Dynamics of Center of Pressure Trajectory in Gait: Unilateral Transfemoral Amputees Versus Non-Disabled Individuals

Yufan He, Mingyu Hu, Abu Jor, Hiroaki Hobara, Fan Gao, and Toshiki Kobayashi

***Abstract*—The primary goal of rehabilitation for individuals with lower limb amputation, particularly those with unilateral transfemoral amputation (uTFA), is to restore their ability to walk independently. Effective control of the center of pressure (COP) during gait is vital for maintaining balance and stability, yet it poses a significant challenge for individuals with uTFA. This study aims to study the COP during gait in individuals with uTFA and elucidate their unique compensatory strategies. This study involved 12 uTFA participants and age-matched non-disabled controls, with gait and COP trajectory data collected using an instrumented treadmill. Gait and COP parameters between the control limb (CL), prosthetic limb (PL), and intact limb (IL) were compared. Notably, the mediolateral displacement of COP in PL exhibited significant lateral displacement compared to the CL from 30% to 60% of the stance. In 20% to 45% of the stance, the COP forward speed of PL was significantly higher than that of the IL. Furthermore, during the initial 20% of the stance, the vertical ground reaction force of PL was significantly lower than that of IL. Additionally, individuals with uTFA exhibited a distinct gait pattern with altered duration of loading response, single limb support, pre-swing and swing phases, and step time. These findings indicate the adaptability of individuals with uTFA in weight transfer, balance control, and pressure distribution on gait stability. In conclusion, this study provides valuable insights into the unique gait dynamics and balance strategies of uTFA patients, highlighting the importance of optimizing prosthetic design, alignment procedures, and rehabilitation programs to enhance gait patterns and reduce the risk of injuries due to compensatory movements.**

***Index Terms*—Amputation, walk, rehabilitation, gait line, prosthetic.**

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I. Background

FOR individuals with unilateral transfemoral amputation (uTFA), a pivotal rehabilitation goal is to regain the capacity for autonomous walking. Efficient ambulation requires effective control of the center of pressure (COP) to maintain balance and stability [1]. A detailed analysis of center of pressure (COP) trajectories can unveil how individuals with uTFA control their COP and modify body movements to maintain balance, enhancing comprehension of the compensation strategy. Moreover, a better understanding of how COP trajectories are influenced by transfemoral amputation is important for refining prosthesis design, achieving an individualized fit, and optimizing rehabilitation programs. However, little is known about alterations in plantar COP trajectories during walking in individuals with uTFA.

Individuals with uTFA absence of the typical postural control strategies, such as an ankle strategy [2]. The lack of a typical ankle strategy entails a temporally delayed initiation of muscle activation, including the ankle, thigh, and trunk muscles, which extend proximally towards the dorsal or ventral aspect of the body [3]. This chain action poses a significant challenge to this population when conducting balance activities [4], [5]. Individuals with unilateral transfemoral amputation (uTFA) must therefore adapt and compensate for the loss of the limb [6], [7]. For example, the COP unconsciously shifts toward the intact limb when standing still [8], and the intact limb bears body weight for a longer time duration than the prosthetic limb during their respective single limb support of the gait [7]. The neuromuscular system in individuals with uTFA also adapts to enhance balance and coordination, regaining an effective functional compensatory strategy for postural control [9]. Given the inherent intact and prosthetic limb asymmetry in the gait of individuals with uTFA and their degraded balance, the compensatory demands on the intact limb can become excessive and lead to hip, knee osteoarthritis and generalized low back pain [10]. Investigating COP is instrumental in comprehending the balance control strategies employed by individuals with uTFA.

Prosthetic alignment influences the relative position of ground reaction force (GRF) lines, impacting joint moments. Notably, for individuals with uTFA, adding the socket extension angle or shifting the socket posteriorly in the sagittal plane increases knee extension moment, while adding the socket abduction angle or laterally moving the socket in the coronal plane increases the knee valgus moment and vice versa [11]. Malalignment significantly alters the magnitude and loading pattern of ground reaction force (GRF) [12], [13]. Investigating COP dynamics may aid in optimizing prosthetic alignment to minimize the risk of injuries. Hence, studying COP trajectories particularly its movements and changes, may potentially aid clinicians in aligning prostheses appropriately.

Despite significant advancements in prosthetic technology and rehabilitation, considerable gaps persist in comprehension of the COP trajectory dynamics in the gait of individuals with uTFA. One study found that individuals with unilateral amputation exhibit an unsteady gait pattern which may caused by a mismatch between the COP and the center of mass (COM) in prosthetic limb (the correlation between the instantaneous positions of COP and COM and COM acceleration was positive, and it was expected to be negative under normal circumstances) [14]. A recent study using butterfly diagrams has revealed that individuals with uTFA may control balance by increasing the standard deviations of lateral displacement during gait [15]. Another study has demonstrated substantial asymmetry in COP profiles of individuals with uTFA during walking could serve as a basis for clinical assessment [16]. Moreover, COP trajectories in individuals with uTFA have been altered in the presence of pathological conditions (e.g. vaulting) [17]. Comparing the COP trajectories between individuals with uTFA and non-disabled persons could help clinicians and/or researchers to identify potential challenges in walking.

Several studies have highlighted the significance of forward COP speed in gait analysis. One study revealed that as walking speed increased, forward COP speed increased during mid-stance, but decreased during terminal stance and pre-swing [18]. COP speed can also serve as an indicator for evaluating fall risk and balance [19], [20]. A typical COP speed curve of normal gait exhibits a three-peak pattern, with the first peak occurring at initial contact, the second at mid-stance, and the third at pre-swing [21]. Nevertheless, COP speed patterns of individuals with uTFA have not been thoroughly studied.

Traditionally, majority of studies only extract discrete parameters including peaks and troughs from the COP profile without considering the temporal and spatial variability of COP and its associated parameters. This can result in some limitations as the peak value may occur at different time points. To solve this issue, one strategy is to compare the differences between or among curves.

The present study aims to examine and contrast variations in COP curves and parameters during walking between non-disabled controls and individuals with uTFA. Understanding the alterations in the gait pattern of individuals with uTFA through the analysis of continuous COP time series can offer valuable insights for enhancing prosthetic gait rehabilitation. It was hypothesized that the dynamics of COP trajectories and related parameters in individuals with uTFA significantly differed from those in non-disabled during gait.

