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Lower-extremity inter-joint coordination variability in active individuals with transtibial amputation and healthy males during running

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Abstract

Background: It is challenging for people with transtibial amputation (TTA) to achieve coordination in locomotion due to biomechanical restrictions of the prostheses used. Even though the lower-extremity inter-joint coordination variability in individuals with TTA has been studied during gait, to the best of our knowledge, systematic analysis of the coordination patterns and coordination variability in multiple lower-limb joint couplings during running has never been done in people with this kind of amputation.

Purpose: To examine continuous inter-joint coordination patterns and coordination variability in running of highly active unilateral TTA patients using SACH prostheses and HIs.

Methods: Eight men with unilateral TTA (K-Level 4, SACH prosthesis) and eight normal men ran at a comfortable self-selected speed. The three-dimensional kinematic data were collected and coupling angles (hip-ankle, hip-knee, and knee-ankle in sagittal plane) were computed using the vector coding method and categorized in eight 45° coordination-angle bins. The pattern frequency is defined as the percentage of time in each bin for every stride and was statistically analyzed using a mixed design ANOVA and followed by Bonferroni-adjusted post hoc tests. The coordination variability (CV) was computed as the circular standard deviation of the coupling angle and was statistically analyzed using SPM1D.

Results: After Bonferroni adjustment, the prosthetic limb (PL) showed had a higher rate of occurrence than healthy individuals (HIs) for two hip-dominant anti-phase hip-to-ankle coordination patterns (hip extension-ankle dorsiflexion and hip flexion-ankle plantarflexion) and a higher rate of occurrence than both HIs and intact limb (IL) for one hip-dominant in-phase hip extension-ankle plantarflexion coordination pattern. In contrast, the PL had a lower rate of occurrence than HIs for in-phase ankle-dominant coordination patterns in the hip-to-ankle and knee-to-ankle coordination patterns. The IL showed no difference to HIs for any bin showing a

significant interaction effect. The CV of knee-to-ankle coordination was higher in the PL than that in the IL and HIs at approximately 56-59% of the running cycle while no differences were observed in the CV of hip-knee and hip-ankle coordination.

Conclusion: In this exploratory investigation of SACH foot users, it was found that the PL made use of more hip-dominant coordination strategies and fewer ankle-dominant coordination strategies compared to the HIs, perhaps as an adaptation to the inactive ankle joints. The increased knee-ankle CV observed in the PL might be due to the biomechanical restrictions of the SACH foot, inactivation of plantar flexion and/or sensory-motor control, but variability alone cannot determine the underlying cause. This research is carried out using only 8 participants per group.

Lower-limb coordination, an intrinsic feature of the spinal neural circuitry, refers to the ability of the joints and muscles to work together to produce goal-directed movements and is considered one of the fundamental components of human locomotion(1).Individuals with limb amputation face challenges in developing optimal and efficient coordination patterns, as the use of a prosthesis—an external factor—alters inter-limb coordination. Following amputation, individuals often become reliant on the prosthetic limb, leads to alterations in their movement patterns. as a result, they adopt distinct coordination patterns accompanied by increased joint loading compared to healthy individuals (HIs)(2). Individuals with transtibial (TTAs) lower limb amputation often experience secondary conditions, including musculoskeletal disorders such as low back pain, knee pain, and osteoarthritis(3)(4)(5). The anatomical loss of the plantar flexor muscles_ primarily the soleus and gastrocnemius which play a critical role in forward propulsion of the body during gait, leads to various neuromechanical impairments (6) .Sibley et al. reported that these individuals exhibit greater asymmetry compared to HIs, which is associated with a reduced stance phase on the prosthetic side, along with decreased peak vertical ground reaction force and step length on the intact side(7). Improper push-off movement leads to alterations in movement pattern and balance, which in turn causes individuals with amputation to change their strategies during activities such as running and walking(8). During walking at higher speeds, the ankle plays a fundamental role in movement control by generating approximately twice the power and energy compared to other joints(9). In HIs, an 80% reduction in plantar flexor muscle capacity leads to a 200% increase in the activity of other muscles(10). Collectively, these factors necessitate the adaptation and reorganization of coordination patterns in individuals with lower-limb amputation.

The neuromuscular mechanisms underlying human movement remain incompletely understood. During running, lower-limb joint coordination and stability play critical roles in shock absorption and injury prevention(11). As running speed increases, the neuromuscular system must adopt specific coordination adjustments to maintain stability while accommodating greater impact loads(12). Because the lower-limb joints are mechanically linked, the neuromuscular system continuously adapts inter-joint coordination to accommodate mechanical constraints and task demands(13)(14). The term coordination variability describes the inherent variation of the inter-joint coordination pattern during consecutive movement cycles and is thought to be part of motor learning and adaptation processes. (15). Coordination variability represents the natural variation in coordination patterns across repeated movement cycles, has been proposed to help individuals adapt to environmental and task constraints(16)(17). Reduced coordination variability has been found to indicate a more rigid coordination structure, which may lead to increased loading on tissues and thus result in injury due to overuse(18) . On the other hand, an increase in movement variability has been observed in those suffering from injuries(19). Hence, movement variability is considered most appropriate when it exists within a specific range, as both excessive and deficient variability might indicate poor control(20). For example, Hafer et al reported that increasing running cadence altered lower-limb coordination and coupling patterns while reducing coordination variability, which the authors suggested may relate to overuse injury risk(21).

Although the effects of running speed on lower-limb coordination have been well documented in HIs, the coordination responses of TTAs to the increased mechanical demands of running remain largely unknown. Though coordination variability at lower extremities' joints has been examined

during walking in those with transtibial amputation(22), it is unknown to us whether or not coordination patterns and coordination variability for multiple lower limb joints have been studied in detail during running in this patient population. Moreover, conventional biomechanical analyses have primarily relied on discrete kinematic and kinetic variables, such as joint angles and ground reaction forces (23). While valuable, these discrete metrics cannot fully characterize the continuous temporal structure of movement or the adaptability of neuromuscular control strategies adopted by individuals using a prosthesis. To overcome these limitations, the modified vector coding technique allows for the quantitative evaluation of continuous inter-joint coordination throughout the entire movement cycle(24)(2). This approach integrates the spatial and temporal relationships between two joints throughout movement and characterizes coordination patterns as in-phase, anti-phase, or out-of-phase while quantifying coordination variability as an indicator of neuromuscular adaptability.

