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Research article

Gait asymmetrical evaluation of lower limb amputees using wearable inertial sensors

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A R T I C L E I N F O

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A B S T R A C T

This study presents an analysis and evaluation of gait asymmetry (GA) based on the temporal gait parameters identified using a portable gait event detection system, placed on the lateral side of the shank of both lower extremities of the participants. Assessment of GA was carried out with seven control subjects (CS), one transfemoral amputee (TFA) and one transtibial amputee (TTA) while walking at different speeds on overground (OG) and treadmill (TM). Gait cycle duration (GCD), stance phase duration (SPD), swing phase duration (SwPD), and the sub-phases of the gait cycle (GC) such as Loading-Response (LR), Foot-Flat (FF), and Push-Off (PO), Swing-1 (SW-1) and Swing-2 (SW-2) were evaluated. The results revealed that GCD showed less asymmetry as compared to other temporal parameters in both groups. A significant difference ($p < 0.05$) was observed between the groups for SPD and SwPD with lower limb amputees (LLA) having a longer stance and shorter swing phase for their intact side compared to their amputated side, resulting, large GA for TFA compared to CS and TTA. The findings could potentially contribute towards a better understanding of gait characteristics in LLA and provide a guide in the design and control of lower limb prosthetics/orthotics.

1. Introduction

Human gait is a complex process that exhibits variability not only between individuals, but also within the same individual across consecutive steps while performing activities of daily living (ADLs) [1]. It has been reported in Ref. [2], that the repetitive nature of human gait is not seamless, even on a very flat surface. This can be ascribed to functional variations in each lower limb's contribution

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to propulsion and control, leading to gait asymmetry (GA) during walking. GA represents a deviation from symmetrical walking patterns and is particularly prevalent among lower limb amputees (LLA), who face challenges in achieving symmetric gait. GA holds significance not only for understanding walking abnormalities but also for its implications in the design and control of prosthetics and orthotics [3], thus ensures that amputees can enjoy improved mobility and reduced discomfort. Moreover, the importance of evaluating GA extends beyond amputees. It is crucial for understanding, treating and rehabilitation of various neurological disorders or medical conditions such as post-orthopedic surgery, falls, stroke and Parkinson's [4,5]. GA can be evaluated through various gait parameters including but not limited to stance time, swing time, stride time, stride length, step length, ground reaction forces, single limb support, double limb support and center of pressure.

GA or symmetry index (SI) is, therefore, commonly used as a key assessment procedure for gait analysis in both clinical and rehabilitation paradigms [6,7]. LLA, during the stance phase, spend more time on their sound (intact) side while less time on their amputated side during the swing phase, and therefore, exhibits GA [8]. Gait asymmetrical patterns in these individuals may arise due to a decrease in muscle volume and force, pain at the stump-socket interface, and less confidence in the prosthesis [8,9], and hence LLA consume more metabolic cost. GA can also lead to numerous further concerns including joint degeneration and osteoarthritis of the sound side, pain in the lower back and joints [9].

One of the essential conditions of assistive devices to perform optimally is the reliability and accuracy of augmented sensory feedback while measuring gait-related parameters. Augmentation of sensory feedback systems can provide effective aid to traditional physiotherapy in the rehabilitation of gait asymmetries [10]. There are many commercial devices such as motion capture labs, floor mats, instrumented treadmills, insole sensor devices that have been used for gait analysis including the GA [11]. Generally, these systems are lab-based, quite expensive, and evaluate the data offline, thus unsuitable for clinical use in general. On the contrary, wearable sensors such as Inertial Measurement Unit (IMU), being a low-cost, low power, easy to don and doff and require low maintenance have been effectively used for objective assessment of GA [12–16]. These wearable devices are capable of estimating spatio-temporal parameters, fatigue prediction, activity recognition and to further evaluate GA, proves to be the most appropriate choice for LLA to be used both in the clinical environment and at home [17–19].

1.1. Related work

Isakov et al. [1] investigated the influence of speed on gait symmetry considering spatio-temporal parameters of 11 below knee amputees using conductive rubber walkway and electro-goniometers. The authors reported that symmetry of knee angle during the loading phase and toe off, and spatio-temporal parameters both were greatly affected by the speed of gait. Blazkiewicz et al. [2] evaluated the GA of spatio-temporal parameters of 58 control subjects by comparing four methods using the ZEBRIS platform. SI for stance, swing and loading phase was reported as 2.38 %, 4 %, and 9.7 % respectively. Gouwanda et al. [6] proposed the use of a gyroscope at the thigh and the shank of each lower extremity to identify GA with the normal walk at first and by imitating the gait of the patient in later trials. Results showed that normalized SI was found to be 20 % greater in asymmetrical gait compared to SI in normal gait during pre and initial swing. Nolan et al. [10] investigated the effect of speed on loading and temporal asymmetry in 4 TFA and 4 TTA. Kinetic data were acquired using CDG (Infotronic) force shoes at 50 Hz while performing level ground walking activities at 0.5, 0.9, 1.2 m/s and at their maximum speed. The authors reported the increase in loading asymmetry with the speed increase however, temporal asymmetry was reduced.

Anwary et al. [3] developed an android app to acquire real-time IMU data from lower extremities to assess gait asymmetry. 10 young and 10 older subjects took part in this study while performing level ground walking and turning. Spatio-temporal parameters such as stride duration, step duration, cadence, step ratio, stance and swing, total distance and velocity were assessed. The validation was carried out with Motion Capture System (Qualisys). The results showed promising results in terms of detection accuracy of the parameters and exhibited high confidence interval. The authors concluded that gait asymmetry can be evaluated using wearable sensors and expensive laboratory based setup can be avoided. Clemens et al. [4] investigated gait symmetry and repeatability of unilateral lower limb amputees using gyroscope placed on thigh and shank of both limbs. Angular velocity signals were acquired through gyroscope and processed using dynamic time warping. The authors concluded that TFA exhibited more segmental asymmetries and less repeatable movements compared to TTA ($p < 0.004$). Han et al. [12] investigated the changes in GA and bilateral coordination in relation to gait velocity using IMUs placed on the shoe outsoles of both sides. Eighty young healthy adults took part in this study walking performing walking trials at various speeds on a treadmill. The analysed parameters affecting GA phase coordination index included cadence, swing time, step length, body mass index and gait velocity. It was reported that GA and bilateral coordination improved during fast walking compared to the other modes of speeds.

Valle et al. [14] investigated the effects of gait and pelvis asymmetries using a single wearable sensor placed at the back. Kinematic data from IMU were captured while walking on a 10-m walkway and using Timed Up and Go (TUG) test. Symmetry index based on antero-posterior acceleration signals of both limbs and 3-D pelvis motions were evaluated for both amputees and CS. Further the duration of TUG test, time of each subcomponent and the velocity of the turning were evaluated. The authors concluded that overall, gait and pelvis asymmetries had a damaging effect on the performance while turning compared to linear walking. Wang et al. [15] proposed a geometric method for estimation of step length and spatial GA based on four IMUs attached to the lower limbs. Yang et al. [16] developed the low-cost lower extremity ambulatory feedback system (LEAFS) to acquire kinetic data. The authors reported an improvement in GA of two TTA whereas the third TTA did not show any sign of improvement. Another study [19] evaluated regularity and gait symmetry in healthy participants ($n = 10$) and TFA ($n = 10$) using foot insoles and a single accelerometer at thorax. The authors reported the utility of a single accelerometer by considering its autocorrelation features, and found adequate for GA assessment. Martini et al. [20], in a recent study, proposed the utility of vibrotactile feedback to improve the temporal GA in LLA. Three TFA took part in this study while performing several walking activities with and without the sensory feedback. The experimental results showed higher SI when using haptic feedback compared to the natural walk. Most of the studies found in the literature reported the GA evaluation for stance, swing, single limb and double limb support in general using motion capture systems, sEMG and/or kinetic measurement devices [1,2,9,11]. Few studies [6,14,17–19] conducted with wearable sensors such as IMU, however; the studies were either confined to control subjects (CS) or limited gait parameters without considering the inner phases of gait were assessed. Table 1 shows a summary of related work.

To the best of the authors' knowledge, this is the first study evaluating the GA for the inner-stance and swing phases while placing IMU at the shank of CS and LLA. By analyzing and evaluation of GA of temporal gait parameters based on events/phases using a single IMU with a focus on the effects of different surfaces (overground and treadmill) and varying speeds, this study contributes to a comprehensive analysis of how GA manifests in various conditions for both control and lower limb amputee participants. The aims of the current study are, therefore.

