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Lower-Extremity Intra-Joint Coordination and Its Variability between Fallers and Non-Fallers during Gait

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Abstract: Falling is one of the most common causes of hip fracture and death in older adults. A comparison of the biomechanics of the gait in fallers and non-fallers older adults, especially joint coordination and coordination variability, enables the understanding of mechanisms that underpin falling. Therefore, we compared lower-extremity intra-joint coordination and its variability between fallers and non-fallers older adults during gait. A total of 26 older adults, comprising 13 fallers, took part in this study. The participants walked barefoot at a self-selected speed on a 10-m walkway. Gait kinematics in the dominant leg during 10 cycles were captured with 10 motion tracking cameras at a sampling rate of 100 Hz. Spatiotemporal gait parameters, namely, cadence, walking speed, double support time, stride time, width, and length, as well as intra-joint coordination and coordination variability in the sagittal plane were compared between the two groups. Results showed that fallers walked with significant lower cadence, walking speed, and stride length but greater double support and stride time than non-fallers. Significant differences in the ankle-to-knee, knee-to-hip, and ankle-to-hip coordination patterns between fallers and non-fallers and less coordination variability in fallers compared to non-fallers in some instants of the gait cycles were observed. The differences in spatiotemporal gait parameters in fallers compared to non-fallers may indicate an adaptation resulting from decreased efficiency to decrease the risk of falling. Moreover, the differences in segment coordination and its variability may indicate an inconsistency in neuromuscular control. It may also indicate reduced ability to control the motion of the leg in preparation for foot contact with the ground and the knee and ankle motions during loading response. Finally, such differences may show less control in generating power during the push-off phase in fallers.

Keywords: elderly faller; non-faller; coordination; coordination variability; gait

1. Introduction

According to the World Health Organization, falling is one of major health challenges of old age, raising extensive discussions among gerontologists and physical therapists [1]. Yamada et al. [2] noted that about one-third of adults aged 65 years and 50% of adults older than 80 falls at least once a year, and 6% of such falls cause fractures. Falling among older adults can lead to severe consequences such as hip and wrist fractures, permanent disability, fear of falling, reduced quality of life, and even death [3,4]. The economic burden of falling among older adults is considerable, with annual direct medical costs about $754 million for the fatal falls, and about $50 billion is related to non-fatal fall [5]. However, researchers suggested that two-third of falls in older persons can be prevented [6]. To prevent falling among the elderly, identifying the biomechanical differences in gait patterns between fallers and non-fallers can be helpful. More specifically, the identification of lower extremity joints coupling and its variability during gait in fallers compared to non-fallers can help to develop rehabilitation exercises to reduce these differences and decrease the probability of falling induced injuries.

Gait disorder is the main reason for falls among older adults and not only it can occur during normal aging, but it can be due to neurological brain diseases or cervical spine and biomechanical diseases [7,8]. More than 70% of falls in older adults take place during gait [9]. For a normal gait, many systems are required, such as sensation and coordination, lower limb and core muscle strength, balance, and function in an integrated fashion [10,11]. A decrease in gait speed with an increase in age can be viewed as an adaptation to compensate for declining sensory-motor function in order to prevent the occurrence of falls. Biomechanical researchers have examined the spatiotemporal gait parameters and their variability among older adults as predictors of the risk of falling, as well as for differentiating between fallers and non-fallers [12,13]. Researchers suggested that the temporal variability and mean spatial parameters of gait have the highest efficacy in predicting the fall risk in clinical measurements [14,15], and that fallers walk with a considerably more irregular gait rhythm (shorter steps) than non-fallers [16]. Furthermore, increased variability of walking speed for defined and self-selected walking speeds and increased step width variability for fast walking speeds in fallers have been reported [12]. However, previous studies used linear measures of variability such as coefficient of variation in spatiotemporal gait parameters. Since the human movement is dynamic, the use of non-linear dynamic measures, such as coordination variability (CoordV), can provide additional insights into the interaction between two joints or segments [17–20].