Despite substantial advances in prosthetic technology, TTAs continue to experience considerable functional limitations, and prosthesis abandonment rates remain high (25–57%) due to poor adaptation to altered movement patterns and secondary complications (16).Improving our understanding of lower-limb coordination during high-demand tasks like running may contribute to the prevention of secondary complications, such as osteoarthritis, and facilitate the development of targeted rehabilitation programs, ultimately improving the quality of life in this population. Therefore, the aim of the present study was to investigate lower-limb coordination patterns and coordination variability during running in individuals with transtibial amputation compared with healthy individuals. We hypothesized that: (1) hip–knee, hip–ankle, and knee–ankle coordination patterns would differ between TTAs and HIs during running; and (2) coordination variability in the sagittal plane would also differ between the two groups, reflecting altered neuromuscular control in the prosthetic limb.

Participants

Eight adult males with unilateral transtibial amputation (TTAs) (age: 37.63 ± 8.55 years; height: 1.75 ± 0.07 m; body mass: 69.88 ± 7.66 kg) and eight healthy adult males (HIs) (age: 31.25 ± 2.76 years; height: 1.73 ± 0.05 m; body mass: 79.00 ± 6.75 kg) participated in this study. Because unilateral transtibial amputation is a relatively uncommon clinical population, and to ensure a homogeneous, highly active cohort capable of running, recruitment for the TTA group was strictly limited to members of the Tehran Province Amputee Football Team. Due to these highly specific inclusion criteria and the specialized athletic nature of the participants, the sample size was limited to eight individuals per group, and the current investigation was conducted as an exploratory study. All participants with TTA were classified as K-Level 4 according to the Medicare Functional Classification Level (MFCL) system(25), indicating the ability to perform high-impact activities. The inclusion criteria for the TTA group were: continuous use of the same Solid Ankle Cushion Heel (SACH) prosthesis (locally manufactured; hard socket with various suspension systems including sleeve and supracondylar/pin-lock) for at least six months prior to testing; daily use of this prosthesis outside of sports activities; absence of lower-limb pain; ability to walk and run independently without assistive devices; and no neurological or musculoskeletal disorders affecting locomotion. None of the participants had a congenital limb deficiency, and all were free

from additional musculoskeletal injuries at the time of testing. This study was performed following the Declaration of Helsinki. All participants signed an informed consent form, and the research protocol and details were approved by the Institutional Review Board of Kharazmi University (approval no. IR.KHU.REC.1403.067). All experiments were performed in accordance with relevant guidelines and regulations. Participant demographic characteristics are presented in Table 1.

Table 1: General demographic of participant

Participants		Age (years)	Side	Foot type	Mass (kg)	Height (m)	Independent t-test for Age		
							T	P-Value	Cohen's d
Amputee	Subject 1	28	Right	SACH	60	1.62	2.00	0.078	1.00
	Subject 2	32	Left	SACH	82	1.87			
	Subject 3	52	Right	SACH	65	1.72			
	Subject 4	27	Left	SACH	73	1.81			
	Subject 5	44	Right	SACH	72	1.72			
	Subject 6	36	Right	SACH	60	1.70			
	Subject 7	40	Right	SACH	72	1.74			
	Subject 8	42	Left	SACH	75	1.80			
	Mean (S.D) for amputees	37.63 (8.55)	-	-	69.88 (7.66)	1.75 (0.07)			
Healthy	Mean (S.D)	31.25 (2.76)	-	-	79 (6.75)	1.73 (0.05)			

Gait data collection (Experimental Setup)

The runway was covered with 10 VICON motion capture system (Oxford Metrics, Oxford, UK) cameras. Three-dimensional kinematic data were collected at 120 Hz, respectively while participants run along a straight, level runway. Twenty reflective markers were attached to the participant's body landmarks based on the lower-body Plug-in-Gait model. Surface markers were attached directly to the skin or prosthesis of amputees. For TTA participants, shank and foot markers on the prosthesis were approximated to match the locations of the corresponding markers on the intact side. The participants stood in an anatomical position to record the static position. Because the SACH device does not have an anatomical ankle joint that articulates, the 'ankle angle' for the prosthetic leg was operationally defined as the angle of a virtual joint. This was obtained

by calculating the relative orientation of the shank segment and the foot segment of the prosthetic leg in 3-D space using the marker trajectory data on the prosthetic device.

Before testing, all participants were familiarized with the experimental procedure and instructed on the proper running technique. They were then asked to run barefoot (participants with TTA completed the trials using their prosthesis without footwear) at a self-selected, comfortable running speed. Because running speed was not objectively measured using timing gates, participants were strictly instructed to maintain a consistent pace throughout the runway, and any trials with visually observable speed fluctuations were discarded. Each participant performed ten running trials. The six highest-quality trials—defined as having complete marker trajectories, no technical errors, and compliance with the experimental protocol—were selected for subsequent data processing.

Data processing

Markers trajectories were labeled and gap filled using Nexus 2.2.3 software (Vicon, Oxford, UK). Thereafter parameters of gait were extracted using ProCalc 1.1 (Vicon, Oxford, UK) gait analysis software. To remove high-frequency noise, kinematic data were filtered using a zero-lag, fourth-order Butterworth low-pass filter with a cutoff frequency of 6 Hz(26). Kinematics data were re-sampled in to 100 data point using a custom MATLAB code (Version R2018b, MathWorks, Natick, MA, USA).