- To evaluate gait asymmetry of inner stance and swing phases based on the temporal gait parameters using a low-cost portable gait measuring sensory system
- To analyse statistically the inter-subject and intra-subject assessment of the symmetry index for CS and LLA during level-ground walking (overground and treadmill)

2. Materials and methods

2.1. Subjects

Seven young male CS (age: 30.6 ± 2.7 years; mass: 71 ± 9.8 kg; height: 171.7 ± 5.3 cm) participated in this study. They were healthy and without any physical or cognitive abnormalities. Two male unilateral LLA with one above-knee amputation (transfemoral amputee (TFA)) and one below-knee amputation (transtibial amputee (TTA)) also participated. Both amputees were without any neurological or orthopaedic disorder except their amputation. Both were left amputated, took part without the use of an ambulation aid and wore normal daily shoes. TFA participated twice with the same prosthetic knee, however, with a different prosthetic ankle/foot. Table 2 shows the demographic data of all the participants and Table 3 shows further details of LLA. All the participants were informed about the objectives of the study and the experimental activities, and informed consent were received from them before performing the trials. All the participants consented to have their images published for research purposes. The experimental procedures carried out in this study were approved by the University of Leeds Ethics Committee (Ref: MEEC 14-011).

2.2. System description

An IMU (MPU 6050, InvenSense Inc.) based on MEMS (Micro Electro Mechanical Systems) technology was used in this study on

Table 1
Summary of related literature work.

Ref.	Sensory system details and placement	Assessed Gait Metrics	Activities	Remarks/Outcome
	GIGA-system (Lab Based), goniometers, metallic strips underneath foot	Walking speed, step rate, stride length, stride duration, stance duration, swing duration, double limb support time and joint rotations	Level ground walking with comfortable and rapid speeds	At comfortable speed, amputees walked 29 % slow compared to control subjects.
	Conductive rubber walkway and eletro-goniometers at hip and knee	Stance, Swing, step time, double limb support and step length. Hip and Knee joint angles	Level ground walking at self-selected speed and fast speed	Apart from level of symmetry, spatio-temporal parameters were significantly affected at fast speed
	ZEBRIS platform having 17,024 sensors	Symmetry index (SI) for stance, swing duration and loading phase	Level ground walking	SI: Stance: 2.38 % Swing: 4 % Loading Phase: 9.7 %
	Gyroscope at thigh and shank	SI based on knee angle	Level ground walking at normal speed and with load on left limb	Normalized SI was 20 % greater in asymmetrical gait than normalized SI in normal gait during initial and pre swing
	Kistler force platform with Infotronic CDG software	Stance, swing duration based on Vertical ground reaction forces, step time	Level ground walking at three different speed	Loading asymmetry increased at fast speed while temporal asymmetry reduced
	2 IMUs at thigh and shank of both lower limbs	Step length	Level ground walking at slow, normal and fast speeds.	For CS and patients with gait impairment, GA for step length estimation was 3 % and 14.7 % respectively.
	2 IMUs at each thigh and shank segment	Stance duration, step length, Double Limb Support, segmental angular velocities	Level ground walking at self-selected speed	TFA exhibited more asymmetries and less repeatable movements compared to TTA
	1 IMU at back	Pelvis and Gait asymmetries	Sit-to-stand, walking and turning	Both asymmetries are appropriate during linear walking but can be harmful during turning
	2 IMUs at each shoe outsole	Step length, gait speed, body mass index, cadence, swing time, GA	Treadmill walking at slow, normal and fast speed	GA and bilateral coordination improved at fast pace.

Table 2
Subject's demographic information.

Subject	Age (years)	Weight (Kg)	Height (cm)	Gender	Health status
A	29	83	175	M	Able bodied
B	32	63.1	166.2	M	Able bodied
C	28	70	176	M	Able bodied
D	29	64	167.6	M	Able bodied
E	36	87.5	180	M	Able bodied
F	29	65	170.9	M	Able bodied
G	31	67	166.6	M	Able bodied
H	77	70	185.1	M	Transtibial Amputee
I	54	65.7	166.1	M	Transfemoral Amputee

Table 3
Details of lower limb amputees.

Sub.	Prosthetic Knee	Prosthetic Ankle/foot	Etiology	Year of Amputation
TFA	3R80 Ottobock	Panthera CF II Medi (TFA-P) Kinterra Freedom Innovation (TFA-K)	Trauma (Chronic infection on the knee)	2009
TTA	–	Panthera CF I Medi	Trauma (Road traffic accident)	2004

each lower limb extremity. MPU 6050 was in the form of a PCB breakout with a size of 21 × 16 × 0.2 mm and weight of about 4 g. It has six degrees of freedom, comprising three-axis accelerometer and a three-axis gyroscope embedded inside a single chip. The IMU was mounted on the acrylic fixture, which then strapped on the lateral side of the shank with the help of a Velcro. The IMU was connected to the PCB using IDC header/socket (8284). Base unit consists of a 3D printed base, a PCB which integrates a wireless microcontroller, electronic components and a power unit. The PCB has a size of 60 mm × 60 mm and was designed as double-sided prints in Eagle software (v6.50, Cadsoft, USA). The communication and data acquisition is handled by a tiny (33 mm × 23 mm) microcontroller (Moteino R4, LowPowerLab LLC, USA), which has an integrated 433 MHz radio frequency (RF) RFM12B transceiver. This feature allows communication and data transfer to be done wirelessly, which is essential for portable application. The power unit consists of 2 rechargeable AA battery (3V supply) which is then regulated to 5V for the PCB operation using a 5V boost voltage regulator (NCP 1402, Pololu Corporation, USA). Fig. 1 shows the ambulatory systems placed on the lower extremities for acquiring the data from IMU and

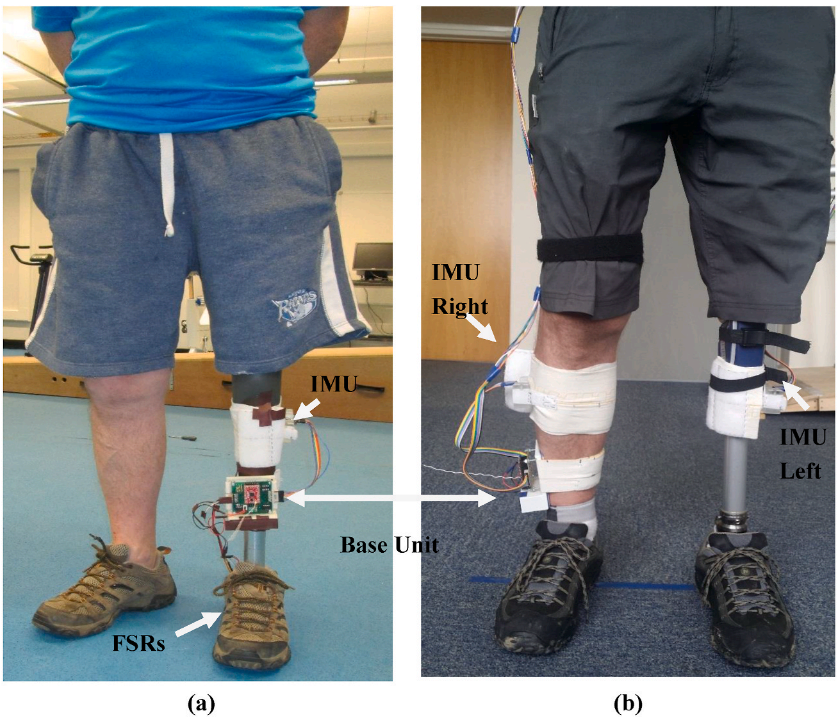

Fig. 1. The ambulatory systems mounted on lower limb amputees (a) IMU, Base unit and Foot Sensitive Resistors (FSRs) and (b) IMUs on both limbs and a Base unit.

FSRs where the later were used for validation of gait events in our previous studies [21,22], where Fig. 1 (a) shows the placement of ambulatory system on an amputated side whereas Fig. 1 (b) shows the system placed on both limbs for GA evaluation.

2.3. Experimental protocol

A single IMU comprising a tri-axial gyroscope (range ± 500 deg/s) and a tri-axial accelerometer (range ± 4g) was placed on the lateral side of the shank of both limbs along with a microcontroller, wireless module and other circuitry as shown in Fig. 2. IMUs were connected to the microcontroller through I2C (inter-integrated circuit). The angular velocity of the shank and the linear acceleration along the longitudinal axis in the sagittal plane were acquired from both IMUs at a sampling rate of 100 Hz. The acquired data from both IMUs were then fed to a microcontroller where temporal gait events were detected in real-time.

All the participants were provided enough time to familiarize themselves with the system. Once acquainted, they were asked to walk along a straight path of 7.2-m walkway with three variations (slow, normal and fast). Each activity was repeated five times for all the participants. The walking speed of each trial was calculated using a stopwatch and then averaged. The average walking speed was 0.89, 1.16 and 1.50 m/s for CS, 0.76, 1.04 and 1.40 m/s for TFA and 0.9, 1.1 and 1.37 m/s for TTA during slow, normal and fast walk respectively. Level ground walking activities were also performed on an instrumented treadmill in addition to overground. The average walking speed obtained during overground walking was set on the treadmill (i.e. the nearest value in km/h). TFA with 3R80 prosthetic knee and Kinterra prosthetic ankle/foot (TFA-K) took part in the treadmill trials. The LLA were provided with the safety harness during the trials on the treadmill. All the participants were given about 2 min to become acquainted with walking on the treadmill. A single trial for each slow, normal and fast walk was carried out for all the participants. Fig. 3 (a) to Fig. 3(c) shows the pictures of TFA, TTA and one of the CS while performing overground (OG) walking and Fig. 3 (d) to Fig. 3 (f) shows the pictures of TFA, TTA and one of the CS while performing treadmill (TM) walking.