Coordination can be viewed as a process in which movements components are sequentially organized over time, producing a functional and synergistic movement pattern [21,22]. The coordination between body segments and joints is coded in a subtle way to accommodate variations imposed by different task demands. Coordination patterns provide information about both the timing and the magnitude of movements, and they represent the organization of multiple degrees of freedom into a simple control strategy [23–26]. The CoordV measure quantifies the variety of movement patterns that an individual uses during a task and can provide a measure of the flexibility/adaptability of the individual's motor system [22,27]. A reduction in the CoordV measure could place older adults at a high risk of falling, because a decreased variety of movement patterns could limit the solutions to perturbations, e.g., obstacles or tripping [28]. As such, a greater variability in knee–ankle inter-joint coordination during the stance phase of gait in fallers than in non-fallers has been reported [29]. Moreover, different segment coordination and different CoordVs during treadmill gait between older adults and young adults have been reported [30]. However, these studies have not examined and compared the coordination and its variability in the joints between fallers and non-fallers [28,29]. To the best of our knowledge, only De Villa et al. [27] compared segments coordination and variability between fallers and non-fallers. They found significant differences in the coordination pattern of the thigh–leg segment pair, with the greatest differences observed at 80% and 120% of the preferred walking speed, among young adults, fallers, and non-fallers older adult, with emphasis on the older adults. Therefore, we believe that there is a lack of evidence about lower limb joint coupling between fallers and non-fallers older adults. Hence, the aim of the present study was to compare lower-extremity intra-joint coordination and its variability during gait between fallers and non-fallers older adults. We hypothesize that the values of spatiotemporal gait parameters are smaller in fallers than in non-fallers. Additionally, based in an interpretation of CoordV, we hypothesize that the ankle-to-knee, knee-to-hip, and hip-to-ankle coordination patterns would be different between fallers and non-fallers. Finally, we test whether coordination variability in the sagittal plane during gait would be smaller in fallers than in non-fallers.

2. Methods

2.1. Participants

A total of 26 healthy older adult fallers (n = 13, age: 72.53 ± 7.18 years, mass: 69.61 ± 9.57 kg, height: 166.45 ± 4.40 cm) and non-fallers (n = 13, age: 73.15 ± 7.15 years, mass: 68.53 ± 8.83 kg, height: 166.38 ± 3.40 cm) were recruited for this study. Fallers were defined as participants who had one fall in the past 12 months, whereas non-fallers were defined as participants who had not fallen in the past 12 months. Individuals who needed help for walking, had difficulties in understanding instructions, were receiving hospice care, or were not willing to participate were excluded. The purposes of the study were explained to the participants, who thereafter signed a consent form. All protocols were previously approved by the Kharazmi University Institutional Review Board.

2.2. Experimental Setup

Retroreflective markers were attached to the participants' body landmarks based on the lower-body Plug-in-Gait model. Data were captured at a frequency of 100 Hz using a 10-camera motion capture system (Motion Analysis, Rohnert Park, CA, USA). The calibration was performed according to the manufacturer's instructions. The participants stood in an anatomical position to record the static position. Thereafter, they were instructed to walk barefoot at a self-selected speed on a 10-m walkway, and kinematic data from 10 gait cycles of the dominant limb were collected at 2 m in the middle of the walkway.

2.3. Data Processing

Kinematic data were processed using Cortex software 8. Marker trajectories were low-pass filtered using a zero-lag fourth-order Butterworth filter with a cutoff frequency of 6 Hz. Heel strike and gait cycles were identified at the lowest point of the trajectory of the heel marker in the vertical direction (z-axis).

For coordination variability waveform analysis, because walking speed differed between the fallers and non-fallers, and we use a statistical tool that compares the entire time series; stance and swing were time normalized to 60 and 40 data points, respectively, and then merged to form a 100 data point gait cycle using a custom MATLAB code (MathWorks, Natick, MA, USA).

2.4. Data Analysis

The hip, knee, and ankle ranges of motion (ROMs) and spatiotemporal gait parameters (cadence, gait speed, stride time, and stride length) were calculated for the faller and non-faller groups.

Stride length was computed as the difference between two consecutives anterior–posterior positions of ipsilateral heel strikes, stride time was computed as the corresponding number of samples plus the inverse of the sampling frequency, walking speed was computed as the ratio between the total walking path and the total walking time, cadence was computed as the number of steps per minute, step width was computed as the medial–lateral distance between heel markers at consecutive heel strikes, double support time was computed as the time between ipsilateral heel strike and toe-off. To determine joints ROMs, the minimum value was subtracted from the maximum value for each gait cycle and then averaged across ten cycles.