II. Methods

A. Participants

This study recruited a total of twelve individuals with uTFA, including ten males and two females. The demographic information included basic characteristics (gender, age, body height, body mass, time since amputation, walking speed, amputated side, and etiology) and prosthetic information (prosthetic knee, prosthetic foot, socket type, suspension type, and liner material) (Table I). All participants demonstrated a good proficiency in utilizing their prostheses. The inclusion criteria were 1) participants must be 18 years old or older, and had experienced a unilateral transfemoral amputation; 2) participants had no neuromusculoskeletal complications other than the amputation itself, 3) prospective participants were required to possess the capability to use prostheses proficiently and ambulate on a treadmill without the need of assistive devices. For each individual with uTFA, an age-matched non-disabled person was included as a control. The study received official approval from the Human Subjects Ethics Sub-Committee of the Hong Kong Polytechnic University (number: HSEARS20220719001). Before participation, each individual involved in the study provided their informed consent.

B. Experimental Procedures

Zebris FDM-T treadmill (Zebris Medical GmbH, Germany) was employed to acquire a comprehensive dataset of COP and spatiotemporal gait parameters (Figure 1 (a)). The treadmill had demonstrated good reliability and validity of spatiotemporal parameters in prior studies [22], [23]. Before the data collection, participants underwent a warm-up period lasting no less than 5 minutes [24]. During this stage, participants were encouraged to acclimatize themselves to the treadmill environment and were requested to ambulate without the assistance of any walking aids. Nevertheless, handrails were available and accessible to participants as an option for support if needed. In the subsequent stage of the experiment, participants' self-selected walking speeds were determined to reflect their natural and comfortable gait patterns. These self-selected speeds were used for the ensuing data collection, which consisted of two separate one-minute trials. The experimental protocol prioritized the participants' safety and well-being. To this end, adequate intervals of rest were provided between the two one-minute trials to minimize the influence of potential fatigue. Furthermore, participants wore safety harnesses, the tension of which was set to an appropriate level, assuming their safety throughout the walking trials.

C. Data Collection and Analysis

Data processing was carried out using the Zebris FDM software (Zebris Medical GmbH, Germany). Vertical ground reaction force (vGRF) data was collected by the embedded force sensors with an area of 0.85*0.85 cm^2 at a 300Hz

TABLE I
DEMOGRAPHIC DATA OF PARTICIPANTS

Participants	Gender (M/F)	Age (year)	Height (m)	Mass (kg)	Time since amputation (year)	Walking speed (m/s)	Amputated side	Etiology	Prosthetic knee	Prosthetic foot	Socket type	Suspension	Liner
1	M	54	1.85	71	34	0.33	L	Tumor	3R80	1C60 Triton	IRC	Suction	Urethane
2	M	43	1.74	100	7	0.39	L	Trauma	Paso	Rush Foot	IC	Suction	N.A.
3	F	55	1.56	53	5	0.47	R	Trauma	KX06	Pro-Flex XC	IRC	Suction	Polyurethane
4	M	44	1.78	101	4	0.42	L	Trauma	Jiguang	Multiflex Foot	IRC	Locking	Silicone
5	M	61	1.63	54	12	0.47	L	Infection	3R78	Single-axis Foot	IC	Suction	Urethane
6	F	56	1.75	93	7	0.47	L	Trauma	3R-106 PRO	Pro-Flex XC	IC	Locking	Silicone
7	M	48	1.69	68	27	0.47	L	Trauma	Covered	Covered	QL	Suction	N.A.
8	M	57	1.77	65	7	0.50	L	Tumor	Paso	Pro-Flex XC	IRC	Locking	Silicone
9	M	52	1.76	78	6	0.50	L	Infection	Mauch	Pro-Flex LP Torsion	IC	Suction	Urethane
10	M	62	1.78	65	45	0.56	R	Trauma	Single-axis Knee	Single-axis Foot	QL	Belt	N.A.
11	M	49	1.76	74	11	0.58	R	Trauma	Paso	Pro-Flex XC	IRC	Suction	Urethane
12	M	66	1.79	71	53	0.78	L	Trauma	RHEO	Pro-Flex XC	IC	Suction	Silicone
Mean	—	53.92	1.74	74.42	18.17	0.50	—	—	—	—	—	—	—
SD	—	6.81	0.07	15.38	16.45	0.11	—	—	—	—	—	—	—
Mean (CT)	—	54.25	1.66	66.83	—	0.65	—	—	—	—	—	—	—
SD (CT)	—	7.21	0.09	10.40	—	0.06	—	—	—	—	—	—	—

Abbreviations: M: male; F: female; CT: control; R: right; L: left; IRC: ischial-ramal containment; IC: ischial containment; QL: quadrilateral; N.A.: not applicable.

Prosthetic knee: KX06 is from Endolite, India; Mauch knee, Paso knee, and RHEO knee are from Ossur, Iceland; 3R-106 PRO, 3R78, and 3R80 are from Ottobock, Germany; Jiguang knee is from Zeanso, China.

Prosthetic foot: Pro-Flex XC and Pro-Flex LP Torsion are from Ossur, Iceland; Rush foot is from PROSTEK, Australia; Multiflex is from Endolite, India; 1C60 Triton is from Ottobock, Germany.

