.

These findings indicate that a higher cadence (P) increases iliopsoas activation during early and late stance (Phases 1 and 2), while a longer step length (SL) leads to elevated activity during late swing (Phase 4), possibly due to greater mechanical demands for limb advancement. No significant differences were found in Phase 3.

Kinematic analysis

Figures *2*-*5* present the time-series plots of hip flexion angle, hip flexion angular velocity, pelvic anterior tilt angle, and pelvic anterior tilt angular velocity across the gait cycle for N, P, and SL. Regarding hip flexion-extension angle, SL showed greater hip extension in the late stance phase and larger hip flexion angles in the early and late swing phases compared to N and P (Figure *2*). Regarding hip flexion-extension angular velocity, SL maintained a flexion-directed angular velocity in the late swing phase, whereas N and P showed a reversal toward extension (Figure *3*). As for pelvic anterior tilt angle, SL walking exhibited a greater anterior tilt during the late swing phase (90-100%), which distinguished it from the other walking conditions (Figure *4*). Furthermore, the highest anterior tilt angular velocity during the late swing was observed in SL walking, suggesting a more dynamic pelvic motion compared to N and P walking (Figure *5*).

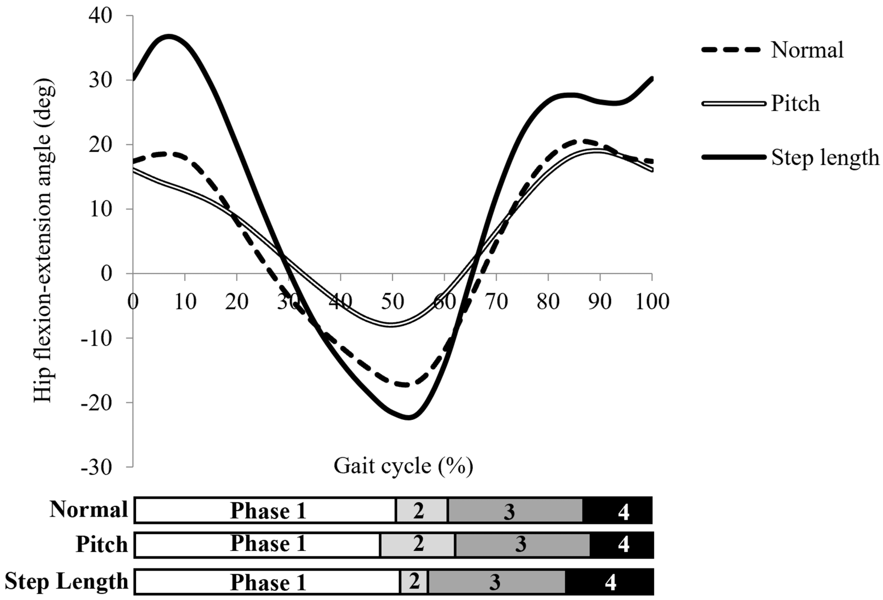

FIGURE 2: Time-series plots of hip flexion-extension angle across the gait cycle for each walking condition.

The vertical axis represents the hip flexion–extension angle (°), and the horizontal axis represents the gait cycle (%). Positive values indicate flexion-directed degree, while negative values represent extension-directed degree. SL showed greater hip extension in late stance and larger hip flexion angles in early and late swing phases compared to N and P. The bars at the bottom represent the gait phases: Phase 1 (white), Phase 2 (light gray), Phase 3 (dark gray), and Phase 4 (black) for each condition (N, P, SL). The curves show the mean values of 10 participants per condition.

SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

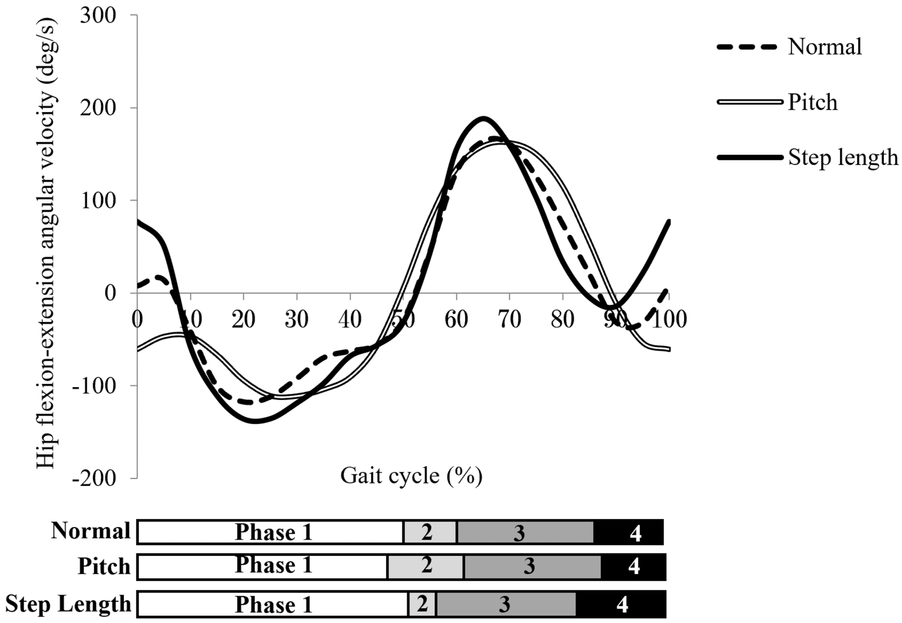

FIGURE 3: Time-series plots of hip flexion-extension angular velocity across the gait cycle for each walking condition.

The vertical axis represents the hip flexion-extension angular velocity (degrees/second), and the horizontal axis represents the gait cycle (%). Positive values indicate flexion-directed velocity, while negative values represent extension-directed velocity. SL maintained a flexion-directed angular velocity in the late swing phase, whereas N and P showed a reversal toward extension. The bars at the bottom of the figure represent the gait phases: Phase 1 (white), Phase 2 (light gray), Phase 3 (dark gray), and Phase 4 (black) for each condition (N, P, SL). The curves represent the mean values of 10 participants for each condition.

SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

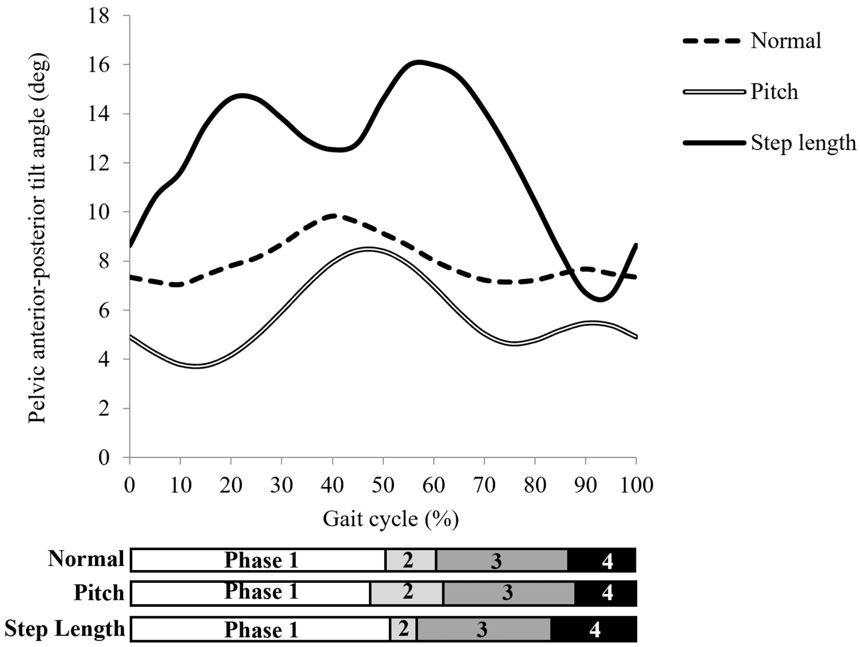

FIGURE 4: Time-series plots of pelvic anterior tilt angle across the gait cycle for each walking condition.

The vertical axis represents the pelvic anterior tilt angle (degrees), and the horizontal axis represents the gait cycle (%). Positive values indicate anterior tilt-directed degree, while negative values represent posterior tilt-directed degree. SL showed greater pelvic anterior tilt in the late swing phase compared to N and P. The bars at the bottom of the figure represent the gait phases: Phase 1 (white), Phase 2 (light gray), Phase 3 (dark gray), and Phase 4 (black) for each condition (N, P, SL). The curves represent the mean values of 10 participants for each condition.

SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

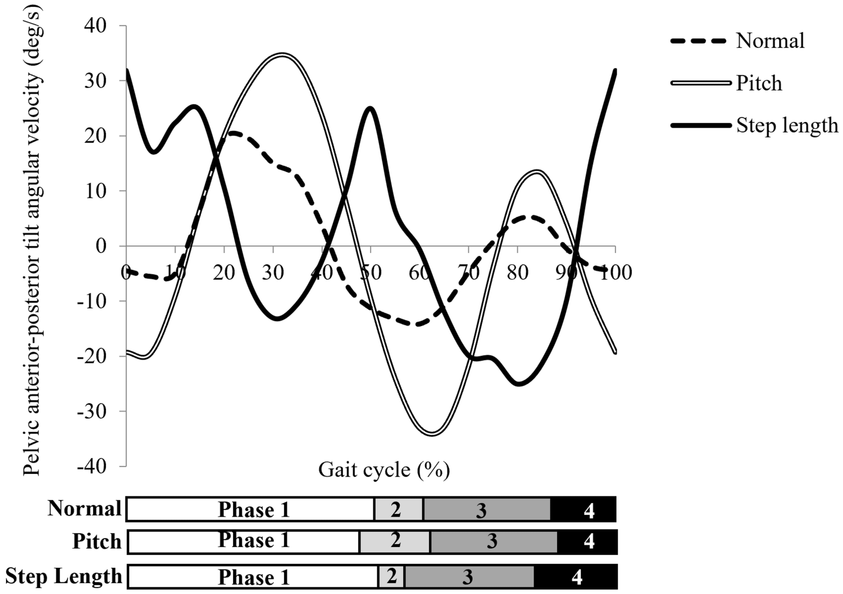

FIGURE 5: Time-series plots of pelvic anterior-posterior tilt angular velocity across the gait cycle for each walking condition.

The vertical axis represents the pelvic anterior-posterior tilt angular velocity (degrees/second), and the horizontal axis represents the gait cycle (%). Positive values indicate anterior tilt-directed velocity, while negative values represent posterior tilt-directed velocity. SL showed greater pelvic anterior-posterior tilt angular velocity changes during the late swing phase compared to N and P. The bars at the bottom of the figure represent the gait phases: Phase 1 (white), Phase 2 (light gray), Phase 3 (dark gray), and Phase 4 (black) for each condition (N, P, SL). The curves represent the mean values of 10 participants for each condition.

SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

Figure 6 illustrates hip flexion-extension angular acceleration throughout the gait cycle. In P, angular acceleration was directed toward flexion immediately after foot contact but soon shifted to extension. Flexion-directed acceleration reappeared around 30% of the gait cycle and peaked near 50%. Figure 7 shows the time-series profiles of iliopsoas activity. In P, activity decreased immediately after foot contact and remained low during early stance. Activity began to rise around 20% of the gait cycle and peaked near 30%, coinciding with the reappearance of flexion-directed angular acceleration observed in Figure 6.

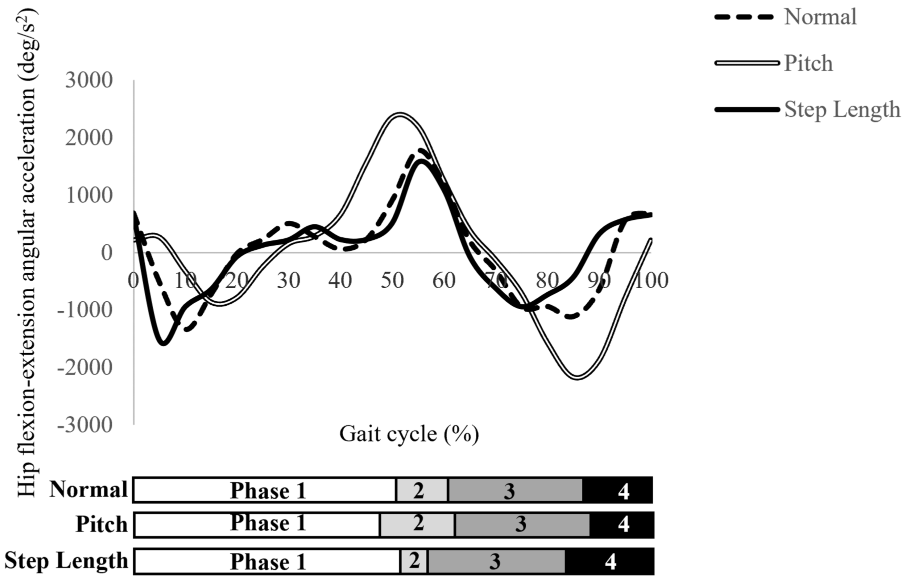

FIGURE 6: Time-series plots of hip flexion–extension angular acceleration across the gait cycle for each walking condition.