2.5. Coordination and Coordination Variability Calculation

Intra-joint coordination and its variability were computed for five gait cycles for each participant, using the modified vector coding technique proposed by Needham et al. [30,31]. Vector coding consists in computing the coupling angle (γ) between different body segments or joints [32]. The ccoupling angle represents the joint coordination pattern, while the standard deviation of the coupling angle at each point of the gait cycle represents the CoordV [33]. Briefly, the coupling angles were calculated as the angle of a vector connecting consecutive data points in a phase space reconstructed using distal and proximal joint angles, according to the following Equations (1) and (2).

$$\gamma_i = \tan^{-1}\left(\frac{\theta_{D(i+1)} - \theta_{D(i)}}{\theta_{P(i+1)} - \theta_{P(i)}}\right)\cdot\frac{180}{\pi}\ \text{with}\theta_{P(i+1)} - \theta_{P(i)} > 0 \tag{1}$$

$$\gamma_i = \tan^{-1}\left(\frac{\theta_{D(i+1)} - \theta_{D(i)}}{\theta_{P(i+1)} - \theta_{P(i)}}\right)\cdot\frac{180}{\pi} + 180\ \text{with}\theta_{P(i+1)} - \theta_{P(i)} < 0 \tag{2}$$

where $0 \leq \gamma \leq 360^\circ$ are the coupling angle, i represents the consecutive samples in a normalized gait cycle, and γ_i is calculated on the basis of the distal joint angles θ_D and proximal joint angles θ_P. To avoid the coupling angle to assume indeterminate values, the following conditions were considered:

$$\gamma_i = \begin{cases} 90^\circ\ if\ \theta_{P(i+1)} - \theta_{P(i)} = 0\ and\ \theta_{D(i+1)} - \theta_{D(i)} > 0 \\ -90^\circ\ if\ \theta_{P(i+1)} - \theta_{P(i)} = 0\ and\ \theta_{D(i+1)} - \theta_{D(i)} < 0 \\ 180^\circ\ if\ \theta_{P(i+1)} - \theta_{P(i)} < 0\ and\ \theta_{D(i+1)} - \theta_{D(i)} = 0 \\ undefined\ if\ \theta_{P(i+1)} - \theta_{P(i)} = 0\ and\ \theta_{D(i+1)} - \theta_{D(i)} = 0 \end{cases} \tag{3}$$

The coupling angle γ_i was corrected according to Equation (4) to assume values between 0° and 360°:

$$\gamma_i = \begin{cases} \gamma_i + 360^\circ\ if\ \gamma_i < 0 \\ \gamma_i\ if\ \gamma_i \geq 0 \end{cases} \tag{4}$$

For an individual (n), and then for a group, γ_i was calculated from the horizontal ($\overline{x}$) and vertical ($\overline{y}$) components along multiple cycles of gait j, for each percentage i of the gait cycle according to Equations (5) and (6):

$$\overline{x_i} = \frac{1}{n}\sum_{j=1}^{n}(\cos\gamma_{ji}) \tag{5}$$

$$\overline{y_i} = \frac{1}{n}\sum_{j=1}^{n}(\text{sen}\gamma_{ji}). \tag{6}$$

Then, the length of the mean coupling vector was defined as Equation (7):

$$\overline{r_i} = \sqrt{\overline{x_i}^2 + \overline{y_i}^2} \tag{7}$$

Finally, the variability of the coupling angle $CoordV_i$, is calculated according to (8).

$$CoordV_i = \sqrt{2\cdot(1-\overline{r_i})}\cdot\frac{180^\circ}{\pi} \tag{8}$$

Coordination was classified into in-phase with proximal dominancy (IPPD), in-phase with distal dominancy (IPDD), anti-phase with proximal dominancy (APPD), and anti-phase with distal dominancy (APDD) [30]. The percentage of gait cycle for each coordination pattern was quantified using frequency plots in order to understand the most prevalent patterns. CoordV was calculated as the standard deviation of the vector connecting the corresponding consecutive time points of the angle–angle plots across all cycles. The coordination and its variability were examined for ankle–knee, knee–hip, and hip–ankle pairs in the sagittal plane.

2.6. Statistical Analysis

Joint ROMs, spatiotemporal gait parameters, and coordination pattern frequencies over gait cycles were assessed with an independent t-test in SPSS (IBM SPSS Statistics version 22; SPSS Inc., Chicago, IL, USA). A statistical parametric mapping (SPM) [34] independent t-test was used to detect significant differences in CoordV waveforms in all gait cycle between the two groups. The statistical significance level for all analyses was set at $\alpha \leq 0.05$. The SPM analyses were implemented using the open-source spm1d code in MATLAB (v.M0.1, www.spm1d.org accessed on 15 March 2021).