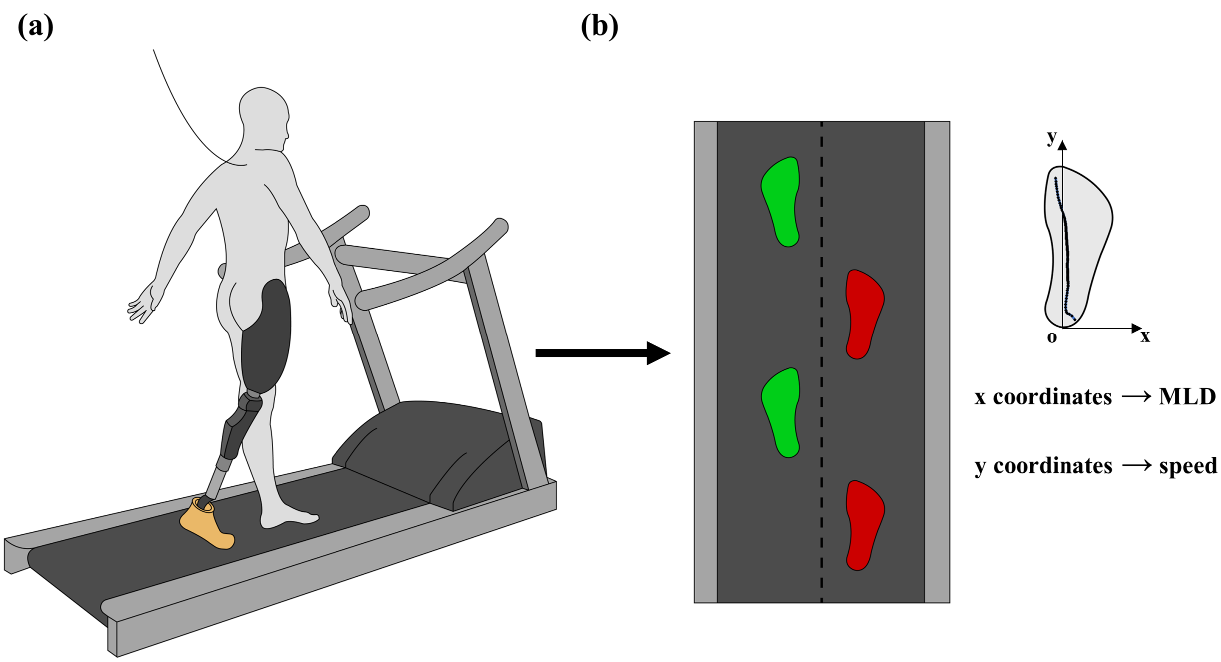

Fig. 1. Experimental setting. Participants performed experimental trails on the treadmill at their selected walking speed while wearing safety belts. (a) Schematic illustration of the experimental setup, (b) Schematic illustration of the treadmill for data collection. COP mediolateral displacement (MLD) was calculated by x coordination, COP forward speed was calculated by y coordination.

sampling rate. The coordinates of the plantar foot COP in the x (mediolateral) direction and y (anterior-posterior) direction was systematically collected for further analysis (Figure 1 (b)). Absolute plantar COP coordinates were automatically transformed into relative coordinates. In this transformation, positive values in the x direction indicated lateral displacement, while negative values indicated medial displacement. Similarly, positive values in the y direction denoted anterior movement, and negative values represented posterior movement. The loading response was defined as the

TABLE II
SUMMARY OF THE DEFINITION OF PARAMETERS

Parameters	Formula	Units
Max LD in LR	$max(X_n), n \in LR$	mm
Max LD in SLS	$max(X_n), n \in SLS$	mm
Max MD in PS	$min(X_n), n \in PS$	mm
Range of MLD	$max(X_n) - min(X_m), n, m = 0, 1,2, \ldots, 100$	mm
Max speed in LR	$max(V_n), n \in LR$	% walking speed
Min speed in SLS	$min(V_n), n \in SLS$	% walking speed
Max speed in PS	$max(V_n), n \in PS$	% walking speed
First peak	$max(F_n), n = 0, 1,2, \ldots, 50$	% body weight
Second peak	$max(F_n), n = 51, 52,53, \ldots, 100$	% body weight
RMS of ML COP	$\sqrt{\frac{1}{N}\sum_n X_n^2}, n = 0,1,2, \ldots, 100$	mm
RMS of speed	$\sqrt{\frac{1}{N}\sum_n V_n^2}, n = 0,1,2, \ldots, 100$	% walking speed
RMS of force	$\sqrt{\frac{1}{N}\sum_n F_n^2}, n = 0,1,2, \ldots, 100$	% body weight

Abbreviations: LD: lateral displacement; MD: medio displacement; MLD: mediolateral displacement; LR: loading response; SLS: single limb support; PS: pre-swing; RMS: root mean square; COP: center of pressure; X_n: mediolateral displacement value of the trajectory of COP in each point of the normalized curve; V_n: the value of the speed in each point of the normalized curve; F_n: the value of the force in each point of the normalized curve; n: numbers of the points in the normalized curve.

phase between initial ground contact and contralateral toe-off, single limb support was defined as the phase between contralateral toe-off and contralateral initial ground contact, pre-swing was defined as the phase between contralateral initial ground contact and toe-off of the ipsilateral side, and the swing phase was defined as the phase when the foot is not in contact with the ground. The maximum lateral displacement in loading response (Max LD in LR) was defined as the maximum value in the x direction during the loading response, the maximum lateral displacement in single limb support (Max LD in SLS) was established as the maximum value in the x direction during single limb support, and the maximum medial displacement in pre-swing (Max MD in PS) was defined as the minimum value in the x direction during the pre-swing. Additionally, the range of medial-lateral displacement (Range of MLD) was calculated as the difference between the maximum lateral displacement and the maximum medial displacement. To facilitate observation and comparison, all left-side COP trajectories were symmetrically labeled as right-side. The mediolateral displacement of COP along x direction was normalized to 101 points, corresponding to the entire stance phase ranging from 0% to 100%.

Forward COP speed was computed as the first-order derivative of the y direction coordinates and was normalized to 101 points from 0% to 100% of the stance phase. Additionally, maximum and minimum speeds were separately determined for distinct gait phases (e.g., loading response, single limb support, and pre-swing). All speeds were normalized as a percentage of walking speed and parameters were extracted according to the percentage of different gait phases. In particular, the maximum speed in the loading response (Max speed in LR) was established as the largest value during the loading response; the minimum speed in the single limb support (Min speed in SLS) was set as the lowest value during the single limb support; and the maximum speed in the pre-swing (Max speed in PS) was defined as the peak value during the pre-swing [25]. The vGRF was also standardized across the entire stance phase (0% – 100%), enabling the calculation of both the first and second peaks. The first peak was determined as the largest value within the initial 50% of the stance phase, while the second peak was identified as the largest value within the latter 50% of the stance phase. All forces were normalized to body weight as a percentage. Root mean square (RMS) values for both medial and lateral displacements, speed of COP, and vGRF were computed individually. The names of parameters, formulas, and units for all these COP related parameters are detailed in Table II. The spatiotemporal parameters including foot rotation angle (toe-in or toe-out angles), length of COP trajectory (distance in y direction), step length, step time, loading response time duration, single limb support time duration, pre-swing time duration, and swing time duration were also collected. Due to the minimal difference between the left and right sides in non-disabled controls, the right limb was consistently chosen as the control limb for this study. Thus, in total three limbs i.e., i) control limb (CL) of non-disabled individuals, ii) intact limb (IL), and iii) prosthetic limb (PL) of individuals with uTFA have been considered for the investigation.