2.4. Description of Gait parameters and gait asymmetry

Five gait phases at each gait cycle were extracted based on the temporal gait events identified from the kinematic data using an IMU at the shank of each limb. The gait events have been successfully identified and validated in our previous work [21,22]. Fig. 4 shows the GC segmentation into its subsequent sub-phases whereas Fig. 5 illustrates the identification of gait events/phases using the acceleration and angular rate signal of an IMU. The stance phase was segmented into three sub-phases namely Loading-Response (LR), Foot-Flat (FF) and Push-Off (PO) whereas the swing phase was segmented into two sub-phases namely Swing-1 (SW-1) and Swing-2 (SW-2). Based on the temporal gait events detection, the duration of subsequent phases of the GC was computed. Stance phase duration (SPD) was computed using Equation (1).

$$SPD_i = TO_i - IC_i \tag{1}$$

where i indicates the occurrence of the temporal events within the stride i detected from the IMU. Similarly, the duration of LR, FF and PO were computed using Equations (2)–(4). Swing phase duration (SwPD) was computed using Equation (5).

$$LR_i = FFS_i - IC_i \tag{2}$$

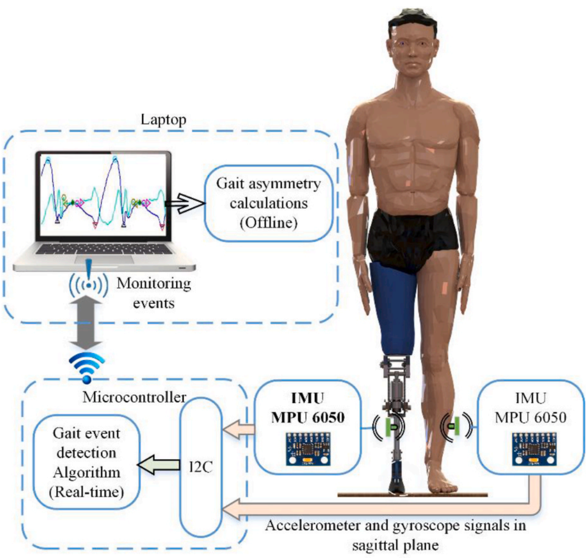

Fig. 2. Experimental setup.

Fig. 3. Participant (a) TFA, (b) TTA and (c) CS-07 during OG walking and (d) TFA, (e) TTA and (f) CS-05 during TM walking.

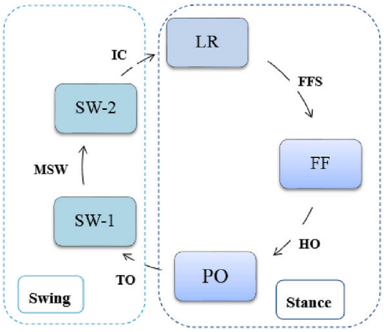

Fig. 4. Segmentation of GC, depicting five gait phases and their transitions.

$$FF_i = HO_i - FFS_i \tag{3}$$

$$PO_i = TO_i - HO_i \tag{4}$$

$$SwPD_i = IC_{i+1} - TO_i \tag{5}$$

where IC_{i+1} is the occurrence of an event in the next stride whereas TO_i was the event in the previous GC. Similarly, SW.1 and SW.2 were computed using Equations (6) and (7).

$$SW.1_i = MSW_{i+1} - TO_i \tag{6}$$

$$SW.2_i = IC_{i+1} - MSW_{i+1} \tag{7}$$

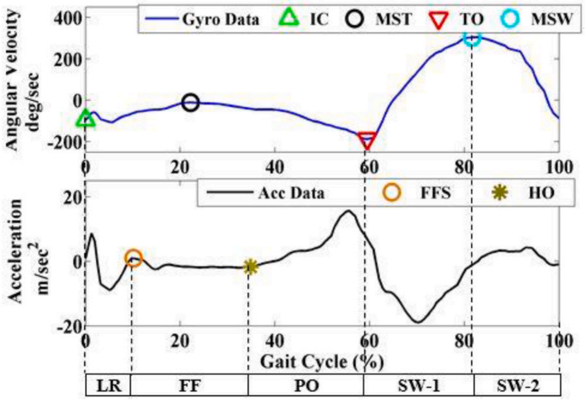

Fig. 5. Description of Gait events and phases. IC: Initial Contact, MST: Mid-Stance, TO: Toe Off, MSW: Mid-Swing, FFS: Foot-flat Start, HO: Heel Off, LR: Loading Response, FF: Foot-Flat, PO: Push Off, SW-1: Swing-1, SW-2: Swing-2.

Finally, the stride duration (StD) or Gait cycle duration (GCD) was computed from two consecutive strides using Equation (8).

$$StD = IC_{i+1} - IC_i \tag{8}$$

GA is often described as an indicator of pathological or abnormal walking and is considered a useful clinical issue that needs to be addressed [12]. Different indicators have been used for the assessment of GA or symmetry index (SI) in the literature [2,9,12]. Each indicator has its own merits and demerits, therefore, it is difficult to rely on any specific indicator to assess symmetry. The authors in Ref. [2] evaluated the symmetry of spatio-temporal gait parameters using four different indicators namely, ratio-index, symmetry index, gait asymmetry and symmetry angle. Based on their analysis, SI was recommended as the most sensitive assessment of gait symmetry analysis on the basis of gait parameters evaluated with 58 healthy subjects [2]. The current study used the same indicator, however, without considering absolute values in the nominator to assess the limb dominance. GA or SI was computed using Equation (9) as reported in.

$$SI = \frac{X_R - X_L}{0.5 \times (X_R + X_L)} \times 100\% \tag{9}$$

where X_R and X_L are the metrics for dominant and non-dominant limbs of the CS. Similarly for the amputees, SI was calculated using Equation (10).

$$SI = \frac{X_I - X_P}{0.5 \times (X_I + X_P)} \times 100\% \tag{10}$$

where X_I and X_P shows the metrics for intact and prosthetic limbs respectively. The value of SI = 0 indicates full symmetry.

2.5. Data collection and analysis

All the participants completed five trials for each activity. As the number of strides for each trial varied between the subjects, therefore, for analysis purposes, an equal number of samples for each activity across all the subjects were considered to avoid any bias. 18, 16, and 14 samples of each gait parameter were considered for slow, normal and fast walk respectively during overground and treadmill walking and then averaged overall. Four hundred and eighty samples of each gait parameter from all the subjects including CS ($48 \times 7 = 336$), TFA (participated twice, $48 \times 2 = 96$) and TTA ($48 \times 1 = 48$) during OG walking and four hundred and thirty-two samples of each gait parameter from both groups during TM walking were analysed. TFA-P participated only for OG walking whereas TFA-K participated for both OG and TM walking. To eliminate acceleration and/or deceleration effect at the start/stop of each trial, only steady-state data were included in the analysis.

The evaluation of SI was carried out based on: 1) the timing duration of the gait phases, 2) the timing duration of each phase as a percentage of the GC. The mean (M) and standard deviation (SD) of SI for all gait parameters were then calculated for all the activities in both groups. The coefficient of variation (CV) was also computed using Equation (11).

$$CV = \frac{SD}{M} \times 100\% \tag{11}$$

Statistical analyses were performed using IBM SPSS Statistics 21 (SPSS, Inc., Chicago, IL, USA). Data were assessed for normality using the Shapiro-Wilk test prior to analysis. In addition, normality was assessed using skewness or kurtosis. The author in Ref. [23] reported that for small samples ($n < 50$), if the z-score lies within ± 1.96 for either skewness or kurtosis, indicating an alpha level 0.05 (95 % confidence interval), then accept the null hypothesis. Based on the z-score, the data were considered as normally distributed. The *t*-test (assuming equal variance) and analysis of variance (ANOVA) are robust enough to moderate small deviation from normality [24]. Data were assessed using two-tailed independent samples *t*-test to determine the significance between the control and amputee groups. In addition, paired sample *t*-test was also carried out to determine the significant differences between OG and TM for the same dependent variables. Results were considered as statistically significant at $p < 0.05$. For statistical analysis, SI data based on the timing duration of gait phases for overall LGW on OG and TM walking were assessed.