3. Results

3.1. ROM of Joints and Gait Spatiotemporal Parameters

Fallers presented significant lower cadence, gait speed, and stride length and significant greater double support and stride time than non-fallers. Moreover, although not significant, fallers presented a greater step width than non-fallers. No significant differences between fallers and non-fallers were found for ankle, knee, and hip ROMs (Table 1).

Table 1. Lower-extremity joint ranges of motion (ROMs) and spatiotemporal parameters of gait in fallers and non-fallers.

Parameters	Group		
	Non-Fallers	Fallers	*p*-Value
Cadence (step/min)	114.79 ± 12.3	97.59 ± 15.85	0.001
Gait speed (m/s)	1.04 ± 0.22	0.74 ± 0.15	0.000
Stride time (s)	1.05 ± 0.12	1.27 ± 0.33	0.008
Stride length (m)	1.08 ± 0.12	0.90 ± 0.09	0.000
Step width (cm)	9.89 ± 4.22	10.25 ± 4.33	0.823
Double support time (s)	0.26 ± 0.02	0.30 ± 0.02	0.000
Ankle ROM (°)	22.48 ± 2.23	25.02 ± 9.79	0.352
Knee ROM (°)	44.28 ± 11.50	47.19 ± 4.41	0.391
Hip ROM (°)	41.36 ± 6.37	41.80 ± 6.74	0.863

3.2. Intra-Joint Coordination Patterns

3.2.1. Ankle-To-Knee Coordination Pattern

The frequency of IPPD for the ankle plantarflexion and knee extension coordination pattern (p = 0.000) and that of APPD for ankle dorsiflexion and knee flexion coordination pattern (p = 0.000) were significantly lower in fallers than in non-fallers. By contrast, the frequency of APDD for the ankle plantarflexion and knee flexion coordination pattern (p = 0.037) was significantly greater in fallers than in non-fallers (Figure 1).

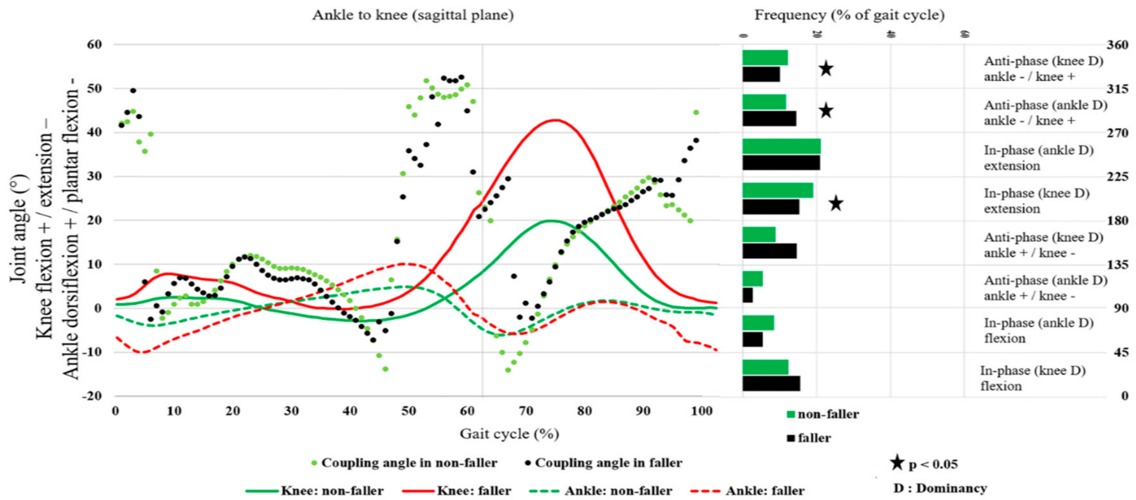

Figure 1. Ankle-to-knee angular displacement diagram (left axis), and frequency of coordination patterns (left axis) in the sagittal plane in fallers and non-fallers. The red and green solid lines are angular displacement of knee joint in fallers and non-fallers, respectively. The red and green dashed lines are angular displacement of ankle joint in fallers and non-fallers,

respectively. The black and green dots are the frequency scatter of coupling angle in fallers and non-fallers, respectively. The black and green bar charts are the frequency percentage of coupling angle in gait cycle in fallers and non-fallers, respectively.