D. Statistical Analysis

For discrete parameters, the assessment of normality for the data was undertaken using the Shapiro-Wilk test. For those datasets found to adhere to a normal distribution, a one-way analysis of variance (ANOVA) was employed to investigate differences between the three limbs (CL, IL, and PL), with subsequent application of Bonferroni post-hoc comparisons to correct for multiple comparisons. In cases where the data did not conform to a normal distribution, the independent-samples Kruskal-Walli's test was used to detect potential

TABLE III
MEAN VALUE, SD VALUE, NORMALITY, P-VALUE, AND POST-HOC OF THE COP PARAMETERS

Parameters	Limbs CL	IL	PL	Normality	p-value (Main effect)	Post-hoc C – I	C – P	I - P
Max LD in LR (mm)	13.30 ± 8.87	20.50 ± 8.82	12.69 ± 7.45	Y	0.07	—	—	—
Max LD in SLS (mm)	3.10 ± 1.38	6.37 ± 5.70	9.92 ± 4.15	Y	**<0.01***	0.22	**<0.01***	0.16
Max MD in PS (mm)	-15.07 ± 4.10	-12.03 ± 6.61	-16.91 ± 10.63	N	0.59	—	—	—
Range of MLD (mm)	28.53 ± 8.16	32.64 ± 8.68	32.27 ± 9.89	Y	0.50	—	—	—
Max speed in LR (%)	117.43 ± 34.49	184.16 ± 67.10	133.05 ± 46.65	N	**0.04***	**0.04***	1.00	0.28
Min speed in SLS (%)	15.32 ± 6.46	-4.12 ± 9.40	22.89 ± 8.37	Y	**<0.01***	**<0.01***	0.12	**<0.01***
Max speed in PS (%)	102.34 ± 13.81	138.51 ± 48.02	134.72 ± 52.12	N	0.10	—	—	—
First peak (%)	104.45 ± 5.54	97.35 ± 4.71	99.71 ± 7.25	Y	**0.03***	**0.02***	0.21	1.00
Second peak (%)	95.50 ± 4.59	95.75 ± 4.82	92.82 ± 4.77	Y	0.29	—	—	—
RMS of ML COP (mm)	5.51 ± 1.35	6.88 ± 2.51	9.14 ± 2.28	Y	**<0.01***	0.42	**<0.01***	0.05
RMS of speed (%)	49.68 ± 6.48	63.81 ± 15.34	66.22 ± 13.14	N	**<0.01***	**0.05***	**0.01***	1.00
RMS of force (%)	78.83 ± 3.29	79.49 ± 3.53	73.31 ± 4.58	Y	**<0.01***	1.00	**<0.01***	**<0.01***

Abbreviations: CL: control limb; IL: intact limb; PL: prosthetic limb; Y: yes; N: no; C: control (right) limb of non-disabled individuals; I: intact limb of individuals with uTFA; P: prosthetic limb of individuals with uTFA; LD: lateral displacement; MD: medio displacement; MLD: mediolateral displacement; LR: loading response; SLS: single limb support; PS: pre-swing; RMS: root mean square; COP: center of pressure. An asterisk (*) represents the statistical significance level $p < 0.05$.

TABLE IV
MEAN VALUE, SD VALUE, NORMALITY, P-VALUE, AND POST-HOC OF THE SPATIOTEMPORAL PARAMETERS

Parameters	Limbs CL	IL	PL	Normality	p-value (Main effect)	Post-hoc C – I	C – P	I - P
Foot rotation angle (°)	11.98 ± 4.37	9.58 ± 4.08	11.37 ± 5.66	Y	0.48	—	—	—
Step length (cm)	38.54 ± 5.84	36.75 ± 6.74	34.46 ± 9.69	Y	0.46	—	—	—
Loading response (%)	17.96 ± 2.34	17.11 ± 2.54	20.77 ± 3.35	Y	**0.01***	1.00	0.07	**0.01***
Single limb support (%)	32.28 ± 2.48	36.33 ± 2.49	25.80 ± 3.44	Y	**<0.01***	**0.01***	**<0.01***	**<0.01***
Pre-swing (%)	17.76 ± 2.34	20.75 ± 3.34	17.10 ± 2.53	Y	**0.01***	**0.05***	1.00	**0.01***
Swing phase (%)	32.00 ± 2.22	25.81 ± 3.43	36.35 ± 2.48	Y	**<0.01***	**<0.01***	**<0.01***	**<0.01***
Step time (s)	0.59 ± 0.06	0.68 ± 0.12	0.78 ± 0.12	Y	**<0.01***	0.14	**<0.01***	0.09
Length of COP trajectory (mm)	229.60 ± 30.11	236.33 ± 19.36	251.94 ± 16.78	Y	0.08	—	—	—

Abbreviations: CL: control limb; IL: intact limb; PL: prosthetic limb; Y: yes; N: no; C: control (right) limb of non-disabled individuals; I: intact limb of individuals with uTFA; P: prosthetic limb of individuals with uTFA; COP: center of pressure. An asterisk (*) represents the statistical significance level $p < 0.05$.

differences among the lower limbs. The entire spectrum of statistical analyses was executed with the utilization of SPSS (IBM SPSS Statistics 26, SPSS Inc., Chicago, IL). A predefined significance level of $p < 0.05$ was adopted as the threshold for determining the presence of statistically significant differences. For curves, the differences in COP mediolateral displacement, forward speed of COP, and vGRF curves were detected using statistical parametric mapping (SPM). The initial step involved applying a one-way ANOVA test to the curves from the control limbs, the intact limbs, and the prosthetic limbs. Subsequently, a Bonferroni post-hoc comparison was executed to adjust the p-value for multiple comparisons and determine the presence of any significant differences between the three limb groups [26]. This process was accomplished by adapting the open-source spm1d code (Version 0.4, available at https://spm1d.org/) in Python (Version 3.11) [27].