Data Analysis

Data for the TTAs for PL and IL were calculated separately. For the control group (HIs), data from the right and left limbs were also processed separately. In the between-group statistical comparisons, the PL of the TTA group was compared with the right limb of the HI group, and the IL of the TTA group was compared with the left limb of the HI group Lower extremity joints range of motion (ROM), coordination and coordination variability were calculated and compared between healthy and amputees. Based on preliminary paired samples SPM{t} analysis, there was no significant difference in either kinematics or coordination between the right and left limbs of the control participants. For this reason, in order to ensure consistency in terms of bilateral comparisons, the right and left limbs of the HIs were systematically paired with the PL and IL of the TTAs, respectively. Inter-joint coordination and its variability for hip-knee, hip-ankle, and knee-ankle in sagittal plane were calculated using modified vector coding technique (24,27). To incorporate the directionality property of angular measurements, the continuous trajectories of coupling angles from the six chosen trials per participant were averaged together through circular statistics. The average coupling angle at each normalized time instant was obtained based on the calculation of the directional mean of sine and cosine values. This average trajectory of each participant was further employed to classify the phase and for statistical analysis purposes. In addition, inter-trial coordination variability (CV) was mathematically computed by taking the circular standard deviation of the coupling angles at each time instant in all six trials using the formula, $CV = \sqrt{2(1-R)} \times \left(\frac{180}{\pi}\right)$where R represents the mean resultant vector length.

Statistical Analysis

Statistical analyses were performed using IBM SPSS Statistics (Version 27.0; IBM Corp., Armonk, NY, USA). The continuous coupling-angle trajectory for each joint coupling (hip–ankle, hip–knee, and knee–ankle) was classified into eight 45° coordination-angle bins representing distinct inter-joint coordination patterns. To compare the occurrence of these patterns, coordination pattern frequency was calculated as the percentage of frames within each 45° bin relative to the total number of frames in the analyzed running cycle, averaged across the selected trials. For each coordination-angle bin and joint coupling, a two-way mixed-design analysis of variance (general linear model with repeated measures) was performed, with group HIs vs TTAs as the between-subject factor and limb (right vs. left for the HIs; prosthetic vs. intact for the TTAs) as the within-subject factor. Because the within-subject factor consisted of only two levels, the assumption of sphericity was inherently satisfied. Normality (Shapiro–Wilk) and homogeneity of variances (Levene's test) were examined; ANOVA was retained because of equal group sizes (n = 8 per group), and Welch's t-tests were used for between-group comparisons. Statistical significance was set at $p < 0.05$ for all analyses. Effect sizes for the GLM analyses were reported as partial eta squared (ηp^2). When a significant group × limb interaction was detected, simple-effects analyses were performed using paired-samples t-tests within each group and Welch's independent-samples t-tests between groups (specifically comparing the PL to the Right limb, and the IL to the Left limb). A Because four post hoc comparisons were made within each bin, p-values were Bonferroni-adjusted (adjusted p = raw p × 4); adjusted p-values are denoted p_adj, and raw p-values are additionally given where a difference was significant only before correction. Effect sizes for post hoc comparisons were calculated as Cohen's d (d_z for paired comparisons).

Coordination variability was analyzed using one-dimensional Statistical Parametric Mapping (SPM1D; Pataky, 2012). Independent-samples SPM{t} tests were used for between-group comparisons (control vs. TTA), whereas paired-samples SPM{t} tests were used for within-group comparisons (right vs. left limbs in the control group and prosthetic vs. intact limbs in the TTA group). All analyses were performed across the entire time-normalized running cycle (0–100%) using a two-tailed significance level of $\alpha = 0.05$. Statistical significance was determined using Random Field Theory (RFT), and significant clusters were identified when the SPM{t} trajectory exceeded the critical threshold.

Results

Hip-to-Ankle Coordination Pattern

Mixed ANOVA revealed significant group × limb interactions in bins 0–45°, 135–180°, 180–225°, 225–270°, and 315–360° ($\eta p^2 = .28–.42$), and significant main effects of group ($F = 5.24$, $p = .038$) and limb ($F = 13.60$, $p = .002$) in bin 90–135° (Table 2); as the interaction in this bin was not significant ($p = .192$), no pairwise comparisons were performed. The frequency of the hip-dominant, in-phase, hip flexion–ankle dorsiflexion coordination pattern was significantly lower in the PL than in the HIs (PL vs. Right: $t = 5.50$, $p_{adj} < .001$, $d = 2.75$; Figure 1). Additionally, the frequency of the ankle-dominant, in-phase, hip extension–ankle plantarflexion coordination

pattern was significantly lower in the PL than in the HIs (PL vs. Right: t = 4.94, p_adj = .001, d = 2.47); the difference from the IL (PL vs. IL: t = −2.97, d = −1.05) was significant before correction but did not survive it (p = .021, p_adj = .084; Figure 1). In contrast, the frequency of the hip-dominant anti-phase hip extension–ankle dorsiflexion coordination pattern was significantly higher in the PL than in the HIs (PL vs. Right: t = −6.01, p_adj < .001, d = −3.00), whereas the difference from the IL (PL vs. IL: t = 3.07, d = 1.08) did not survive correction (p = .018, p_adj = .072; Figure 1). Similarly, the frequency of the hip-dominant, in-phase, hip extension–ankle plantarflexion coordination pattern was significantly higher in the PL than in both the IL (PL vs. IL: t = 3.72, p_adj = .028, d = 1.31) and the HIs (PL vs. Right: t = −3.30, p_adj = .021, d = −1.65; Figure 1). In addition, the frequency of the hip-dominant, anti-phase, hip flexion–ankle plantarflexion coordination pattern was significantly higher in the PL than in the HIs (PL vs. Right: t = −4.59, p_adj = .002, d = −2.30), whereas the difference from the IL (PL vs. IL: t = 2.57, d = 0.91) did not survive correction (p = .037, p_adj = .148; Figure 1). No differences between IL and left limb of HIs in any bin were found. (Post hoc p-values are Bonferroni-adjusted (p_adj)).