3. Results

3.1. *SI based on the timing duration of the gait phases*

The evaluation of the SI based on the timing duration of temporal gait phases in terms of mean (M) and standard deviation (SD) all expressed in percentage (%) for all the activities and all the participants are shown in Table 4. Positive and negative values of SI indicate that the higher value of the temporal variable was recorded on the right limb (intact side in case of amputees) and left limb (prosthetic side in case of amputees) respectively. Table 4 shows the mean, standard deviation and CV values of gait cycle duration (GCD) of all the participants during OG walking at slow, normal and fast speed. Stride duration or GCD is decreased in both groups as the speed is increased. There was no significant difference in CV for CS during three locomotion modes, however, LLA, in general, showed a reduction in CV with an increase in speed with few exceptions as shown in Table 4.

Fig. 6 (a) to Fig. 6(h) shows the distribution of SI values of temporal gait parameters for CS, TFA and TTA during a level ground walk on OG and TM. The lowest SI was observed for StD (less than ±5 %) for all the CS during OG and TM walking as shown in Fig. 6 (a). The SI values for SPD and SwPD were also found to be less than 15 % for all the CS as shown in Fig. 6 (b) to Fig. 6(c). The inner phases of stance and swing, however, showed higher SI values for all the CS during OG and TM walking as shown in Fig. 6 (d) to Fig. (h). It was also observed that SI values of StD, SPD, SwPD and PO during TM walking were lower than the OG walking, however, for LR, FF, SW-1 and SW-2, variation was observed between the locomotion modes for OG and TM.

Similar to the CS, the lowest SI was observed for StD (i.e. (less than ±5 %) followed by the SPD where the SI lies less than 25 % across both amputees during OG and TM walking as shown in Fig. 6 (a) to Fig. 6 (b). SI values for SwPD and SW-1 also lie in a range of ±30 %, however, LR, FF, PO and SW-2 showed a high mean (SD) in both groups during OG and TM walking. It was also observed that there is a slight improvement in SI for StD, SPD and SwPD for both amputees during TM walking compared to OG walking. The rest of the gait parameters showed variation in results between OG and TM walking. Overall, a significant improvement was observed in SD of all the gait parameters during TM walking compared to OG. Overall, CS has the lowest SI distribution for all the gait parameters compared to TFA and TTA during OG and TM walking. Between TFA-P and TFA-K there was not much significant difference in SI for StD and SW-2. TFA-P showed lower SI values for SPD, SwPD, LR and PO whereas TFA-K showed lower SI for FF and SW-1 Results also showed significant improvement for TFA-K during TM walking for all parameters except for LR which showed high SI in case of TM walking.

The mean for the temporal gait phases in terms of percentage of GC during slow, normal and fast walk on OG respectively are presented in Fig. 7 (a) to Fig. 7(c). It was observed that with increase in speed i.e. from slow to normal and then normal to fast, percentage (PT) of SPD decreased whereas PT of SwPD increased in both groups (CS and amputees) as shown in Fig. 7 (a) to Fig. 7(c). PT of inner-stance phases varies largely with speeds compared to inner-swing phases. There was no significant variation in the mean PT of LR for CS with the increase in speed. However, LLA showed large variation between the three locomotion modes for instance, the mean PT of LR varies from 5.8 to 6.9 between slow to normal and then 6.9 to 12.2 between normal to fast speed for the intact side of TFA-P as shown in Fig. 7 (a) to Fig. 7(c). The mean PT of FF for CS showed about 2 % reduction with the increase in speed from slow to fast. Large variation was observed for LLA especially for the intact side of TFA-P and prosthetic sides of TFA-K and TTA. The mean PT of PO showed in general a reduction when the speed varies from slow to normal and normal to fast in both groups except for TTA where the mean PT of PO increased slightly at first from slow to normal and then reduced at fast speed. It was also showed that both TFA and TTA tend to spend more time on the intact side during PO compared to the prosthetic side irrespective of the variation in speed as TTA showed low SI values for all the parameters in general during both OG and TM walking compared to TFA-K with the exception in FF during TM walking where SI for TFA-K was found lower than TTA.

Table 4
M(s), SD (s) and CV (%) values of GCD in CS, TFA and TTA during Overground Walking (OG), M: Mean, SD: Standard Deviation, CV: Coefficient of Variation, R: Right, L: Left, I: Intact, P: Prosthetic.

Mode	V	CS		TFA-P		TFA-K		TTA	
		R	L	I	P	I	P	I	P
Slow	M	1.33	1.33	1.4	1.38	1.45	1.43	1.53	1.53
	SD	0.14	0.14	0.05	0.06	0.064	0.072	0.085	0.085
	CV	10.5	10.5	3.6	4.3	4.4	5	5.5	5.5
Normal	M	1.14	1.14	1.2	1.2	1.17	1.17	1.25	1.26
	SD	0.12	0.12	0.037	0.033	0.046	0.048	0.047	0.083
	CV	10.5	10.5	3	2.75	3.9	4.1	3.76	6.5
Fast	M	1.03	1.03	1.06	1.08	1.1	1.07	1.11	1.1
	SD	0.102	0.108	0.026	0.038	0.022	0.027	0.067	0.034
	CV	9.9	10.5	2.5	3.5	2	2.5	6	3

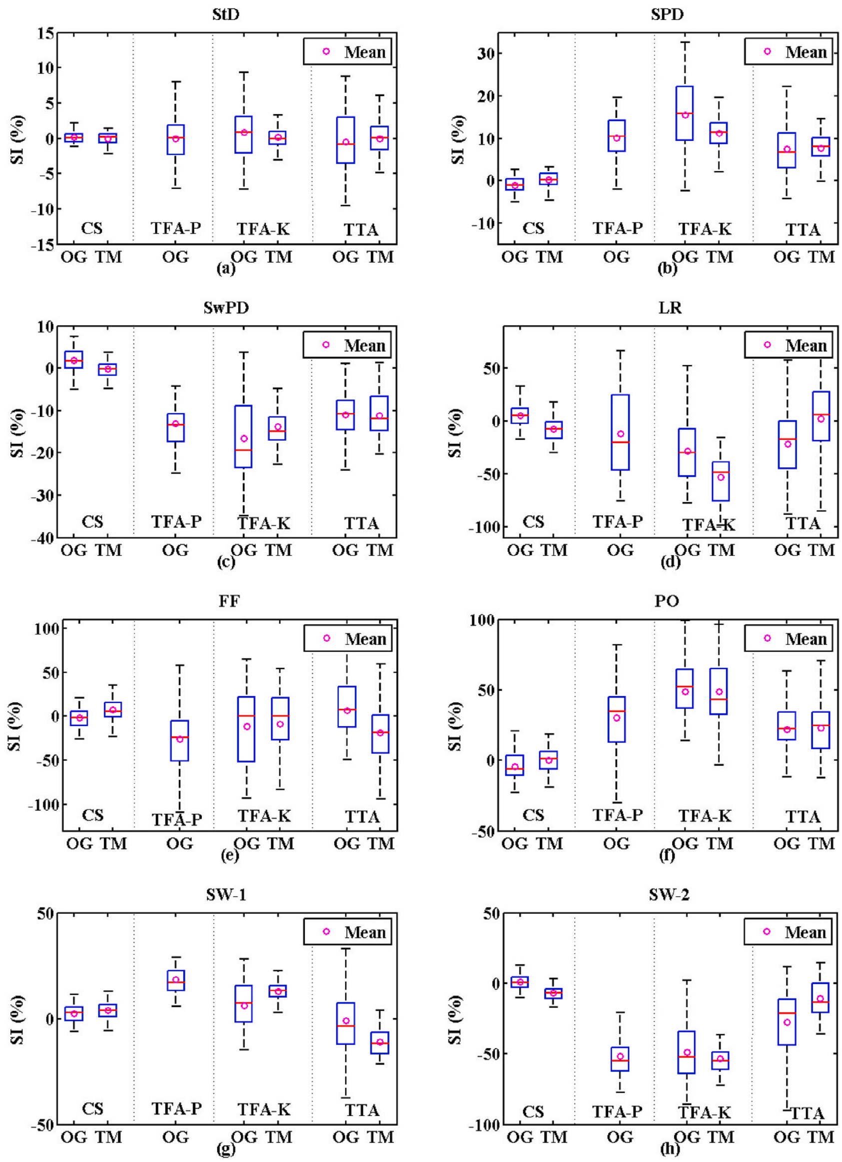

Fig. 6. SI (%) distribution of temporal gait parameters (a) StD (b) SPD (c) SwPD (d) LR (e) FF (f) PO (g) SW-1 (h) SW-2 for CS, TFA and TTA during OG and treadmill walking.

shown in Fig. 7 (a) to Fig. 7(c). The mean PT of SW-1 showed a slight increment in the values when speed varies from slow to fast except for one instance (TTA prosthetic side) where it was decreased about 0.2 % and 0.3 % from slow and normal speed. The mean PT of SW-2 showed a variation in the trend of increment/decrement between the three locomotion modes in both groups as shown in Fig. 7 (a) to Fig. 7(c).