III. RESULTS

The findings of COP-related parameters and discrete spatiotemporal parameters are shown in Table III and Table IV, respectively. The mean mediolateral COP displacement curves, the mean forward speed of COP curves, the mean vGRF curves, and their corresponding SPM results were presented in Figure 2, Figure 3, and Figure 4, respectively.

A. Mediolateral Displacement of COP Parameters and Curves

There were no differences in the maximum lateral displacement in loading response, the maximum medial displacement in pre-swing, and range of mediolateral displacement. However, there were significant differences in the maximum lateral displacement in single limb support, with the lateral displacement of the PL being greater than that of the CL ($p < 0.01$, CL: 3.10 ± 1.38 mm, IL: 6.37 ± 5.70 mm, PL: 9.92 ± 4.15 mm) (Table III). The RMS values of mediolateral COP displacement of the PL were significantly larger than that of the CL ($p < 0.01$, CL: 5.51 ± 1.35 mm, IL: 6.88 ± 2.51 mm, PL: 9.14 ± 2.28 mm). In terms of the mediolateral displacement of the COP curve (Figure 2 (a)), the primary distinction was evident during the 30% to 60% of the stance phase (Figure 2 (a1)). Subsequent post-hoc tests revealed that within this period, the PL exhibited greater lateral displacement than the CL (Figure 2 (a4)). Figure 2 (b) displayed the averaged plantar pressure distributions across the three limbs during the stance phase. Notably, two-dimensional SPM did not reveal any significant differences between them.

B. Forward Speed of COP Parameters and Curves

The COP forward speed of CL exhibited a characteristic three-peak curve, with elevated speeds during the loading

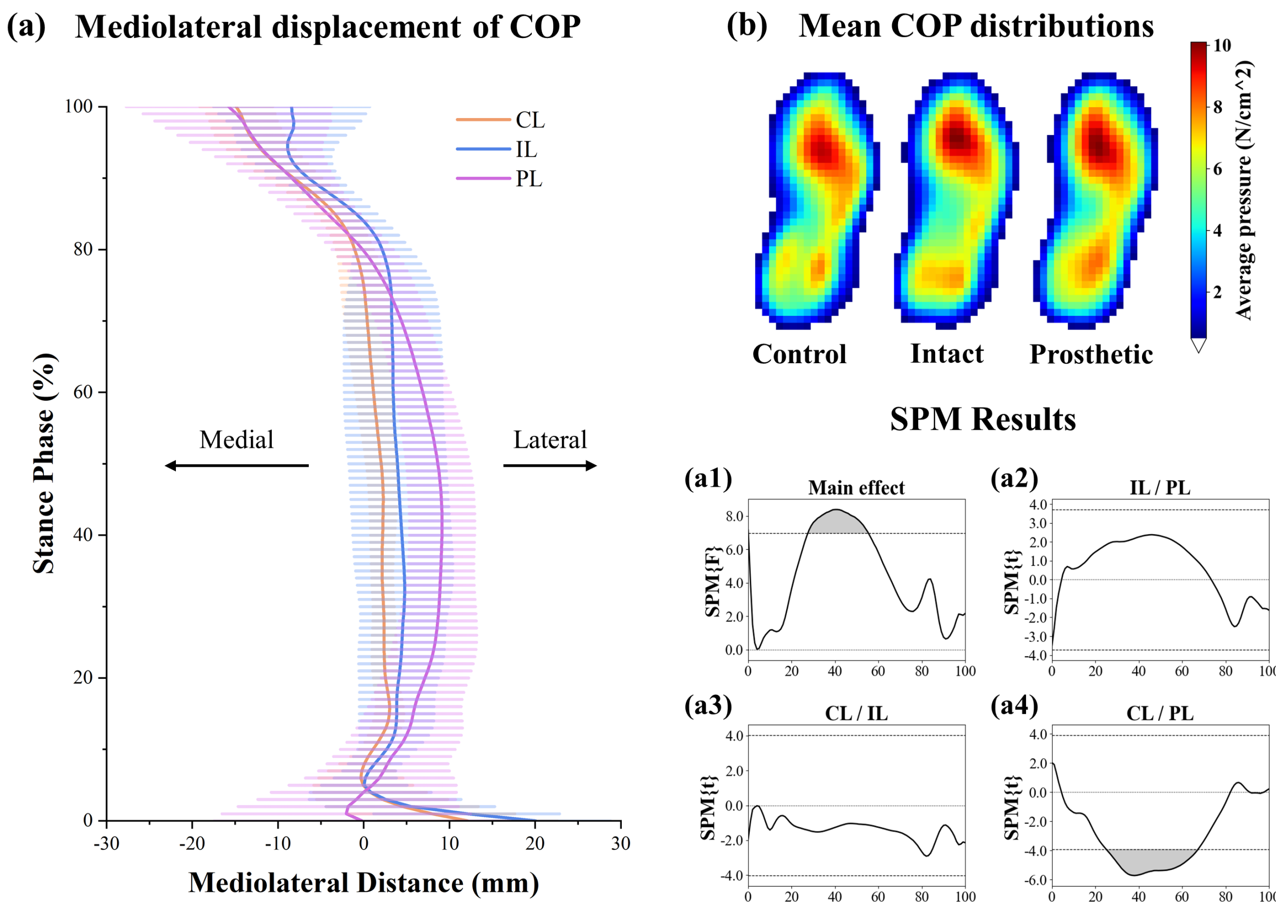

Fig. 2. Results of mediolateral displacement of COP, SPM analysis, and mean COP distribution. The orange, blue, and pink lines represent the control limb (CL), intact limb (IL), and prosthetic limb (PL), respectively. The (a) is the COP trajectory during the stance phase with standard deviations of the three limbs. The (a1), (a2), (a3), and (a4) show the SPM results for the main effect and post-hoc test using Bonferroni correction. The grey parts indicate significant differences. The (a1) shows the main effects among the three limbs. The (a2) shows the statistical significance between the IL and the PL. The (a3) shows the statistical significance between the CL and the IL. The (a4) shows the statistical significance between the CL and the PL. The (b) shows the mean COP distribution during the stance of the three limbs.