The vertical axis represents hip flexion–extension angular acceleration (degrees/second²), and the horizontal axis represents the gait cycle (%). Positive values indicate flexion-directed acceleration, while negative values represent extension-directed acceleration. In P, flexion acceleration was initially positive just after foot contact, then became negative during mid-stance, and turned positive again around 30% of the gait cycle, reaching a peak near 50%. The curves represent the mean values of 10 participants for each condition (N, P, SL). The bars at the bottom of the figure represent gait phases: Phase 1 (white), Phase 2 (light gray), Phase 3 (dark gray), and Phase 4 (black).

SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

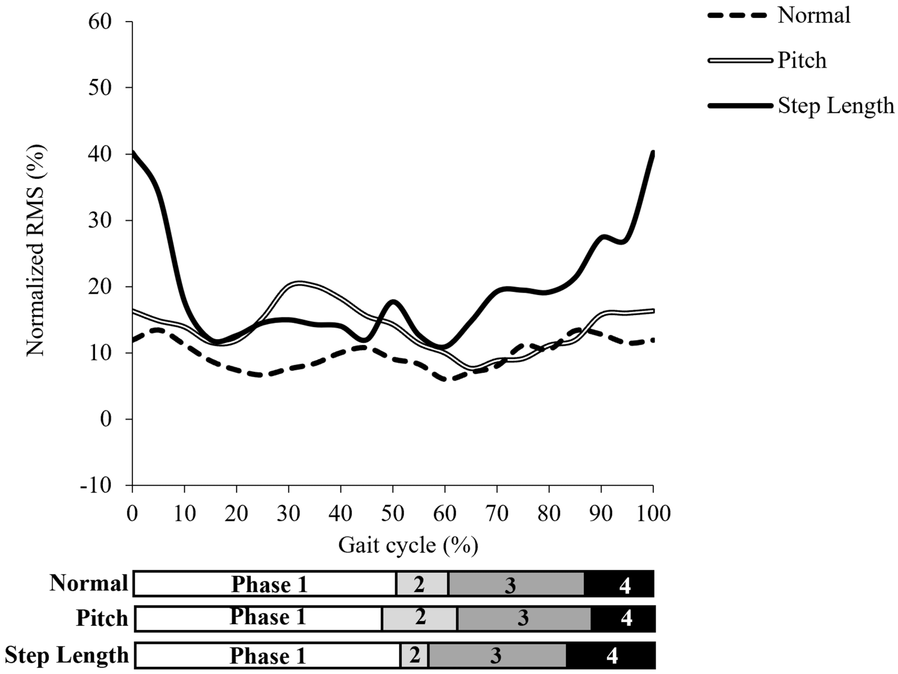

FIGURE 7: Time-series plots of iliopsoas muscle activity across the gait cycle for each walking condition.

The vertical axis represents normalized iliopsoas muscle activity (%), and the horizontal axis represents the gait cycle (%). In P, iliopsoas activity decreased immediately after foot contact and remained low during the earlier portion of stance. From around 20% of the gait cycle, activity began to increase and peaked near 30%, suggesting preparatory activation preceding the swing phase. Each curve represents the mean of 10 participants. The shaded bars at the bottom of the figure indicate gait phases: Phase 1 (white), Phase 2 (light gray), Phase 3 (dark gray), and Phase 4 (black) for each condition (N, P, SL).

SL: Step-Length Increased Walking; N: Normal Walking; P: Pitch Increased Walking.

Discussion

This study examined the effects of gait pattern modifications, specifically step length and cadence, on iliopsoas muscle activation and hip/pelvic kinematics during treadmill walking at a constant speed of 5 km/h. Three walking conditions were compared: SL (longer step length with lower cadence), N (self-selected step length and cadence), and P (shorter step length with higher cadence). These findings revealed distinct patterns of iliopsoas activation and joint kinematics associated with each gait strategy.