3.2. SI based on the percentage of the gait phases

The mean, SD and CV values for the temporal gait parameters during OG and TM walking respectively are presented in Table 5. GCD were found almost identical in both lower limbs across all the participants during overall LGW on OG and TM. TFA-K showed high CV compared to CS, TFA-P and TTA during OG walking whereas CV was higher for TTA during TM walking. It was also observed that CV for GCD reduced in TM walking for TFA and TTA however, no significant difference was observed in case of CS. The CV for SPD and SwPD were less than 10 % across all the participants in both locomotion terrains (OG and TM). High variability was observed in case of inner-phases of stance and swing for all the participants in particular for stance inner-phases.

The smallest CV was found to be 2.4 % and 1.7 % in case of SPD for the intact side of TFA-P during OG walking and intact side of TTA during TM walking respectively. The largest CV was found to be 43.7 % and 33.7 % in case of LR for the intact side of TFA-P during

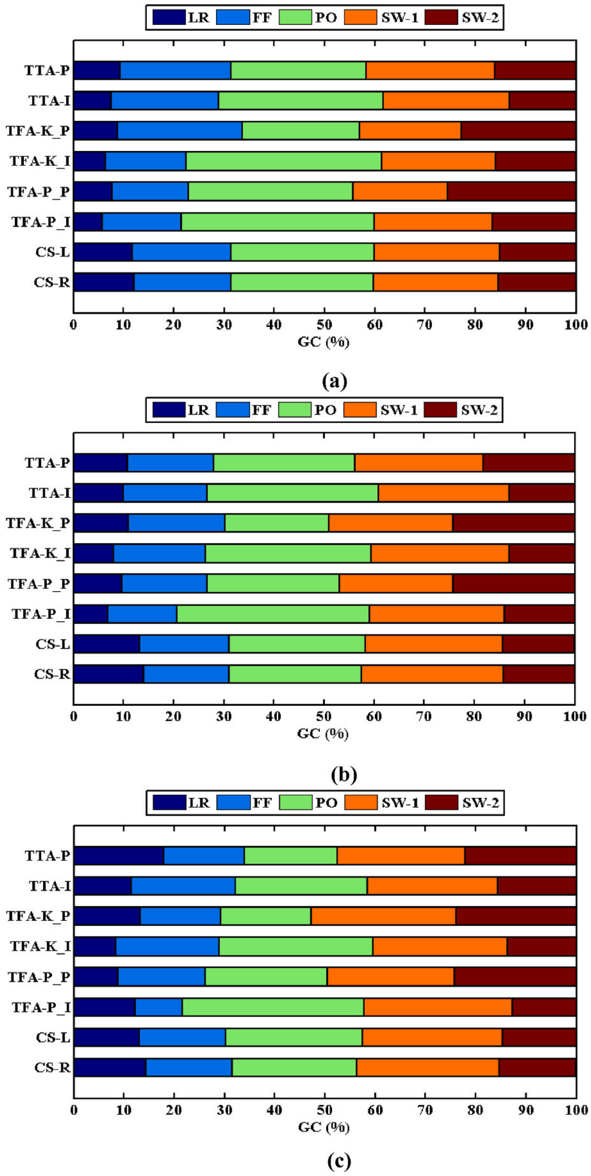

Fig. 7. Mean percentage distribution of inner-phases of stance and swing during OG walk at (a) slow (b) normal and (c) fast speed.

OG walking and prosthetic side of TFA-K during TM walking respectively as shown in Table 5. Since the CS were more in numbers (seven in this study) compared to LLA, SD was expected to be a high and hence large CV. Overall, TFA showed large asymmetry for all the inner gait phases compared to CS and TTA during OG and TM walking as shown in Table 5. At only one instance, it was observed (i.e. during TM walking) that asymmetry of FF was large for TTA compared to TFA.

The mean stance phase was 58 % and 58.7 % for right and left limb respectively of CS during OG walking showing more time spent on the left limb during the stance with mean SI of −1.1 % as shown in Table 5. The mean swing phase was 42 % and 41.3 % for right and left limb of CS with mean of 1.64 % during OG walking. TFA-P showed a mean stance phase of 59 % and 53.3 % for intact and prosthetic limbs with mean SI of 10.3 % whereas TFA-K showed a mean stance phase of 60.2 % and 52.1 % with mean SI of 14.7 % during OG walking. The mean swing phase for TFA-P was 41 % and 46.7 % for intact and prosthetic limb respectively with mean SI of −12.9 % whereas TFA-K showed a mean swing phase of 39.8 % and 47.9 % with mean SI of −17.9 % during OG walking. The mean stance phase for TTA intact and prosthetic side was 60.5 % and 55.9 % respectively, with mean SI of 7.5 % and the mean swing phase for the intact and prosthetic side was 39.5 % and 44.1 % respectively with mean SI of −10.9 % during OG walking as shown in Table 5.

Table 5
M (s), SD (s), and CV (%) values of temporal GP in terms of percentage of GC for CS, TFA and TTA during OG and TM Walking. GP: Gait Parameters, V: Variable, M: Mean, SD: Standard deviation, CV: Coefficient of Variation, R: Right, L: Left, I: Intact, P: Prosthetic.

GP	V	OG								TM					
		CS		TFA-P		TFA-K		TTA		CS		TFA-K		TTA	
		R	L	I	P	I	P	I	P	R	L	I	P	I	P
GCD (s)	M	1.18	1.18	1.2	1.2	1.2	1.2	1.32	1.32	1.13	1.13	1.15	1.15	1.43	1.43
	SD	0.17	0.17	0.14	0.13	0.19	0.19	0.19	0.18	0.12	0.12	0.1	0.1	0.16	0.16
	CV	10.5	14.6	11.6	10.8	15.8	15.8	14.4	13.6	10.6	10.7	8.7	9	11.3	11.4
SPD (%)	M	58	58.7	59	53.3	60.2	52.1	60.5	55.9	57	57	59.2	53	62	57.4
	SD	3.2	3.4	1.4	2.6	1.6	4.8	1.9	3.1	2.9	3.4	2.2	2.9	1.1	2.4
	CV	5.5	5.8	2.4	4.9	2.6	9.2	3.1	5.5	5	5.9	3.7	5.5	1.7	4.2
SwPD (%)	M	42	41.3	41	46.7	39.8	47.9	39.5	44.1	43	43	40.8	47	38	42.6
	SD	3.2	3.4	1.4	2.6	1.6	4.8	1.9	3.1	2.9	3.4	2.2	2.9	1.1	2.4
	CV	7.6	8.2	3.4	5.5	4	10	4.8	7	6.7	7.9	5.4	6.2	2.9	5.6
LR (%)	M	13.4	12.6	8	8.7	7.5	10.8	9.5	12.4	14.5	15.3	8.4	15.4	7.8	7.7
	SD	4.4	4.1	3.5	2.2	1.2	3.4	2.7	5.4	4.4	4.5	1.3	5.2	1.6	2.4
	CV	32.8	32.5	43.7	25.3	16	31.5	28.4	43.5	30.3	29.4	15.5	33.7	20.5	31.2
FF (%)	M	18	18.4	13.2	16.4	18.2	20.5	19.7	18.6	15.5	14.2	13.3	14.8	14.7	18.2
	SD	5.1	5.4	5	3.7	5.8	6.3	4.3	5.5	3.9	3.2	2.8	4	3.5	5.4
	CV	28.3	29.3	37.9	22.6	31.8	30.7	21.8	29.5	25.2	22.5	21	27	23.8	29.6
PO (%)	M	26.6	27.7	37.8	28.2	34.5	20.8	31.3	24.9	27	27.5	37.5	22.9	39.5	31.5
	SD	6.4	5.9	3.9	6	6.4	5.4	5.3	6.8	6.3	6.5	4.4	6.2	4.4	5.9
	CV	24	21.3	10.3	21.3	18.5	25.9	16.9	27.3	23.3	23.6	11.7	27	11.1	18.7
SW-1 (%)	M	27	26.6	26.4	22	25.5	24.3	25.7	25.6	29.3	28.2	27.9	24.6	24.3	27.1
	SD	3.4	3.9	2.8	3.1	3.3	4	2.4	3.4	2.6	3.7	2.5	3.4	1.3	2
	CV	12.6	14.6	10.6	14.1	12.9	16.5	9.3	13.3	8.9	13.3	8.9	13.8	5.3	7.4
SW-2 (%)	M	15	14.7	14.6	24.7	14.3	23.6	13.8	18.5	13.7	14.8	12.9	22.3	13.7	15.5
	SD	2.9	2.4	2.2	2.3	2.7	2.5	2.3	4.7	1.8	3	1.2	4	0.7	1.9
	CV	19.3	16.3	15	9.3	18.8	10.6	16.6	25.4	13.1	20.3	9.3	17.9	5.1	12.3

The mean SI values for SPD and SwPD were observed low during TM walking for CS and TTA whereas TFA showed an increase in SI values when compared to OG walking as shown in Table 6. In general, the inner phases of stance and swing showed high variation in SI across all the participants during OG and TM walking. Overall, the lowest SI range was observed during the stance phase in both groups and in both terrains (OG and TM) whereas the highest range was observed for PO and LR in case of TFA-K during OG and TM walking respectively as shown in Table 6. TFA-K showed an improvement in SI during SPD, SwPD and FF during TM walking whereas TTA showed an improvement during FF and SW-2. There was no significant difference during SPD and SwPD for TTA in OG and TM walking. However, mean SD became less during TM walking in general for all the gait phases in both groups with few exceptions as shown in Table 6.