response, around 40% of the stance phase, and at push-off. In contrast, the second peak in IL diminishes, while the first and last peaks amplify in comparison to CL. Notably, the PL did not exhibit discernible peaks across the entire of the stance phase. All forward COP speed-related parameters displayed noteworthy disparities except maximum speed in pre-swing. Specifically, the maximum forward COP speed of the IL in loading response was significantly higher than that of the CL ($p = 0.04$, CL: $117.43 \pm 34.49\%$, IL: $184.16 \pm 67.10\%$, PL: $133.05 \pm 46.65\%$). Furthermore, in the single limb support phase, the minimum forward COP speed of the IL was significantly less than that of the CL and PL ($p < 0.01$, CL: $15.32 \pm 6.46\%$, IL: $-4.12 \pm 9.40\%$, PL: $22.89 \pm 8.37\%$). Additionally, the RMS of forward COP speed of the CL was significantly smaller than that of the IL and PL ($p < 0.01$, CL: $49.68 \pm 6.48\%$, IL: $63.81 \pm 15.34\%$, PL: $66.22 \pm 13.14\%$) (Table III). The main effects and post-hoc tests indicated differences in forward COP speed during two specific periods: the first half of the mid-stance (20% - 45%) and the middle of the terminal stance (80% - 85%) (Figure 3 (a1)). Notably, in the first half of the single limb support, the forward COP speed of the PL was significantly higher than that of the IL (Figure 3 (a2)). While at the initial of the pre-swing, the forward COP speed of the PL surpassed that of the CL significantly (Figure 3 (a4)).

C. Vertical Ground Reaction Force Parameters and Curves

The first peak in the CL significantly exceeded that in the IL ($p = 0.03$, CL: $104.45 \pm 5.54\%$, IL: $97.35 \pm 4.71\%$, PL: $99.71 \pm 7.25\%$), while the second peak did not exhibit significant differences across all limbs. Additionally, the RMS of force in the PL was notably smaller compared to both the CL and IL ($p < 0.01$, CL: $78.83 \pm 3.29\%$, IL: $79.49 \pm 3.53\%$, PL: $73.31 \pm 4.58\%$) (Table III). Significant differences in the vGRF curves were primarily concentrated within the initial 20% of the stance phase and around 30% to 35% of the stance phase (Figure 4 (a1)). During the first 10% of the stance phase, the vGRF of the CL was significantly smaller than that of the IL (Figure 4 (a3)). Moreover, in the first 20% of the stance phase, the vGRF of the PL was notably lower compared to that of the IL (Figure 4 (a2)). At around 30% to 35% in the stance phase, the vGRF of the PL was also significantly lower than that of the CL (Figure 4 (a4)).

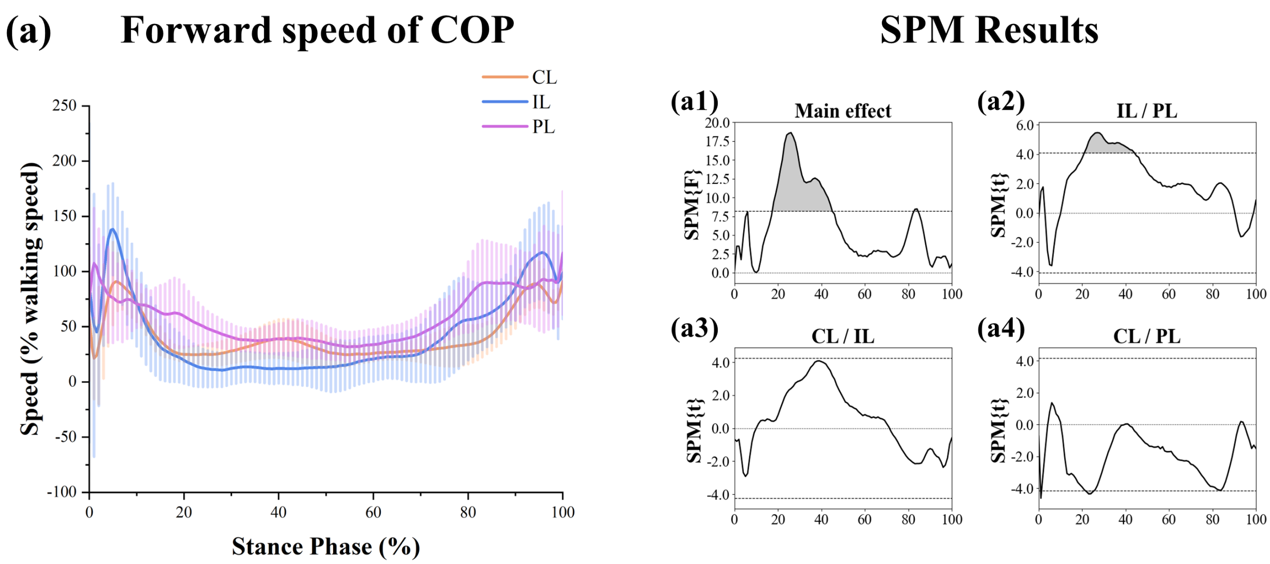

Fig. 3. Results of forward speed of COP and SPM analysis. The orange, blue, and pink lines represent the control limb (CL), intact limb (IL), and prosthetic limb (PL), respectively. The (a) is the COP trajectory during the stance phase with standard deviations of the three limbs. The (a1), (a2), (a3), and (a4) show the SPM results for the main effect and post-hoc test using Bonferroni correction. The grey parts indicate significant differences. The (a1) shows the main effects among the three limbs. The (a2) shows the statistical significance between the IL and the PL. The (a3) shows the statistical significance between the CL and the IL. The (a4) shows the statistical significance between the CL and the PL.