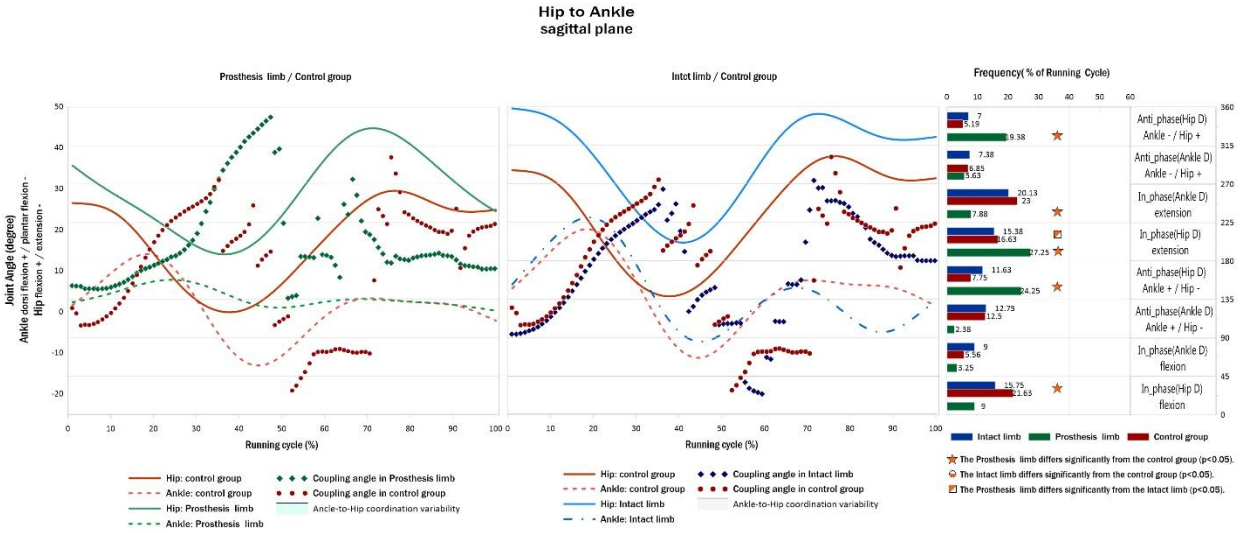


Figure 1. Hip-to-Ankle angular displacement diagram (left axis) and frequency of coordination patterns (right axis) in the sagittal plane in TTAs and control group. The 360° coupling-angle space was classified into eight 45° bins, representing distinct inter-joint coordination patterns.

			Hip To Ankle																	
			General linear model								Paired sample t-test						Independent t-test					
			Group			Limb			Group*Limb								R*P			L*I		
Phase	Limb	Mean (SD)	F	P-Value	ηp^2	F	P-Value	ηp^2	F	P-Value	ηp^2	T	Bonferroni-adjusted p-value	Cohen's d_z	T	Bonferroni-adjusted p-value	Cohen's d_z	T	Bonferroni-adjusted p-value	Cohen's d_z	T	Bonferroni-adjusted p-value	Cohen's d_z
0-45	Left	20.13(5.38)	36.64	0.001	0.72	0.85	0.371	0.05	5.76	0.031	0.29	1.71	0.524	0.60	1.84	0.432	0.65	5.5	0.001	2.75	1.73	0.402	0.87
	Right	23.13(2.64)																					
	Intact	15.75(4.65)																					
	Prosthesis	9(6.74)																					
45-90	Left	6.37(5.44)	0.09	0.766	0.07	1.93	0.186	0.12	0.60	0.450	0.04												
	Right	4.75(3.77)																					
	Intact	9(7.83)																					
	Prosthesis	3.25(2.11)																					
90-135	Left	14.87(8.23)	5.24	0.038	0.273	13.60	0.002	0.49	1.88	0.192	0.11												
	Right	10.13(3.44)																					
	Intact	12.75(6.60)																					
	Prosthesis	2.38(1.01)																					
135-180	Left	8.38(3.29)	18.09	0.001	0.56	6.866	0.020	0.32	10.21	.006	0.04	-0.89	1.000	-0.31	3.07	0.072	1.08	-6.01	0.001	-3.00	-0.90	1.000	-0.45
	Right	7.13(1.64)																					
	Intact	11.63(9.56)																					
	Prosthesis	24.25(7.88)																					
180-225	Left	15.88(2.50)	2.81	0.115	0.16	10.97	0.005	0.43	6.60	0.022	0.32	0.60	1.000	0.21	3.72	0.028	1.31	-3.30	0.021	-1.65	0.13	1.000	0.06
	Right	17.38(2.50)																					
	Intact	15.38(7.9)																					
	Prosthesis	27.25(8.06)																					
225-270	Left	22.63(9.14)	8.52	0.011	0.37	4.24	0.058	0.23	5.43	0.035	0.27	0.19	1.000	0.07	-2.97	0.084	-1.05	4.94	0.001	2.47	0.50	1.000	0.25
	Right	23.38(5.42)																					
	Intact	20.13(10.68)																					
	Prosthesis	7.87(7.01)																					
270-315	Left	6(3.46)	0.06	0.808	0.004	0.00	1.000	0.00	1.77	0.204	0.11												
	Right	7.75(1.28)																					
	Intact	7.37(6.07)																					
	Prosthesis	5.63(3.73)																					
315-360	Left	4.88(2.03)	28.23	0.001	0.66	7.03	0.019	0.33	5.74	0.031	0.29	0.63	1.000	0.22	2.57	0.148	0.91	-4.59	0.002	-2.30	-0.77	1.000	-0.39
	Right	5.50(2.26)																					
	Intact	7(5.44)																					
	Prosthesis	19.38(8.22)																					

Table 2. Ankle-to-hip angular coupling (mean ± SD) and between/within-group statistical comparisons across the running cycle in TTAs and control group participants (sagittal plane).

Hip-to-Knee Coordination Pattern

Mixed ANOVA revealed a significant main effect of group in bin 90–135° ($F = 12.87$, $p = .003$, $\eta p^2 = .48$) and significant group × limb interactions in bins 180–225° ($F = 6.52$, $p = .023$) and 270–315° ($F = 4.69$, $p = .048$; Table 3). In bin 90–135°, because the interaction was not significant ($p = .435$), no pairwise comparisons were performed; descriptively, the knee-dominant, anti-phase hip extension–knee flexion pattern was least frequent in the PL (11.88%) compared with the IL

(19.13%) and the right and left limbs of the HIs (21.63% and 24.75%).The frequency of the hip-dominant, in-phase hip extension–knee extension coordination pattern was higher in the PL than in the IL (PL vs. IL: t = 2.52, d = 0.89); this difference was significant before correction but did not survive it (p = .040, p_adj = .159; Figure 2). In bin 270–315°, the interaction was significant, but none of the pairwise comparisons reached significance after correction. In bins with a significant interaction, the IL did not differ from the left limb of the HIs. (Post hoc p-values are Bonferroni-adjusted (p_adj)).