Table 6
MEAN (SD) % values of SI for temporal GP as percentage of GC during LGM on OG and TM.

	OG				TM		
Parameter	CS	TFA-P	TFA-K	TTA	CS	TFA-K	TTA
SPD	−1.1 (5.8)	10.3 (4.2)	14.7 (8.3)	7.9 (4.5)	0.34 (5.8)	10.9 (3.4)	7.7 (3.2)
SwPD	1.64 (8.4)	−12.9 (5)	−17.9 (9.2)	−11 (5.6)	−0.3 (7.9)	−13.8 (4)	−11.2 (4.3)
LR	5.4 (39.5)	−11.8 (43.1)	−31.6 (30.9)	−20.8 (36.4)	−5.94 (32.6)	−53.5 (25.9)	2.5 (34)
FF	−2.1 (34.8)	−25.9 (45.2)	−12.1 (46)	7.1 (32.7)	7.9 (29.7)	−9.6 (36.4)	−18.4 (35.6)
PO	−4.1 (26)	30.5 (24.1)	50.3 (25.5)	24.5 (28.9)	−0.72 (27)	50.8 (26.2)	23.4 (19)
SW-1	1.9 (12.6)	18.4 (8.3)	5.3 (13.8)	0.6 (19.3)	4 (13)	12.4 (5.4)	−11 (6.5)
SW-2	1.16 (19.7)	−51.6 (13.4)	−49.4 (20.2)	−27 (25.8)	−7.2 (15.7)	−52.9 (10.8)	−10.9 (12.7)

Table 7
Assessment of Mean SI between Control and Amputee Groups during OG and TM Walking using *t*-test, * Indicate Significance.

	P Value (Two-Tailed) OG			*P* Value (Two-Tailed) TM		
Parameters	CS v TFA	CS v TTA	TFA v TTA	CS v TFA	CS v TTA	TFA v TTA
StD	0.26	0.33	0.13	0.67	0.92	0.68
SPD	0.00*	0.00*	0.00*	0.00*	0.00*	0.00*
SwPD	0.00*	0.00*	0.00*	0.00*	0.00*	0.02*
LR	0.00*	0.00*	0.31	0.00*	0.07	0.00*
FF	0.15	0.11	0.03*	0.00*	0.00*	0.16
PO	0.00*	0.00*	0.00*	0.00*	0.00*	0.00*
SW-1	0.05	0.23	0.02*	0.00*	0.00*	0.00*
SW-2	0.00*	0.00*	0.00*	0.00*	0.06	0.00*

3.3. Statistical analysis

Table 7 shows the inter-subject assessment of mean SI for CS and LLA during OG and TM walking. Since TFA with Kinterra (prosthetic foot) participated in both OG and TM, therefore, SI results of TFA-K were assessed in statistical analysis only. StD or GCD was the only common parameter found with no significant difference ($p > 0.05$) when compared CS with either TFA or TTA or comparison between TFA v TTA was made during both OG and TM walking as shown in Table 7. There was no significant difference for FF and SW-1 when CS were compared with either TFA ($p = 0.15$, $p = 0.05$) or TTA ($p = 0.11$, ($p = 0.23$). No significant difference was observed for LR ($p = 0.31$) between TFA and TTA during OG walking. During TM walking, LR ($p = 0.07$) and SW-2 ($p = 0.06$) were found as not significant between CS and TTA whereas FF ($p = 0.16$) was not significant between TFA and TTA as shown in Table 7. Significant results are shown with an asterisk (*).

Statistical assessment of intra-subject was also carried out between OG and TM walking using paired sample *t*-test as shown in Table 8. Similar to inter-subject results, StD was the only common parameter found not significant for CS, TFA and TTA. SW-1 ($p = 0.08$) showed no significant different for CS. Amputee results showed no significant differences in SwPD ($p = 0.06$), FF ($p = 0.65$), PO ($p = 0.99$) and SW-2 ($p = 0.21$) for TFA, and SPD ($p = 0.86$), SwPD ($p = 0.76$) and PO ($p = 0.76$) for TTA as shown in Table 8. Significant results are shown with an asterisk (*).

4. Discussion

This study aimed to investigate the GA or SI analysis for temporal kinematic gait parameters in both CS and LLA based on a single IMU attached to the shank or prosthetic pylon on both limbs during level-ground walking. SI of the following gait parameters was investigated: stride time duration (StD) or gait cycle duration (GCD), stance phase duration (SPD), swing phase duration (SwPD), the inner phases of stance, loading response (LR), foot-flat (FF), push-off (PO) and the inner phases of swing, SW-1 and SW-2.

The TFA walked slower than CS during slow, normal and fast speeds, and agrees with other studies [9,25]. The trend was similar for TTA during normal and fast speed, however, at slow speed, TTA walked at a higher or similar speed as of CS (see section 2). It was observed that GCD reduced as speed varies from slow to fast in both groups. LLA showed longer GCD compared to CS irrespective of the locomotion mode i.e. slow, normal and fast walk. In addition, TFA showed large asymmetry in GCD during slow and fast walk compared to CS and TTA, however, found perfect symmetry during normal speed as shown in Table 4. Despite large asymmetry in TFA compared to CS and TTA, statistical results showed no significance ($p > 0.05$) in SI for GCD not only during OG walking but also in TM walking as shown in Table 7. A significant difference was observed between the groups in terms of SPD and SwPD with LLA having a longer stance and shorter swing phase for their intact side compared with their prosthetic side as shown in Table 5. SPD as a percentage of the GC was found to decrease whereas SwPD increased as speed increases in both groups which are in agreement with [9]. Due to this large variation in stance and swing phase duration, TFA showed greater asymmetry for SPD and SwPD compared to TTA and CS in all the activities. The results were found in agreement with [9,10]. Statistical results also showed that there was a significant difference ($p = 0.00$) between CS and lower limb amputees during OG and TM walking as shown in Table 7. The authors in Ref. [10] investigated the effect of walking speed on loading and temporal asymmetry in young and active LLA based on the kinetic information using eight force sensors, reported that temporal asymmetry decreases with an increase in walking speed, however, it contradicts with our results during OG walking in stance and swing phase asymmetry especially for LLA as shown in Table 10 and Table 11. One of the probable reasons for this disagreement is that data (n = number of strides) are varying between the locomotion mode in this study (i.e. $n = 18$ for slow, $n = 16$ for normal and $n = 14$ for fast walk).

The results, however, in Table 11 for CS and TTA during the swing phase showed higher asymmetry during fast walk compared to slow and normal walk that shows resemblance with our results. SI results, however, during TM walking, showed strong agreement especially for TTA with [10], as shown in Tables 10 and 11. The walking speeds reported in Ref. [10] were 0.5,0.9,1.2 m/s for a slow, normal and fast walk and the maximum walking speed was between 1.5 and 1.9 m/s whilst in this study, a range of 0.76–0.95 m/s, 1.04–1.16 m/s, and 1.37–1.50 m/s was recorded for slow, normal and fast walk respectively in both groups. It is imperative to mention here that the former study [10] carried out an investigation based on kinetic information from force sensors whereas this study investigated asymmetry analysis using temporal information from shank attached IMU on both lower extremities. A direct comparison of SI for the inner phases of stance and swing using IMU at shank could not be made for LGW as the authors were not aware of any data to be compared in the literature. Tashman et al. [27] evaluated the prosthetic shank with variable inertial properties of a TFA during normal and fast speed. The authors reported the mean swing time asymmetry of 19.5 % and 32.4 % during normal and fast walks respectively, however, only one TFA was assessed [27]. Isakov et al. [1] studied the influence of speed on gait parameters and asymmetry in TTA and reported that fast gait speed significantly affected all spatio-temporal parameters, however, the level of symmetry was not significantly different between the limbs. The evaluation was carried out with 11 TTA while walking at normal (0.94 m/s) and fast (1.38 m/s) speed. The present study also showed that level of symmetry was not significantly affected during normal and fast speed for CS and TTA as shown in Tables 10 and 11. TFA however, exhibited a large variation in SI between the locomotion modes. Blazkiewicz et al. [2] evaluated the asymmetry of spatio-temporal parameters of gait using four different indicators. The mean SI for the stance and swing phase was 2.38 % and 4 % respectively. The inner phases of stance and swing exhibited large asymmetry overall in both groups. The study also reported the mean asymmetry for LR as 9.73 % with a minimum and maximum values of 0.03 % and 36.31 % respectively [2]. The current study showed a mean (SD) asymmetry for LR as 5.4 ± 39.5 % for CS, −31.6 ± 30.9 % for TFA-K and −20.8 ± 36.4 % for TTA during OG walking.