3.2.2. Knee-To-Hip Coordination Pattern

The frequency of APPD for the hip extension and knee flexion coordination pattern was significantly lower in fallers than in non-fallers (p = 0.003) (Figure 2).

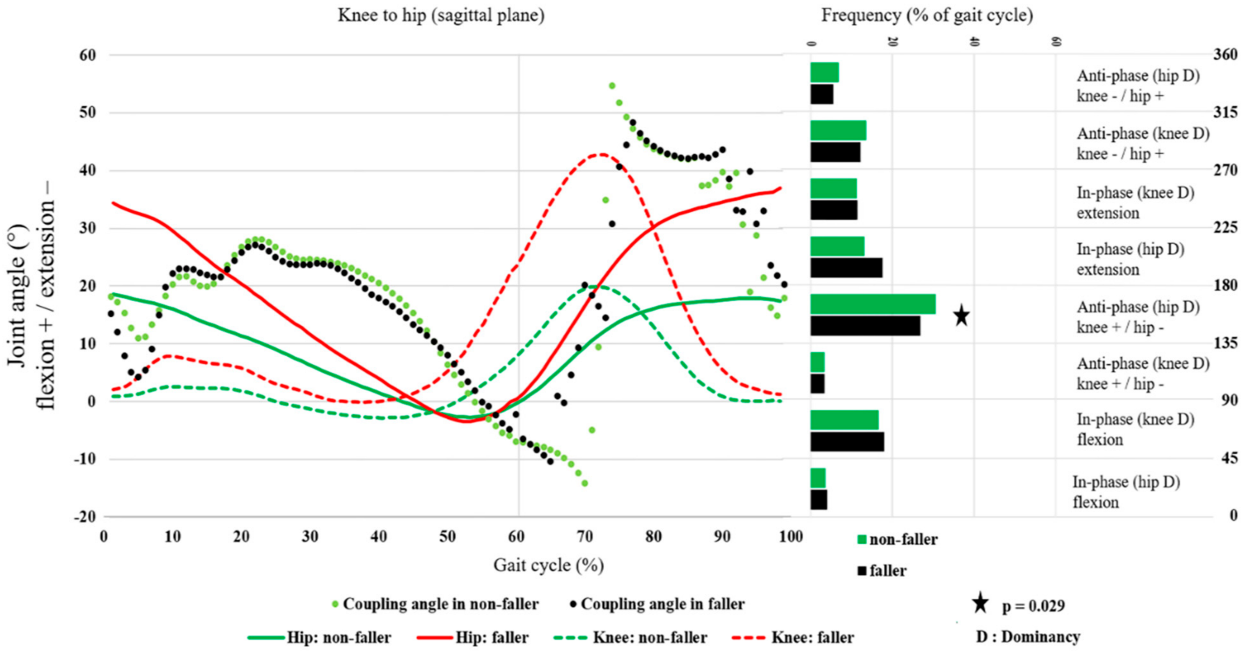

Figure 2. Knee-to-hip angular displacement diagram (left axis) and frequency of coordination patterns (left axis) in the sagittal plane in fallers and non-fallers. The red and green solid lines are angular displacement of knee joint in fallers and non-fallers, respectively. The red and green dashed lines are angular displacement of ankle joint in fallers and non-fallers, respectively. The black and green dots are the frequency scatter of coupling angle in fallers and non-fallers, respectively. The black and green bar charts are the frequency percentage of coupling angle in gait cycle in fallers and non-fallers, respectively.

3.2.3. Hip-To-Ankle Coordination Pattern

The frequency of IPPD for the hip flexion and ankle dorsiflexion coordination pattern (p = 0.003) and that of IPDD for the hip extension and ankle plantarflexion coordination pattern (p = 0.006) were significantly greater in fallers than in non-fallers (Figure 3).

3.3. Coordination Variability

The results of a vector analysis SPM independent t-test showed that the ankle-to-hip CoordV for the late swing phase (Figure 4a), ankle-to-knee CoordV for the late swing phase (Figure 4b), and knee-to-hip CoordV for the loading response phase (Figure 4c) were significantly lower in fallers than in non-fallers.