Iliopsoas activity was significantly higher during SL and P compared to N in Phases 1, 2, and 4, with no significant difference observed in Phase 3 (early swing). Notably, SL produced greater hip flexion and anterior pelvic tilt during the late swing phase, likely increasing the iliopsoas moment arm [18] and thereby enhancing its mechanical effectiveness. In Phase 4, the increased hip flexion angle and anterior pelvic tilt observed in the SL condition may have placed the iliopsoas in a more mechanically advantageous position. Greater hip flexion and anterior tilt during terminal swing may have enlarged the iliopsoas moment arm relative to other hip flexors, thereby improving the efficiency of torque production. This joint configuration may have contributed to the elevated iliopsoas activity during Phase 4. Therefore, the increased activation could be partly attributed not only to higher mechanical demands but also to potentially more favorable biomechanical conditions for force generation. This may explain the elevated muscle activation observed during Phase 4.

These results suggest that increasing step length, as seen in SL, may facilitate iliopsoas engagement during terminal swing, a phase often underemphasized in conventional gait training. Encouraging longer strides in individuals with adequate strength and flexibility may be beneficial in terms of muscle activation and dynamic limb preparation for stance.

In contrast, P appeared to increase iliopsoas activity primarily during the stance phase (Phases 1 and 2). Time-series analysis revealed that in P, iliopsoas activity began to rise around 20% of the gait cycle and peaked near 30%, preceding the peak of hip flexion angular acceleration at approximately 50% (Figures 6, 7).

In contrast, P appeared to increase iliopsoas activity primarily during the stance phase (Phases 1 and 2). Time-series analysis revealed that in P, iliopsoas activity began to rise around 20% of the gait cycle and peaked near 30%, preceding the peak of hip flexion angular acceleration at approximately 50% (Figures *6*, *7*). The timing of changes in Figures *6*, *7* may indicate that the increased iliopsoas activity during stance in P reflects a preparatory action for the upcoming swing phase. The rise in muscle activity starting around 20% of the gait cycle, followed by flexion-directed angular acceleration near 30%, may imply that the iliopsoas activates prior to observable joint motion. This may reflect a feedforward-like control strategy, where the muscle prepares in advance for the demands of rapid cadence. Such early activation might help ensure efficient swing initiation under time-constrained conditions imposed by the short step length and high stepping frequency.

The increased cadence in P likely imposed a demand for rapid leg swing initiation but without the accompanying increase in hip joint flexion observed in SL. The iliopsoas is considered a primary hip flexor, exhibiting angular velocity during flexion two to three times greater than that of the rectus femoris, particularly at lower flexion angles [19]. Thus, the role of the iliopsoas in P may be limited to limb acceleration rather than generating substantial hip flexion.

Effect sizes were also notable. Phase 4 showed a strong effect (Kendall's W = 0.91), indicating that SL has a substantial impact on iliopsoas activation during late swing. Phase 1 showed a moderate effect (W = 0.63), and Phase 2 showed a small-to-moderate effect (W = 0.49), reflecting a more modest influence of cadence on stance-phase activation. These findings support the notion that different gait parameters engage the iliopsoas at different points in the gait cycle and to varying degrees.

This pattern aligns with compensatory strategies often observed in older adults, who may increase cadence (as in P) to maintain walking speed despite limited joint mobility [5,7,8]. Although this compensation may preserve function in the short term, it could lead to excessive loading of the iliopsoas during stance. Over time, this may contribute to muscle fatigue, reduced hip mobility, and increased gait instability [4].

Early identification of these compensatory patterns may help physical therapists intervene before functional decline occurs. These findings highlight the need to assess both kinematic and neuromuscular aspects of gait in older adults.

Previous studies have shown that the iliopsoas, particularly the psoas major, is prone to age-related atrophy [20]. Therefore, interventions targeting iliopsoas strength, especially during the late swing phase, and improving hip flexion range of motion are essential for maintaining functional gait. Although SL may enhance hip and pelvic kinematics, it should be applied based on individual capacity. For example, in individuals with lumbar spine issues or poor flexibility, excessive anterior pelvic tilt could increase lumbar stress. Careful screening is needed before prescribing such gait modifications.