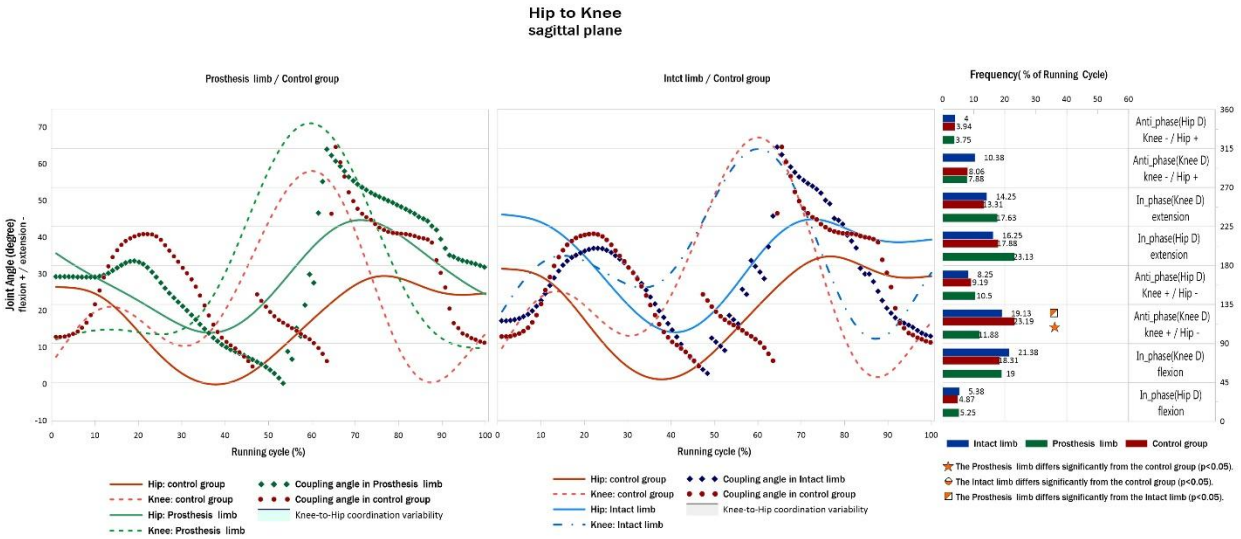


Figure 2. Hip-to-Knee angular displacement diagram (left axis) and frequency of coordination patterns (right axis) in the sagittal plane in TTAs and control group. The 360° coupling-angle space was classified into eight 45° bins, representing distinct inter-joint coordination patterns.

		Hip to Knee																					
		Mean (SD)	General linear model									Paired sample t-test						Independent t-test					
			Group			Limb			Group*Limb			control			TTAs			R*P			L*I		
			F	P-Value	ηp^2	F	P-Value	ηp^2	F	P-Value	ηp^2	T	Bonferroni-adjusted p-value	Cohen's d	T	Bonferroni-adjusted p-value	Cohen's d	T	Bonferroni-adjusted p-value	Cohen's d	T	Bonferroni-adjusted p-value	Cohen's d
0-45	Left	4.75(1.488)	0.48	0.500	0.033	0.01	0.916	0.001	0.10	0.753	0.007												
	Right	5(1.19)																					
	Intact	5.37(2.56)																					
	Prosthesis	5.25(1.28)																					
45-90	Left	18.50(4.81)	1.00	0.333	0.067	0.29	0.599	0.02	0.153	0.701	0.01												
	Right	18.13(5.11)																					
	Intact	21.38(8.91)																					
	Prosthesis	19.00(5.60)																					
90-135	Left	24.75(10.76)	12.87	0.003	0.479	4.08	0.063	0.226	0.64	0.435	0.044												
	Right	21.63(4.24)																					
	Intact	19.13(5.64)																					
	Prosthesis	11.88(3.60)																					
135-180	Left	8.75(2.60)	0.01	0.907	0.001	0.82	0.380	0.05	0.15	0.69	0.01												
	Right	9.63(2.72)																					
	Intact	8.25(2.81)																					
	Prosthesis	10.50(8.05)																					
180-225	Left	18.38(2.61)	0.64	0.434	0.04	4.87	0.044	0.25	6.52	0.023	0.31	-0.53	1.000	-0.18	2.52	0.159	0.89	-1.86	0.409	-0.93	1.09	1.000	0.54
	Right	17.88(1.45)																					
	Intact	16.25(4.80)																					
	Prosthesis	23.13(7.84)																					
225-270	Left	12.63(3.37)	2.43	0.14	0.14	1.32	0.270	0.08	0.23	0.633	0.01												
	Right	14.00(1.19)																					
	Intact	14.25(8.87)																					
	Prosthesis	17.63(4.71)																					
270-315	Left	7.50(1.60)	1.00	0.334	0.067	0.67	0.425	0.04	4.69	0.048	0.25	1.46	0.742	0.51	-1.68	0.547	-0.59	0.821	1.000	0.41	-1.71	0.488	-0.85
	Right	8.63(1.30)																					
	Intact	10.38(4.47)																					
	Prosthesis	7.88(2.23)																					
315-360	Left	3.75(0.70)	0.043	0.839	0.003	0.037	0.849	0.003	0.93	0.350	0.06												
	Right	4.13(0.35)																					
	Intact	4.00(1.07)																					
	Prosthesis	3.75(1.17)																					

Table 3. Knee-to-hip angular coupling (mean ± SD) and between/within-group statistical comparisons across the running cycle in TTAs and control group participants (sagittal plane).