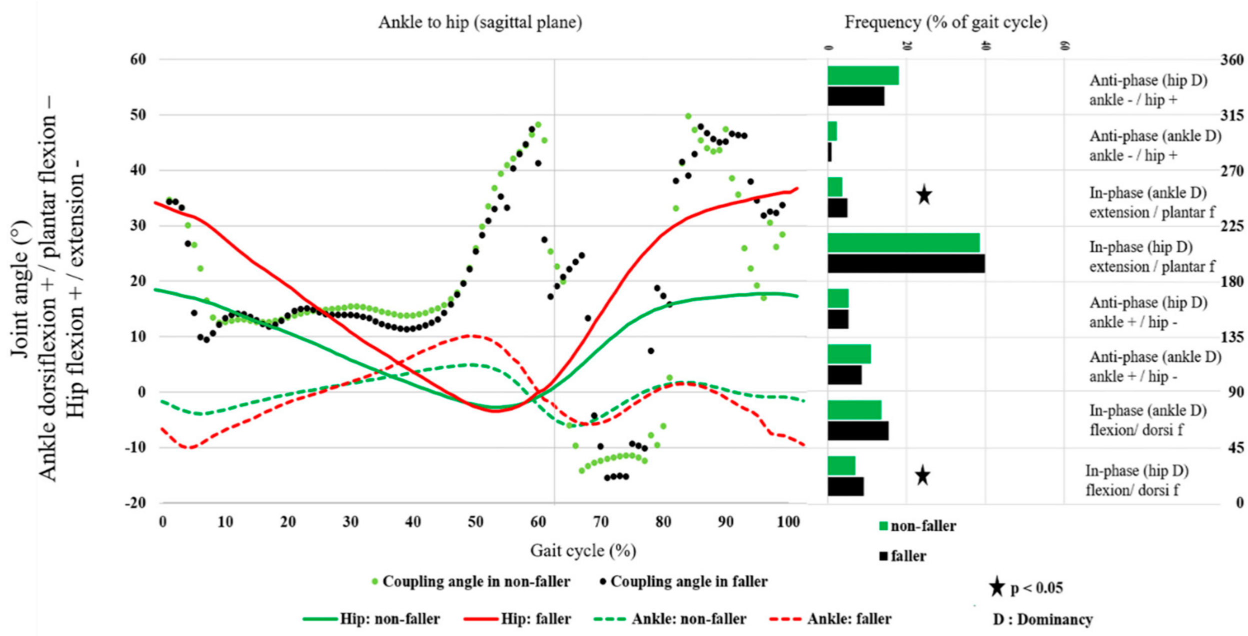

Figure 3. Ankle-to-hip angular displacement diagram (left axis), and frequency of coordination patterns (left axis) in the sagittal plane in fallers and non-fallers. The red and green solid lines are angular displacement of knee joint in fallers and non-fallers, respectively. The red and green dashed lines are angular displacement of ankle joint in fallers and non-fallers, respectively. The black and green dots are the frequency scatter of coupling angle in fallers and non-fallers, respectively. The black and green bar charts are the frequency percentage of coupling angle in gait cycle in fallers and non-fallers, respectively.

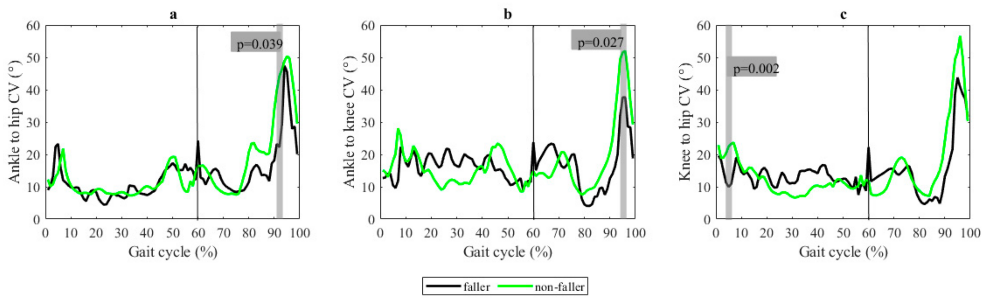

Figure 4. (**a**) Ankle-to-hip coordination variability (CoordV), (**b**) ankle-to-knee CoordV, and (**c**) knee-to-hip CoordV during gait cycles in fallers and non-fallers.