These findings emphasize the importance of individualized gait-training strategies. In older adults, cadence control should be prioritized. However, in those with sufficient muscle strength and flexibility, increasing step length as in SL may improve gait efficiency and help reduce fall risk. Beyond older adults, the insights from this study may also be relevant to athletes, post-operative patients, and individuals with neuromuscular conditions, where gait efficiency and motor control are crucial. Understanding phase-specific muscle engagement may help guide personalized gait retraining protocols in diverse clinical populations.

A key implication of this study may be the potential role of the iliopsoas as an active contributor not only during the stance-to-swing transition but also during late swing. The increased hip flexion angle and pelvic tilt observed in this phase suggest that the iliopsoas plays an important role in limb preparation for initial contact. This under-recognized function may inform more targeted, phase-specific rehabilitation strategies.

This study has several limitations. First, only healthy young men were included, which may limit generalizability to older adults. Additionally, the sample size was relatively small (n = 10). Although a priori power analysis indicated that this number was sufficient to detect moderate-to-strong effects in the measured parameters, a larger sample would increase the statistical robustness and generalizability of the results. Second, treadmill walking under fixed speed and controlled conditions may not reflect natural variability. Third, only the iliopsoas was analyzed, without evaluating other muscles such as the rectus femoris, gluteus medius, adductors, hamstrings, or triceps surae. Furthermore, ultrasound imaging was not used to evaluate muscle architecture or dynamic behavior during gait. Although ultrasound was used to guide electrode placement over the iliopsoas, real-time assessment of muscle-tendon dynamics was not conducted. This limits interpretation regarding structural contributions such as muscle stiffness, tendon recoil, or fascicle length changes. Fourth, kinetic data such as joint moments and ground reaction forces were not collected, limiting biomechanical interpretation. Fifth, although the time-series data provided important temporal insights, the lack of kinetic data prevents full understanding of mechanical demands on the iliopsoas throughout the gait cycle. Sixth, although the study focused on active muscle activation as measured by surface EMG, passive force contributions, such as those generated by muscle elongation and stiffness of the iliopsoas, were not assessed. These passive elements may also influence hip flexion during swing, and their contribution should be considered in future work. While stiffness may facilitate leg swing by storing and releasing elastic energy, it may also reduce joint range of motion, which could be a disadvantage in certain populations.

Future studies should include older adults, assess multiple lower-limb muscles, and collect kinetic data such as joint moments and ground reaction forces. In addition, incorporating markers of muscle fatigue and conducting longitudinal studies may help determine whether early, phase-specific interventions can slow functional decline in at-risk populations.

Conclusions

This study suggests that changes in step length and cadence may influence the timing and magnitude of iliopsoas muscle activation during walking. Specifically, longer steps (SL) may be associated with increased activity during the late swing phase, while shorter steps and higher cadence (P) may enhance activation during the stance phase.

Enhancing hip flexion mobility during walking could potentially contribute to safer and more efficient gait in older adults. The observed temporal relationship between muscle activation and joint acceleration may offer insight for tailoring training interventions to specific gait phases. These findings may inform the development of phase-specific strategies focusing on iliopsoas strengthening and mobility improvement.

Additional Information

Author Contributions

All authors have reviewed the final version to be published and agreed to be accountable for all aspects of the work.

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Disclosures

Human subjects: Consent for treatment and open access publication was obtained or waived by all participants in this study. Ethics Committee of Kanazawa Orthopedic Sports Medicine Clinic issued approval Kanazawa-OSMC-2022-008. **Animal subjects:** All authors have confirmed that this study did not involve animal subjects or tissue. **Conflicts of interest:** In compliance with the ICMJE uniform disclosure form, all authors declare the following: **Payment/services info:** All authors have declared that no financial support was received from any organization for the submitted work. **Financial relationships:** All authors have declared that they have no financial relationships at present or within the previous three years with any organizations that might have an interest in the submitted work. **Other relationships:** All authors have declared that there are no other relationships or activities that could appear to have influenced the submitted work.

Acknowledgements

The authors would like to thank all those who contributed to this study. During the preparation of this work, the authors used the generative AI tool/ChatGPT to assist with editing. After using this tool/service, the authors reviewed and edited the content as needed and takes full responsibility for the content of the published article.

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