Statistical results also showed no significance between CS and lower limb amputees for FF during OG as shown in Table 7. The longer SPD for the intact side of LLA especially for TFA confirms that LLA heavily rely on the intact side during walking. The shorter

Table 8
Intra-subject Assessment of Mean SI between OG and TM Walking using Paired *t*-test, * Indicate Significance.

	P Value (Two-Tailed)		
Parameters	CS	TFA	TTA
StD	0.46	0.41	0.46
SPD	0.00*	0.00*	0.86
SwPD	0.00*	0.06	0.76
LR	0.00*	0.00*	0.00*
FF	0.00*	0.65	0.00*
PO	0.02*	0.99	0.76
SW-1	0.08	0.00*	0.00*
SW-2	0.00*	0.21	0.00*

Table 9
Cost Comparison of the proposed system with existing systems.

Sr. No	Sensory system and brief description	Estimated Price (US Dollars)	Reference
1	2 × Shimmer3 IMU Units excluding dock	916	https://shimmersensing.com/product/shimmer3-imu-unit/ (accessed on Sep 13, 2023)
2	2 × Trigno IM Sensor (EMG +3D Gyro +3D Accelerometer 3D Magnetometer) with other accessories	3700*	Quoted price from the quotation (Q-7663) received from the delsys Inc. obtained on December 3, 2018
3	2 × IMUs, 2 × sEMG, 2 × Potentiometers, 2 × FSRs and 1 × myRio microcontroller board	850	[26]
4	2 × IMUs, 1 × moteino microcontroller and other electronic accessories	100	Proposed developed system

Table 10
Mean (SD) of Stance Phase Asymmetry during OG and TM Walking, Ref: Reference, PS: Present study, Max: Maximum, NA: Not available.

Mode	CS			TFA			TTA		
	Ref [10]	PS (OG)	PS (TM)	Ref [10]	PS (OG)	PS (TM)	Ref [10]	PS (OG)	PS (TM)
Slow	2.98 (2.29)	−0.05 (6.1)	−0.35 (6.2)	33.49 (14.5)	9 (7.8)	9.6 (5.4)	5.47 (4.68)	6 (4.2)	4.9 (4)
Normal	2.60 (1.84)	−1.2 (6.5)	0.58 (5.8)	25.8 (9.10)	15.3 (5.6)	11.6 (2.6)	6.27 (4.50)	6.1 (6.4)	9.7 (2.3)
Fast	2.78 (2.11)	−2.1 (5.7)	0.34 (5.9)	23.92 (5.80)	24 (5.3)	12 (2.2)	5.15 (3.54)	11 (7.4)	8.7 (2.6)
Max	3.1 (1.35)	NA	NA	23.42 (3.68)	NA	NA	3.52 (3.72)	NA	NA

Table 11
Mean ± SD of Swing Phase Asymmetry during OG and TM Walking, Ref: Reference, PS: Present study, Max: Maximum, NA: Not available.

Mode	CS			TFA			TTA		
	Ref [10]	PS (OG)	PS (TM)	Ref [10]	PS (OG)	PS (TM)	Ref [10]	PS (OG)	PS (TM)
Slow	3.72 (2.02)	1.1 (10.5)	0.4 (9.1)	−64.56 (30.3)	−9.6 (9.7)	−12.4 (7.9)	−10.3 (9.2)	−8.6 (7)	−7.7 (5.1)
Normal	3.62 (3.3)	1.9 (8.2)	−0.5 (7.6)	−50.39 (19.2)	−18.7 (8.3)	−15 (3)	−12.2 (8.23)	−11.9 (3.9)	−13.6 (4.1)
Fast	4.36 (4.23)	2.9 (8.1)	−0.27 (7)	−42.17 (10.9)	−24.7 (6.7)	−13.9 (2.8)	−10.7 (4.07)	−12.7 (6)	−13 (3.3)
Max	5.47 (2.26)	NA	NA	−37.38 (3.37)	NA	NA	−8.06 (11.1)	NA	NA

SwPD for the intact side and longer SwPD for the prosthetic side are mainly due to the differences in the inertial properties and mass of the prosthesis in comparison to the intact side [28]. The authors, however, reported large gait asymmetry when an attempt was made to manipulate the mass and inertial properties of the prosthetic side to match with the intact side [28]. TFA showed large asymmetry in stance and swing phase compared to CS and TTA due to the absence of an active mechanism for example in CS knee flexion starts at the end of weight-bearing whereas TFA tends to flex their prosthetic knee just before toe-off.

Valle et al. [14] evaluated the effects of gait and pelvis asymmetries of lower limb amputees using a single wearable inertial sensor, placed at the back. Kinematic data were acquired from the body and the pelvis during level ground walking on a 10-m walkway, and using timed Up and Go (TUG) test, effects of gait cycle and pelvis asymmetries were evaluated. The authors concluded that, gait cycle and pelvis asymmetries had an unfavorable effect while turning compared to linear walking. Further, it was reported that a single wearable sensor contributed significantly in relation to detection of various measures from motor contexts. Sung et al. [29] proposed an approach to predict the multi-joint angles of lower extremity based on a long short-term memory (LSTM) recurrent neural network using a single IMU placed at the shank. The experimental results showed promising results when joint angles from an IMU and the motion capture system were compared. Few other studies also reported the use of inertial sensors especially the IMU, acquiring kinematic data of different segments of human body for gait analysis and related studies has the potential to improve the quality of life of patients with amputation, fall avoidance, and enhance rehabilitation protocols with ease and reliability [30,31]. Table 9 shows a cost comparison of the developed system with few existing systems. It is clearly evident that the proposed system is a low-cost solution capable of acquiring kinematic data for gait analysis.

One of the key advantages of the proposed system is that only a single IMU at the shank is sufficient to identify the temporal gait parameters and further the evaluation of GA in CS and LLA during ADLs. On the other hand, the limitation of the current study includes a low number of participants, especially the lower limb amputees. The other probable limitation was analyzing small data (number of strides <50). Therefore, it is difficult to draw any generalized conclusion from the current findings. However, most of the findings agreed with the literature. Identifying and addressing gait asymmetries in individuals with lower limb amputation can mitigate long-term health complications related to the amputation, enhancing both mobility and overall well-being. It is believed that the significant findings drawn in this study with the limited number of participants could be beneficial in the assessment of stroke patients, lower limb amputees, patients with spinal cord injury, osteoarthritis, and other related pathological disorders. Also, the segmentation of GC into different phases can be utilized for user intent recognition of activities of daily livings. Future study will be evaluated with a large number of participants, lower limb amputees in particular, and for other activities such as ramp up and down, stair ascent/descent.

5. Conclusion

This study presented the use of a low-cost portable sensory system comprising an IMU placed at the shank of each lower extremity and is capable of estimating real-time gait events/phases, and further evaluating the GA or SI of temporal parameters of inner phases of stance and swing. Two approaches were considered to evaluate SI; based on: 1) the timing duration of each parameter separately and 2) the duration of each parameter as a percentage of the GC. Gait cycle duration showed less asymmetry in comparison to other temporal parameters in both groups, whereas the inner phases of stance such as LR and PO showed large variation in SI across all the participants. Overall, TFA exhibited large asymmetry for all the gait parameters compared to CS and TTA. Despite the limited number of participants, it is believed that the proposed system with current findings, in particular the inner phases of stance and swing provides additional insight and could be beneficial for the assessment of patients with neurological disorders and lower limb amputations. Further evaluation will be carried out with both spatial and temporal parameters on larger pool size, and for large number of strides in the future.

Ethical approval

This study has been approved by the University of Leeds Ethics Committee (Ref: MEEC 14-011).

Data availability statement

The data acquired and analysed for this study is confidential due to embargo period till June 30, 2025. Afterwards, it will be deposited in any related repository.