4. Discussion

The aim of the present study was to compare the lower-extremity intra-joint coordination and its variability in fallers and non-fallers older adults during gait. The results showed that there were no significant differences in the hip, knee, and ankle ROMs between the faller and non-faller groups. However, the spatiotemporal parameters were significantly different between the two groups, confirming our first hypothesis about the differences in spatiotemporal gait parameters between fallers and non-fallers. In our study, fallers walked with lower cadence, gait speed, and stride length but greater double support and stride time than non-fallers, indicating a cautious gait pattern adopted by fallers during walking. The decreased cadence and short stride length and greater double support and stride time seen in the faller group may be a consequence of decreased efficiency and an adaptation to decrease the risk of falling. The results of the present study were similar to those of a previous research [35], suggesting that a decrease in stride length, cadence, and gait speed and an increase in stride time are the most common parameters that can distinguish between fallers and non-fallers. Moreover, in 2018, Kwon reported that fallers had slower speed, shorter steps, and longer stance phase, with increased double-limb support, than non-fallers [16]. However, our results on spatiotemporal gait parameters were not in line with the results of the studies by Svoboda in 2017, Cebolla in 2015, and Mankino in 2017, which showed no significant differences in kinematic gait parameters between fallers and non-fallers [12,36,37].

The intra-joint coordination pattern of the lower extremity and its variability in the sagittal plane were significantly different between fallers and non-fallers in some instants of the gait cycle. This confirms, to some extent, our second and third hypotheses about significant differences in the intra-joint coordination pattern and its variability in the lower extremity between fallers and non-fallers. Fallers showed a more anti-phase coordination pattern in knee flexion-ankle plantar flexion with ankle dominancy (270–315°) compared to non-fallers; this is related to the greater knee flexion in all gait cycles and greater plantar flexion in loading response and late swing in fallers (Figure 1). The greater knee flexion in all gait cycles was probably a result of knee extensor weakness in older people with a history of falling. Knee extensors eccentrically control knee flexion during the stance phase of gait [38], and individuals with knee extensor weakness cannot efficiently extend their knee in the stance phase. Moreover, studies on differences between young and older adults showed that the contribution of the hips and ankles increased and decreased, respectively, with aging [39,40]. Fallers seem to have more ankle plantar flexion during late swing and initial contact to compensate more knee flexion and be more cautious about foot contact with ground.

Fallers showed a less in-phase coordination pattern in knee extension-ankle plantarflexion with knee dominancy (180–225°) in forward progression than non-fallers (Figure 1). Although, overall, fallers had more knee flexion in all gait cycle and less ankle plantarflexion than non-fallers during forward progression, this result suggests that fallers used more knee extension but less ankle plantarflexion for forward progression. Moreover, fallers showed less anti-phase coordination patterns in knee flexion-ankle plantar flexion with knee dominancy (315–360°) than non-fallers, which is related to the loading response and late swing. Fallers may have more knee flexion and more ankle plantarflexion during the late swing phase and initial contact, resulting in sooner foot contact with the ground and shorter contralateral single-limb support time. This pattern may help fallers in keeping their center of mass low and in walking with more caution. Moreover, as fallers have more plantarflexion during the late swing phase, they might use more knee flexion during the swing phase to increase foot clearance.

Our results showed that the CoordV in the ankle-to-knee pattern was significantly lower in fallers than in non-fallers in the late swing phase (Figure 4b), which is in agreement with the results of Chiu et al., who reported a lower variability of knee–ankle inter-joint coordination during the swing phase in older adults [28]. In our study, vector coding was used to quantify the coordination, and all phases of the gait cycle were analyzed. This lower CoordV in fallers could indicate an inconsistency in neuromuscular control and reduced ability to control the motion of the leg in preparation for foot contact. Overall, the results for the ankle-to-knee coordination pattern indicate that fallers walk with more ankle dorsiflexion and less plantarflexion pattern during forward progression than non-fallers, and that the knee of fallers is in greater flexion than that of non-fallers in all gait cycles. However, the less plantarflexion indicates that fallers cannot generate enough power to walk at the same cadence and gait speed as non-fallers.

Our results showed that fallers had a less anti-phase coordination pattern in hip extension–knee flexion with hip dominancy (135–180°) compared to non-fallers (Figure 2), which occurred in the loading response and push-off phases. Fallers showed more knee flexion and less hip extension in the loading response and push-off phase than non-fallers (Figure 2). Moreover, fallers also had more ankle plantarflexion during the loading response phase than non-fallers (Figure 1). These patterns of joint motion in fallers may be a strategy to land the leg on the ground with more ankle plantarflexion, knee flexion, and less hip extension to control the foot contact with the ground and the loading response. More ankle plantarflexion and less hip extension with more knee flexion cause less power generation during the push-off phase in fallers, which may indicate going to the single-leg support phase with caution. This result is in line with the results of the study by Barak, which showed smaller ankle plantarflexion and hip extension during push-off in fallers than in non-fallers [41]. Our results showed that fallers exhibit smaller CoordV in the knee-to-hip coordination during loading response than non-fallers (Figure 4c), which may indicate a reduction in degree of freedom in this phase. They may try to reduce their knee-to-hip CoordV due to more cautious loading response compared to non-fallers, at the expense of less ability to adapt to disturbances.