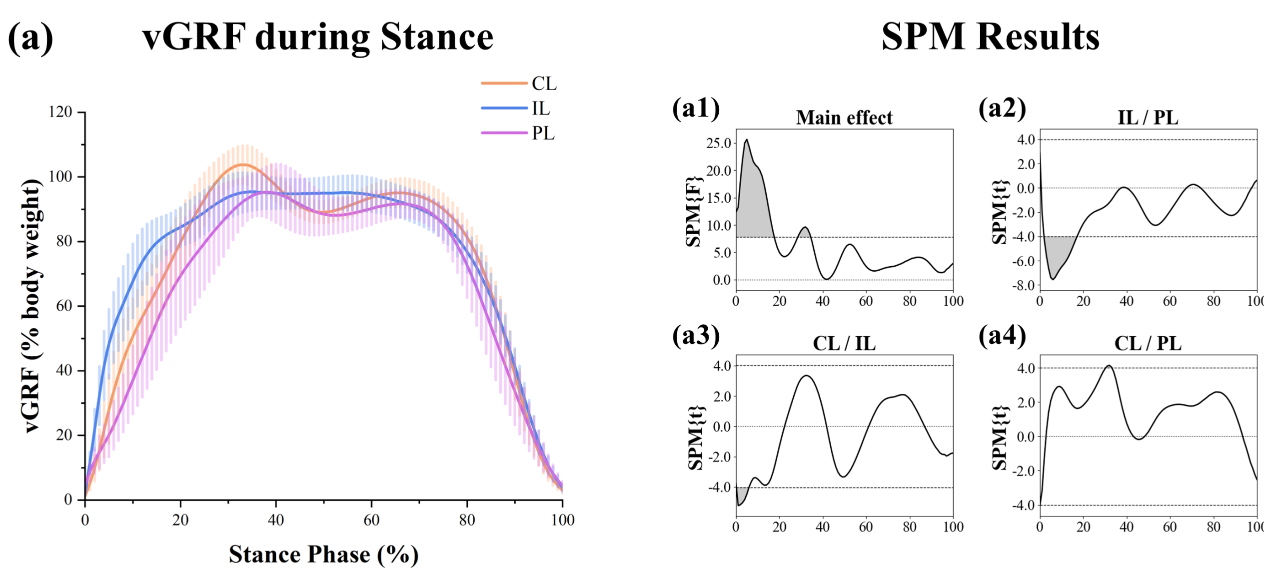

Fig. 4. Results of vGRF and SPM analysis. The orange, blue, and pink lines represent the control limb (CL), intact limb (IL), and prosthetic limb (PL), respectively. The (a) is the COP trajectory during the stance phase with standard deviations of the three limbs. The (a1), (a2), (a3), and (a4) show the SPM results for the main effect and post-hoc test using Bonferroni correction. The grey parts indicate significant differences. The (a1) shows the main effects among the three limbs. The (a2) shows the statistical significance between the IL and the PL. The (a3) shows the statistical significance between the CL and the IL. The (a4) shows the statistical significance between the CL and the PL.

D. Spatiotemporal Parameters

There were no differences observed in foot rotation angle, step length, or length of COP trajectory between the three limbs. However, the loading response time duration of the PL was significantly longer than that of the IL ($p = 0.01$, CL: 17.96 ± 2.34%, IL: 17.11 ± 2.54 %, PL: 20.77 ± 3.35 %). In single limb support time duration, significant differences were observed between all the three limbs, with the CL exhibiting a longer time duration than the PL and a shorter time duration than the IL ($p < 0.01$, CL: 32.28 ± 2.48%, IL: 36.33 ± 2.49%, PL: 25.80 ± 3.44%). Furthermore, the pre-swing time duration in the IL was significantly longer compared to both the CL and PL ($p = 0.01$, CL: 17.76 ± 2.34%, IL: 20.75 ± 3.34%, PL: 17.10 ± 2.53%). In regard to the swing phase, the CL had a significantly shorter swing time duration than the PL but a longer swing time duration than the IL ($p < 0.01$, CL: 32.00 ± 2.22%, IL: 25.81 ± 3.43%, PL: 36.35 ± 2.48%). Notably, the step time of PL was significantly longer than that of the CL ($p < 0.01$, CL: 0.59 ± 0.06 s, IL: 0.68 ± 0.12 s, PL: 0.78 ± 0.12 s) (Table IV).

IV. Discussion

This study compared plantar COP mediolateral displacement curves, forward COP speed curves, and vGRF curves, along with their associated discrete parameters, between individuals with uTFA and non-disabled individuals during gait. The hypothesis was supported as notable variations were evident in the period of 30% to 60% of the COP mediolateral displacement curve (Figure 2), the 20% to 45% of the forward COP speed curve (Figure 3), and the initial 20% of the vGRF curve (Figure 4). These results signified unique COP dynamics in individuals with uTFA, particularly during the first half of the stance.

In the mediolateral COP trajectory, the PL exhibited an initial medial displacement trend compared to the IL, succeeded by a large lateral movement of the COP trajectory. Further examination of the trajectory curves suggested that the disparity primarily occurs within the period of 20% to 60% of the stance phase. The medial initial contact can be attributed to the adoption of a conservative abduction alignment strategy during prosthesis fitting. Research has demonstrated that the prosthesis alignment can influence the lower limb force line, consequently affecting the gait of individuals with uTFA [11], [12], [28]. It is important to note that individuals with uTFA often present symptoms of residual limb abduction [29] and to accommodate this, the prosthesis was adjusted by abducting the socket during alignment. When the socket fails to provide sufficient room for residual limb abduction, it results in a relative abduction of the lower prosthetic components as the prosthesis must conform to the direction of the residual limb. Consequently, the medial side of the foot made initial contact with the ground, shifting the COP medially. Worth mentioning is that the majority of individuals with uTFA in this study had socket types of IC (ischial containment) (41.7%) or IRC (ischial ramal containment) (41.7%). The IC or IRC type of socket exerts a group of 3-points-force to the greater trochanter, ischial or ischial-ramal, and the lateral aspect of the residual limb, inducing adduction of the residual limb (Long's Line) [30], to some extent, contributing to a lack of abduction. Moreover, amputation results in an imbalance of muscle control between the medial and lateral regions of the residual limb, with the adductors' strength decreasing as the amputation level gets higher, while the abductors remain relatively unaffected [31]. This muscle control imbalance may also cause the prosthesis imbalance during stance. To preserve balance, individuals with uTFA tend to employ a lateral trunk lean (lateral COP movement) to reduce the moment arm of muscle force, which could be the reason why PL exhibited a lateral displacement during 30% to 60% of the stance. Notably, the variance in COP trajectory was less pronounced after 60% of the stance. This may be due to postural adjustments made by individuals with uTFA that shifting their COP medially to ensure the loading response on the contralateral foot [32].