Knee-to-Ankle Coordination Pattern

Mixed ANOVA revealed significant group × limb interactions in bins 225–270° (F = 6.48, p = .023) and 270–315° (F = 5.53, p = .034), and a large main effect of group in bin 180–225° (F = 17.74, p = .001, ηp^2 = .56; Table 4). The frequency of the ankle-dominant, in-phase knee extension–ankle plantarflexion coordination pattern was significantly lower in the PL than in the HIs (PL vs. Right: t = 3.81, p_adj = .015, d = 1.90); the difference from the IL (PL vs. IL: t = −2.99, d = −1.06) was significant before correction but did not survive it (p = .020, p_adj = .080; Figure 3). Furthermore, the frequency of the ankle-dominant, anti-phase knee flexion–ankle plantarflexion coordination pattern was significantly lower in the PL than in the HIs (PL vs. Right: t = 4.33, p_adj = .004, d = 2.17; Figure 3). No significant effects were found in bins 45–90° and 315–360°. In bin 180–225°, because the interaction was not significant (p = .391), no pairwise

comparisons were performed; descriptively, the knee-dominant, in-phase knee extension–ankle plantarflexion pattern was most frequent in the PL (27.00%) compared with the IL (17.75%) and the right and left limbs of the HIs (18.38% and 14.13%). In bins with a significant interaction, the IL did not differ from the left limb of the HIs. (Post hoc p-values are Bonferroni-adjusted (p_adj)).

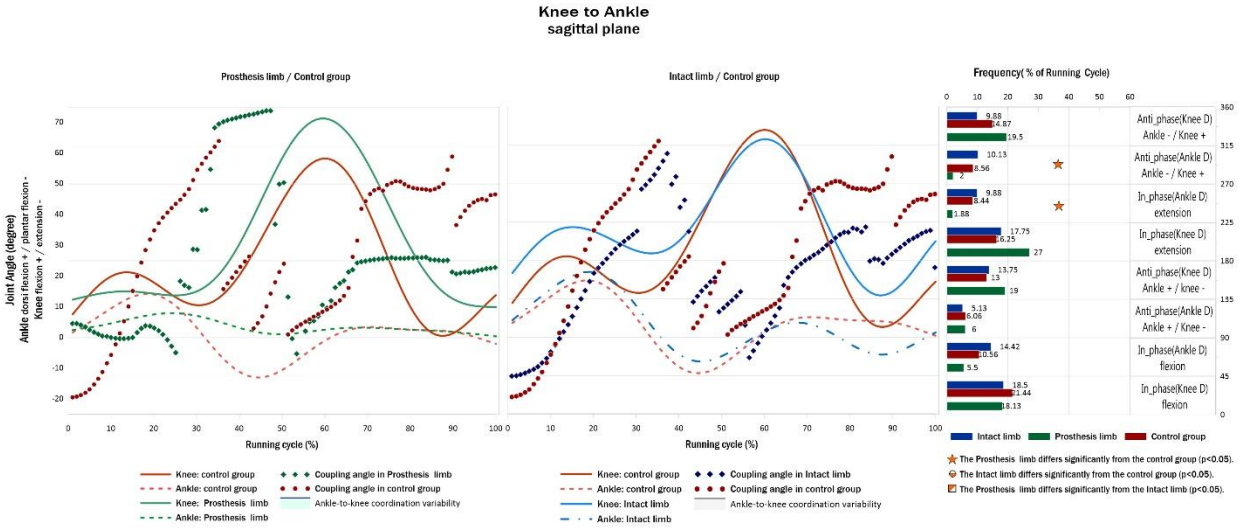


Figure 3. Knee-to-Ankle angular displacement diagram (left axis) and frequency of coordination patterns (right axis) in the sagittal plane in TTAs and control group. The 360° coupling-angle space was classified into eight 45° bins, representing distinct inter-joint coordination patterns.

		Knee to Ankle																			
		Mean (SD)	General linear model									Paired sample t-test						Independent t-test					
			Group			Limb			Group*Limb			control			TTAs			R*P			L*I		
			F	P-Value	ηp^2	F	P-Value	ηp^2	F	P-Value	ηp^2	T	Bonferroni-adjusted p-value	Cohen's d	T	Bonferroni-adjusted p-value	Cohen's d	T	Bonferroni-adjusted p-value	Cohen's d	T	Bonferroni-adjusted p-value	Cohen's d
0-45	Left	23.75(11.24)	0.81	0.38	0.05	0.88	0.364	0.05	0.635	0.43	0.04												
	Right	19.13(8.02)																					
	Intact	18.50(8.12)																					
	Prosthesis	18.13(7.01)																					
45-90	Left	10.12 (3.18)	0.19	6.69	0.01	2.28	0.153	0.14	3.44	0.085	0.20												
	Right	11.00 (4.81)																					
	Intact	14.00 (10.42)																					
	Prosthesis	5.50 (3.96)																					
90-135	Left	6.38 (1.41)	0.26	0.617	0.02	0.03	0.856	0.002	1.23	0.287	0.080												
	Right	5.75 (1.39)																					
	Intact	5.13 (1.46)																					
	Prosthesis	6.00 (4.07)																					
135-180	Left	12.25(5.44)	3.09	0.100	0.18	2.52	0.134	0.15	0.78	0.392	0.05												
	Right	13.75(3.19)																					
	Intact	13.75(4.95)																					
	Prosthesis	19.00(8.15)																					
180-225	Left	14.13(4.70)	17.74	0.001	0.55	5.72	0.031	0.29	0.78	0.391	0.053												
	Right	18.38(3.29)																					
	Intact	17.75(6.98)																					
	Prosthesis	27.00(8.91)																					
225-270	Left	8.88(2.69)	4.73	0.047	0.25	10.05	0.007	0.41	6.47	0.023	0.31	-1.05	1.000	-0.37	-2.99	0.080	-1.05	3.80	0.015	1.90	-0.49	1.000	-0.24
	Right	8.00(1.85)																					
	Intact	9.88(5.05)																					
	Prosthesis	1.88(1.15)																					
270-315	Left	7.00(3.54)	1.73	0.209	0.11	1.09	0.314	0.07	5.53	0.034	0.28	2.40	0. 188	0.84	-1.76	0.484	-0.62	4.33	0.004	2.16	-0.80	1.000	-0.40
	Right	10.13(2.69)																					
	Intact	10.12(9.41)																					
	Prosthesis	2.00(1.02)																					
315-360	Left	16.50(13.41)	0.005	1.000	0.00	0.75	0.439	0.04	3.68	0.076	0.21												
	Right	12.87(4.22)																					
	Intact	9.88(7.01)																					
	Prosthesis	19.5(6.48)																					

Table 4. knee-to-Ankle angular coupling (mean ± SD) and between/within-group statistical comparisons across the running cycle in TTAs and control group participants (sagittal plane).