The results also showed that fallers had a more in-phase coordination pattern in hip flexion-ankle dorsiflexion with hip dominancy (0–45°) compared to non-fallers (Figure 3), which occurred in the initial and mid swing phases. This pattern indicates that fallers have more hip flexion and ankle dorsiflexion during the initial and mid swing phases, which may cause lower foot clearance and increased risk of falling. Moreover, fallers showed a more in-phase coordination pattern in hip extension-ankle plantarflexion with ankle dominancy (225–270°) compared to non-fallers, which occurred in the loading response phase (Figure 3). This pattern indicates that fallers have more ankle plantarflexion during the loading response phase, which may be a result of the greater plantarflexion at the late swing phase, and that fallers use a strategy of touching the ground faster than non-fallers, thus decreasing their stride length. The results showed that fallers have significantly less ankle-to-hip CoordV during the late swing phase than non-fallers (Figure 4a). This may indicate that fallers presented reduced ankle and hip degree of freedom to contact foot with ground with, decreasing their CoordV in late swing. This may indicate reduced ability to adapt to disturbances.

Finally, a number of study limitations need to be considered. First, we examined the intra-joint coordination pattern and its variability only in the sagittal plane; however, the intra-joint and intra-segment coordination pattern in other planes of motion also need to be considered. Second, the examination of coordination patterns and their variability together with electromyography and analysis of kinetic parameters such as joint torques and power flows may provide additional insights into the biomechanical differences of gait patterns between fallers and non-fallers. Third, we used a motion analysis system with reflexive markers in the laboratory setting for kinematic analysis that may affect the participants' natural gait pattern; however, long-term kinematic analysis with IMU-based systems out of a laboratory may give a more accurate results about gait pattern in fallers and non-fallers [42]. Finally, in this study, the non-fallers were defined as participants who had not fallen within the last 12 months; thus, the time domain for differentiating between fallers and non-fallers may be small.

5. Conclusions

This study showed that fallers walked with lower cadence, gait speed, and stride length but with greater double support and stride time than non-fallers, which decreased their efficiency and may be indicators of an increased risk of falling. Moreover, fallers walked with more knee flexion and ankle dorsiflexion, indicating a cautious walking strategy. Fallers also had less ankle plantarflexion and hip extension during forward progression, which suggests lower power generation for forward progression. Fallers showed less ankle-to-knee CoordV during late swing, less knee-to-hip CoordV during loading response, and less ankle-to-hip CoordV during late swing than non-fallers. This could indicate an inconsistency in neuromuscular control and a reduced ability to control the motion of the leg in preparation for foot contact with the ground and the knee and ankle motions during loading response, as well as less control to generate power during the push-off phase, thus reducing the efficiency, decreasing the foot clearance, reducing the ability to adapt to disturbances, and, hence, increasing the risk of fall during gait.

Author Contributions: Conceptualization, H.S., A.A., and S.S.S.; methodology H.S., A.A., and M.F.V.; software, A.A., and E.A.; validation, H.S., A.A., K.N., and M.F.V.; formal analysis, A.A., K.N., and E.A.; investigation, H.S., A.A., S.S.S., M.F.V., Z.S., and K.N.; resources, H.S., and A.A.; data curation, A.A., M.F.V., K.N., and H.S.; writing—original draft preparation, H.S., A.A., M.F.V., E.A., and K.N.;

writing—review and editing, A.A., M.F.V., K.N., and Z.S.; visualization, A.A., E.A., and M.F.V.; supervision and project administration, A.A. and S.S.S.; funding acquisition, S.S.S.; All authors have read and agreed to the published version of the manuscript.

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Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Ethics Committee of KHARAZMI UNIVERSITY (protocol code: IR.KHU.REC.1398.038 and date of approval: 12/01/2020).

Informed Consent Statement: Informed consent was obtained from all participants involved in the study.

Data Availability Statement: The data for this paper is not available online. They are available on request from the corresponding author.

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