The maximal forward COP speed in the IL was notably higher during the loading response and pre-swing phases. A higher speed implied a quicker movement of COP and weight transfer in the double limb support phase. Consequently, this elongated the single limb support time duration for the IL [33]. Additionally, the reduction in forward COP speed during single limb support of the IL also served as a strategy to extend the phase. Notably, during the single limb support phase, the IL's minimum speed was found to be a negative value in some individuals, indicating a gradual backward shift of the COP. However, it should be noted that the IL lacked a distinct peak COP speed during mid-stance (in comparison to the CL). This may be due to the PL's inability to be voluntarily controlled, the force must be transmitted from the residual limb to the socket first, and then drives the entire prosthetic limb during the swing phase. The PL is often "threw forward" during the swing phase. As a result, an accelerating disturbance occurred in the sagittal plane. To prevent imbalance of the trunk, the gluteus maximus on the IL side needs to contract additionally to assist hip extension, counteracting the imbalanced moment of force. Therefore, IL adopted a conservative strategy to control body balance during this period, leading to excessive stability in COP. Conversely, the forward COP speed of the PL aligned with the findings from the previous study [7]. Our results further revealed that both the PL and IL exhibited a greater range of variability in forward COP speed compared to the CL, the curves of the PL lacked a distinct peak pattern [21], possibly indicating poor dynamic stability.

The first peak vGRF in the IL was lower than that in the CL, aligning with prior research findings [34]. The results from the SPM analysis indicate that vGRF curve variability primarily resided within the initial 20% of the stance phase. In this period, the vGRF loading rate of the IL was significantly faster than that of the PL, as well as faster than that of the CL in the initial 10%. This issue may arise from an inappropriate selection of heel cushion stiffness for the prosthetic foot. Excessive softness or rigidity in the heel can result in either a delayed or hastened completion of the heel rocker (during the loading response, the foot pivots on heel until the plantar fully contact with ground) [35]. The heel cushion used by individuals with uTFA in this study may have been overly soft, leading to a decelerated loading of the vGRF during heel rocker [35]. Another possible explanation could be that individuals with uTFA do not have a clear "terminal stance phase" when walking at a relatively low speed (0.50 ± 0.11 m/s) [25]. Strictly speaking, this meant that the heel-off of IL was delayed, resulting in a reduced emphasis on push-off during the IL's terminal stance (thus lacking a second vGRF peak). Conversely, the push-off of the PL in the terminal stance also became less pronounced, and the vGRF loading rate in the loading response was slower. The inherent challenges of the single limb support phase for the PL necessitate a sudden initiation of the stance phase for the IL to maintain stability. This abrupt transition led to an increased loading rate for the IL during the loading response. Notably, previous research had reported individuals with uTFA walking at higher speeds exhibited increased loading asymmetry, highlighting the risk of osteoarthritis and musculoskeletal disorders [36].

Similar to previous studies, our research affirmed time duration variations in gait phases in individuals with uTFA while walking [37], [38]. Specifically, the PL spent more time on loading response compared to the IL. Given the demanding nature of the single limb support phase, this potentially reduced the single limb support time duration for the PL, allowing IL to share the load as much as possible [33]. This also indirectly impacted the pre-swing and swing phases, as they were inversely related to the loading response and single limb support time duration. Notably, the PL had significantly longer step time than the CL, aligning with previous research findings [39]. This might suggest individuals with uTFA adopt a relatively conservative walking pattern to enhance stability and postural control.

This study has several limitations. First, all participants wore non-standardized shoes during testing, potentially affecting plantar pressure distribution and the trajectory of the COP. However, it should be noted that walking without shoes or using standardized shoes could also alter the gait of individuals with uTFA [40]. This is because the prosthesis alignment is influenced by the effective heel height of the shoes typically worn by individuals. Second, the participants in this study were relatively old (53.92 $\pm$ 6.81 years). Future research should encompass a broader age range of individuals with uTFA to capture potential age-related variations in COP. Thirdly, specific prosthetic alignment angles were not measured in this study. Future research should investigate the correlation between alignment angle changes and COP. Fourthly, existing research suggests the reliability of the majority of gait parameters measured in healthy older adults (age: 64.8 $\pm$ 3.2 years) utilizing instrumented treadmill [23]. However, some studies propose that measurements of vGRF over an extended period and measurements taken in healthy young adults (age: 21.5 $\pm$ 2.8) might yield less reliable results [41], [42]. Although there is no similar research reported in the amputee population, it is imperative to approach the measurement results in this study with caution. Finally, the subjects opted for relatively slow walking speeds. The choice of a relatively slower walking speed may be associated with various factors, such as age, physical fitness, psychological factors, and individual interpretations of a “comfortable” walking speed, etc. And future studies should include a wider range of walking speeds.

V. Conclusion

In summary, this study focused on investigating the COP trajectory dynamics during gait in individuals with uTFA and compared to non-disabled individuals. The research uncovered distinctive gait patterns among those with uTFA, especially during the initial contact and 30% to 60% of the stance phase. These differences manifested as unique COP dynamics, involving mediolateral displacement, forward speed, and vGRF variations. Individuals with uTFA may adopt conservative abduction alignment strategies and lateral trunk lean strategies, resulting in a medial pressure bias during the initial contact and a lateral pressure bias during the first half of single limb support, respectively. Moreover, individuals with uTFA faced challenges related to balance and stability, as evidenced by variations in forward COP speed and loading rates, particularly during the loading response and single limb support phases. These findings have clinical implications, offering insights of prosthetic design, alignment procedures, and rehabilitation programs. A better understanding of the compensation strategies and unique gait dynamics of individuals with uTFA can help to optimize their gait patterns and minimize the risk of injuries due to compensatory movements.

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