Coordination variability

The results of a Independent-samples SPM{t} tests were used for between-group comparisons (control vs. TTA) showed that the knee-to-ankle coordination variability for the initial swing phase (Figure 4) was significantly greater for the PL in the TTAs compared to the IL and the HIs. However, no significant differences between TTAs and HIs were found for Hip-to-Ankle and Knee-to-Hip coordination variability (Figure 4).

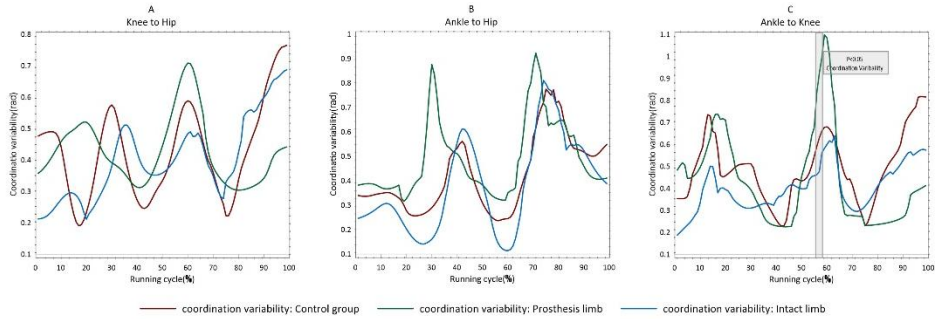


Figure 4. (A) knee-to-Hip coordination variability, (B) Ankle-to-Hip coordination variability, and (C) Ankle-to-Knee coordination variability.

Discussion

The primary aim of the present study was to compare lower-limb joint coordination patterns and coordination variability during running between individuals with transtibial amputation (TTAs) and healthy individuals (HIs). The present findings demonstrated significant differences in lower-limb coordination patterns and coordination variability during specific phases of the running cycle, thereby supporting our first hypothesis and, for the knee–ankle coupling only, our second hypothesis. No difference was found between intact limb (IL) and HIs in any bin where interaction was significant. During mid-stance (15–25% of the running cycle), the prosthetic limb (PL) exhibited a significantly higher frequency of a hip-dominant anti-phase hip extension–ankle dorsiflexion coordination pattern than the HIs. This finding likely reflects the absence of active ankle function in the PL, causing ankle motion to rely primarily on the passive mechanical properties of the prosthesis. Consequently, the hip may to assume a more dominant role in advancing the limb as a compensatory strategy to maintain forward progression and dynamic stability. By analogy, Błażkiewicz et al. demonstrated that in individuals with impaired ankle function, such as those with foot drop, the reduced force-generating capacity of the ankle can be compensated for by an increased reliance on adjacent joints, particularly the hip (28). They reported substantial altered movement strategies at the hip, which contribute to knee flexion and hip extension control. These findings suggest that when ankle function is compromised, an altered coordination strategy involving proximal joints, particularly the hip, may serve as an adaptive mechanism to maintain limb support and dynamic stability. The present results also showed that, during terminal stance, the PL exhibited a higher frequency of a hip-dominant in-phase hip extension–ankle plantarflexion coordination pattern than both the IL and the HIs. Joint kinematics also showed that, during terminal stance, both the IL and the HIs exhibited ankle plantarflexion, enabling effective push-off, while the hip simultaneously reached greater extension. In contrast, the PL exhibited less hip extension than both the IL and the HIs. Baum et al. also reported that hip extension moment during the stance phase was lower in the PL than in the IL(29). The reduced hip extension observed in the present study may be consistent with this finding, suggesting a reduced contribution of the hip to propulsion and the transition from stance to swing. In addition, during the push-off and pre-swing phases, the PL exhibited a significantly lower frequency of an ankle-dominant in-phase hip extension–ankle plantarflexion coordination pattern than the HIs. In

this coordination pattern, the ankle plantarflexes while the hip extends (ankle-dominant). Similarly, Previous studies have reported that increasing prosthetic push-off power reduced positive hip work and hip power while increasing step length and forward propulsion(30). These findings suggest that improving prosthetic push-off function reduces the need for compensatory strategies at the hip. Because the participants in the present study used SACH prostheses, which lack an articulated ankle joint and are unable to generate active push-off, the prosthetic limb provides only limited forward propulsion. Therefore, the lower frequency of the ankle-dominant in-phase coordination pattern may reflect a greater dependence on the hip to compensate for the reduced forward propulsion of the prosthesis and to facilitate the initiation of the swing phase. Joint kinematic analysis also showed that the PL exhibited a smaller hip range of motion during this phase than both IL and the HIs. Furthermore, during initial swing, the PL demonstrated a higher frequency of a hip-dominant anti-phase hip flexion–ankle plantarflexion coordination pattern than the HIs. In this coordination pattern, the hip flexes while the ankle moves toward plantarflexion rather than dorsiflexing to lift the foot. This may reflect the lack of dorsiflexors activity in PL since it usually helps clear the toes during swing. Therefore, this finding can be attributed to an increased use of hip strategies to compensate for the fact that the prosthetic ankle cannot perform dorsiflexion, even though toe clearance was not measured. Consistent with this interpretation, previous studies have reported that individuals with transtibial amputation increase hip joint torque while reducing knee flexor torque in the prosthetic limb to elevate the limb and achieve sufficient toe clearance during swing, thereby compensating for the absence of active ankle function (31). This increase is likely due to the use of compensatory strategies to increase toe clearance and prepare the limb for initial contact; strategies such as hip hiking, vaulting, and limb rotation in the transverse plane (circumduction) and increased knee flexion, which amputees use to compensate for the impaired active dorsiflexion of the prosthetic ankle during the swing phase (32).