CRediT authorship contribution statement

Hafiz Farhan Maqbool: Writing – original draft, Validation, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Imran Mahmood:** Writing – review & editing, Visualization, Investigation, Data curation. **Ahmad Ali:** Writing – review & editing, Visualization. **Nadeem Iqbal:** Writing – review & editing, Validation, Investigation, Data curation. **Jin-Taek Seong:** Writing – review & editing, Resources, Investigation, Funding acquisition. **Abbas Ali Dehghani-Sanij:** Writing – review & editing, Supervision, Resources, Project administration, Funding acquisition. **Sundas Naji Alaziz:** Writing – review & editing, Visualization. **Mohammed Ibrahim Awad:** Writing – review & editing, Visualization, Validation, Supervision, Investigation.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests:

Jin-Taek Seong reports financial support was provided by National Research Foundation of Korea, and EPSRC. Jin-Taek Seong reports a relationship with Graduate School of Data Science, Chonnam National University, Gwanju 61186, Republic of South Korea that includes: employment. If there are other authors, they declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Acronyms

Acronym	Definition	Acronym	Definition
Acc	Accelerometer	M	Mean
ADLs	Activities of Daily Living	MD	Mean Difference
CS	Control Subjects	MST	Mid-Stance
CV	Coefficient of Variation	MSW	Mid-Swing
FF	Foot-Flat	OG	Overground
FFS	Foot-Flat Start	P	Prosthetic
FSW	Foot-Switches	PO	Push-Off
Gyro	Gyroscope	SD	Standard Deviation
GA	Gait Asymmetry	SI	Symmetry Index
GC	Gait Cycle	StD	Stride Duration
GP	Gait Parameters	SPD	Stance Phase Duration
GCD	Gait Cycle Duration	SwPD	Swing Phase Duration
HS	Heel-Strike	SW-1	Swing-1
HO	Heel-Off	SW-2	Swing-2
IC	Initial Contact	TM	Treadmill
I2C	Inter Integrated Circuit	TFA	Transfemoral Amputee
I	Intact	TO	Toe-Off
IMU	Inertial Measurement Unit	TS	Toe-Start/Strike
LR	Loading Response	TTA	Transtibial Amputee
LLA	Lower Limb Amputees		
LGW	Level Ground Walking		

References

[1] E. Isakov, H. Burger, J. Krajnik, M. Gregoric, C. Marincek, Influence of speed on gait parameters and on symmetry in transtibial amputees, Prosthet. Orthot. Int. 20 (1996) 153–158.
[2] M. Błażkiewicz, I. Wiszomirska, A. Wit, Comparison of four methods of calculating the symmetry of spatial-temporal parameters of gait, Acta Bioeng. Biomech. 16 (2014).
[3] A.R. Anwary, H. Yu, M. Vassallo, An automatic gait feature extraction method for identifying gait asymmetry using wearable sensors, Sensors 18 (2018) 676.
[4] S. Clemens, K.J. Kim, R. Gailey, N. Kirk-Sanchez, A. Kristal, I. Gaunaurd, Inertial sensor-based measures of gait symmetry and repeatability in people with unilateral lower limb amputation, Clin. BioMech. 72 (2020) 102–107.
[5] M.D. Lewek, C.E. Bradley, C.J. Wutzke, S.M. Zinder, The relationship between spatiotemporal gait asymmetry and balance in individuals with chronic stroke, J. Appl. Biomech. 30 (2014) 31–36.
[6] D. Gouwanda, S.A. Senanayake, Identifying gait asymmetry using gyroscopes—a cross-correlation and Normalized Symmetry Index approach, J. Biomech. 44 (2011) 972–978.
[7] M. Plotnik, N. Giladi, Y. Balash, C. Peretz, J.M. Hausdorff, Is freezing of gait in Parkinson's disease related to asymmetric motor function? Ann. Neurol.: Official Journal of the American Neurological Association and the Child Neurology Society 57 (2005) 656–663.
[8] S. Winiarski, A. Rutkowska-Kucharska, M. Kowal, Symmetry function – an effective tool for evaluating the gait symmetry of trans-femoral amputees, Gait Posture 90 (2021/10/01/2021) 9–15.
[9] S.M. Jaegers, J.H. Arendzen, H.J. de Jongh, Prosthetic gait of unilateral transfemoral amputees: a kinematic study, Arch. Phys. Med. Rehabil. 76 (1995) 736–743.
[10] L. Nolan, A. Wit, K. Dudziñski, A. Lees, M. Lake, M. Wychowañski, Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees, Gait Posture 17 (2003) 142–151.
[11] G. Chamorro-Moriana, A.J. Moreno, J.L. Sevillano, Technology-based feedback and its efficacy in improving gait parameters in patients with abnormal gait: a systematic review, Sensors 18 (2018) 142.
[12] S.H. Han, C.O. Kim, K.J. Kim, J. Jeon, H. Chang, Quantitative analysis of the bilateral coordination and gait asymmetry using inertial measurement unit-based gait analysis, PLoS One 14 (2019) e0222913.
[13] H. Sadeghi, P. Allard, F. Prince, H. Labelle, Symmetry and limb dominance in able-bodied gait: a review, Gait Posture 12 (2000) 34–45.
[14] M.S. Valle, A. Casabona, I. Sapienza, L. Laudani, A. Vagnini, S. Lanza, et al., Use of a single wearable sensor to evaluate the effects of gait and pelvis asymmetries on the components of the timed up and Go test, in persons with unilateral lower limb amputation, Sensors 22 (2022) 95.
[15] L. Wang, Y. Sun, Q. Li, T. Liu, Estimation of step length and gait asymmetry using wearable inertial sensors, IEEE Sensor. J. 18 (2018) 3844–3851.
[16] L. Yang, P. Dyer, R. Carson, J. Webster, K.B. Foreman, S. Bamberg, Utilization of a lower extremity ambulatory feedback system to reduce gait asymmetry in transtibial amputation gait, Gait Posture 36 (2012) 631–634.
[17] D. Trojaniello, A. Cereatti, E. Pelosin, L. Avanzino, A. Mirelman, J.M. Hausdorff, et al., Estimation of step-by-step spatio-temporal parameters of normal and impaired gait using shank-mounted magneto-inertial sensors: application to elderly, hemiparetic, parkinsonian and choreic gait, J. NeuroEng. Rehabil. 11 (2014/11/11 2014) 152.
[18] A. Tura, M. Raggi, L. Rocchi, A.G. Cutti, L. Chiari, Gait symmetry and regularity in transfemoral amputees assessed by trunk accelerations, J. NeuroEng. Rehabil. 7 (2010) 1–10.
[19] W. Zhang, M. Smuck, C. Legault, M.A. Ith, A. Muaremi, K. Aminian, Gait symmetry assessment with a low back 3d accelerometer in post-stroke patients, Sensors 18 (2018) 3322.
[20] E. Martini, I. Cesini, J. D'Abbraccio, G. Arnetoli, S. Doronzio, A. Giffone, et al., Increased symmetry of lower-limb amputees walking with concurrent bilateral vibrotactile feedback, IEEE Trans. Neural Syst. Rehabil. Eng. 29 (2020) 74–84.
[21] H.F. Maqbool, M.A.B. Husman, M.I. Awad, A. Abouhossein, N. Iqbal, A.A. Dehghani-Sanij, A real-time gait event detection for lower limb prosthesis control and evaluation, IEEE Trans. Neural Syst. Rehabil. Eng. 25 (2016) 1500–1509.
[22] H.F. Maqbool, M.A.B. Husman, M.I. Awad, A. Abouhossein, N. Iqbal, M. Tahir, et al., Heuristic real-time detection of temporal gait events for lower limb amputees, IEEE Sensor. J. 19 (2018) 3138–3148.
[23] H.-Y. Kim, Statistical Notes for Clinical Researchers: Assessing Normal Distribution (2) Using Skewness and Kurtosis, vol. 38, Restorative dentistry & endodontics, 2013, pp. 52–54.
[24] M. DouglasC, in: Douglas C. Montgomery (Ed.), Design and Analysis of Experiments, Wiley, London, 2009.
[25] Y. Liu, K. Lu, S. Yan, M. Sun, D.K. Lester, K. Zhang, Gait phase varies over velocities, Gait Posture 39 (2014) 756–760.

[26] M. Zubair, H.N. Ishaq, A. Raza, H.F. Maqbool, S. Zafar, Development and evaluation of a low-cost data acquisition system using heterogeneous sensors, Int. J. Sens. Netw. 40 (2022) 45–56.
[27] S. Tashman, R. Hicks, D. Jendrzejczyk, Evaluation of a prosthetic shank with variable inertial properties, Clin. Prosth. Ortho 9 (1985) 23–28.
[28] S.J. Mattes, P.E. Martin, T.D. Royer, Walking symmetry and energy cost in persons with unilateral transtibial amputations: matching prosthetic and intact limb inertial properties, Arch. Phys. Med. Rehabil. 81 (2000) 561–568.
[29] J. Sung, S. Han, H. Park, H.-M. Cho, S. Hwang, J.W. Park, et al., Prediction of lower extremity multi-joint angles during overground walking by using a single IMU with a low frequency based on an LSTM recurrent neural network, Sensors 22 (2021) 53.
[30] A. Demeco, A. Frizziero, C. Nuresi, G. Buccino, F. Pisani, C. Martini, et al., Gait alteration in individual with limb loss: the role of inertial sensors, Sensors 23 (2023) 1880.
[31] E. Simonetti, E. Bergamini, G. Vannozzi, J. Bascou, H. Pillet, Estimation of 3D body center of mass acceleration and instantaneous velocity from a wearable inertial sensor network in transfemoral amputee gait: a case study, Sensors 21 (2021) 3129.