In contrast, during terminal stance and initial swing, the PL demonstrated a different coordination strategy from the IL and HIs, characterized by a lower frequency of an anti-phase hip extension–knee flexion coordination pattern with knee dominance (90-135). This is probably due to the reduced forward propulsion generated by the PL because of the absence of active plantar-flexor function. Accordingly, the transition from stance to swing is altered, resulting in changes in movement timing and locomotor strategy (33). .although this could not be quantitatively measured. Reduced capability of propelling with a prosthetic leg leads to compensations in proximal joints, according to Hadj-Moussa et al(34). The PL displayed greater occurrences of the hip-dominant hip extension-knee extension in phase pattern compared to the IL in bin 180-225°, but this result was not statistically significant even after correction for multiple comparisons ($p_{adj} = .159$). This finding is believed to be associated with the abnormal sensorimotor control of the PL. As there is no physiological sensory input in case of prostheses, it has been stated in the literature that the TTA patients use indirect sources of sensory information for controlling the position of the prosthesis limb (35)(36). As a result, they often adopt compensatory locomotor strategies, including altered lower-limb kinematics and greater reliance on proximal joints, to maintain stable locomotion. The greater reliance on proximal joints observed in the present study may therefore represent one of these compensatory adaptations. While this technique can help runners develop stability, there is speculation that such proximal adjustments could compromise locomotor

efficiency and lead to a higher risk of developing secondary musculoskeletal problems due to increased metabolic costs associated with this adjustment (37).

The present results also revealed significant differences in the knee–ankle coordination pattern during late stance. In late stance (push-off), the PL exhibited a lower frequency of an ankle-dominant in-phase knee extension–ankle plantarflexion coordination pattern than the HIs. This observation is consistent with reduced functional coordination between the knee and ankle during mid-stance. In HIs, the plantar-flexor muscles eccentrically control tibial progression while knee extension occurs simultaneously, providing favorable conditions for elastic energy storage and release and subsequent propulsion generation. In contrast, because the PL lacks the plantar-flexor muscles and active ankle control, its ability to generate and control ankle moment during stance is reduced. Accordingly, the functional coupling between the knee and ankle may also be diminished during this phase. Sepp et al. reported that reduced ankle work was accompanied by altered knee and hip function, supporting the notion that reduced ankle contribution is associated with changes in inter-joint coordination during locomotion (38). In addition, during terminal stance, the PL showed a lower frequency of an ankle-dominant anti-phase ankle plantarflexion–knee flexion coordination pattern than the HIs. These results may reflect altered knee–ankle intersegmental coordination during the forward propulsion of running. Hamner et al. demonstrated that the plantar-flexor muscles, particularly the soleus and gastrocnemius, are the primary contributors to forward propulsion during terminal stance and play a critical role in facilitating the transition from stance to swing during running(39). The findings of the present study are consistent with this interpretation (39). The greater knee flexion observed in the IL may reflect the preserved function of the plantar-flexor muscles, facilitating forward propulsion and a more effective transition from stance to swing. In contrast, because the PL cannot actively generate plantarflexion, it appears to adopt a different compensatory strategy. Differences between the PL and the IL did not survive correction, and the IL did not differ from the HIs.

Movement variability is considered a fundamental characteristic of the human motor system, reflecting the capacity of the neuromuscular system to adapt and respond appropriately to varying task demands. Although high movement variability was traditionally regarded as an indicator of impaired motor control, contemporary perspectives suggest that an optimal level of variability is essential for maintaining motor flexibility and adaptability. Within this framework, healthy motor behavior is characterized not by minimal variability, but by the ability to generate organized, structured, and task-specific variations in movement (40)(20). Analysis of coordination variability revealed that, although the PL exhibited greater coordination variability than both the IL and the HIs during some portions of the running cycle, particularly between 25% and 30% of the running cycle, these differences did not achieve statistical significance ($p > 0.05$) for either the hip–knee or hip–ankle coupling. Nevertheless, although these differences were not statistically significant, the relatively small sample size ($n = 8$) may have limited the statistical power to detect between-c

Conclusion

The present study confirmed that subjects with TTA adopted a modified lower limb coordination pattern in order to counteract the lack of active ankle joint movements while running. The prosthesis limb mainly depends upon the proximal joints and hip-dominant coordination strategy

is adopted in order to propel and stabilize the body in the anteroposterior direction. Consequently, due to the lower propelling action of the prosthesis, the intact limb adopts changes in its stance-to-swing transition as well as increases the role of hips in this regard. Similarly, TTA subjects exhibited significantly higher variability in knee-ankle coordination pattern of their prosthesis limb in relation to the intact limb and HIs subjects during the stance-to-swing transition. Increased variability in the coordination pattern could be attributed to the mechanical limitations of the SACH prosthesis, lack of active plantar flexors movement, and change in sensory motor system after the amputation.

However, there are several limitations of the current study which should be mentioned. Firstly, the small sample size and heterogeneity of the participants in terms of age and cause of amputation can affect the generalizability of results to the wider population of people with TTA. Secondly, as all participants wore SACH prostheses, it is difficult to generalize the obtained results to other prosthetic devices like ESR or powered prostheses that have different effects on running economy and coordination. Finally, coordination patterns and coordination variability were assessed only in the sagittal plane; joint coordination in the frontal and transverse planes could also play an important role in compensation strategies during running. One of the obvious limitations of the current study is the lack of objective measure of the running speed. As participants were running at a self-selected and comfortable speed without the help of the timing gates, it cannot be excluded that there are potential differences in the actual running velocity. As running speed is one of the major determinants of the lower-limb kinematics and coordination variability, some differences may be caused not by the amputation but by the running speed. Besides, as the members of the provincial amputee football team were recruited, they may have had a significant mismatch in age, physical condition and running experience comparing to the healthy participants. Also, detailed clinical information of the TTA participants (the cause of amputation, time after amputation, specific socket/suspension design) was not gathered. Due to the small number of participants, this aspect is considered to be a limitation, as these clinical characteristics can add mechanical variability at the socket/residual limb interface.

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Declaration of Generative AI and AI-assisted technologies in the writing process

No generative artificial intelligence tools were used to generate, analyze, or interpret the study data. AI-assisted language translation and editing were used solely to improve the English clarity and grammar of the manuscript. All scientific content was thoroughly reviewed and verified by the authors.

Data availability

The datasets used and/or analysed during the current study are available from the corresponding author on reasonable request.

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