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A Real-Time Gait Event Detection for Lower Limb Prosthesis Control and Evaluation

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Abstract— Lower extremity amputees suffer from mobility limitations which will result in a degradation of their quality of life. Wearable sensors are frequently used to assess spatio-temporal, kinematic and kinetic parameters providing the means to establish an interactive control of the amputee-prosthesis-environment system. Gait events and the gait phase detection of an amputee's locomotion are vital for controlling lower limb prosthetic devices. The paper presents an approach to real-time gait event detection for lower limb amputees using a wireless gyroscope attached to the shank when performing level ground and ramp activities. The results were validated using both healthy and amputee subjects and showed that the time differences in identifying Initial Contact (IC) and Toe Off (TO) events were larger in a transfemoral amputee when compared to the control subjects and a transtibial amputee (TTA). Overall, the time difference latency lies within a range of ± 50 ms while the detection rate was 100% for all activities. Based on the validated results, the IC and TO events can be accurately detected using the proposed system in both control subjects and amputees when performing activities of daily living and can also be utilized in the clinical setup for rehabilitation and assessing the performance of lower limb prosthesis users.

Index Terms—Event detection, Gyroscope, Lower limb prosthesis, Transfemoral amputee, Transtibial amputee

I. INTRODUCTION

Worldwide, numbers of individuals undergo the amputation of their lower limbs every year as a result of vascular disease and complications associated with conditions such as diabetes, cancer and trauma have increased. On average, approximately 185,000 persons undergo an amputation every year in the U.S. and it is estimated that this number will double by 2050 [1]. In the UK, around 34,109 lower limb amputations were carried out in 151 hospitals during 2007-2010 period [2]. Limb loss has a significant impact on individual's physical, mental and vocational abilities, generally resulting in the degradation of amputees' quality of life (QOL). Following an amputation, prosthetic devices can improve the amputees' QOL. Rapid technological advancement in the prosthetic field over the last few decades have caused prosthetic devices to evolve from purely passive (mechanical) devices to more advanced devices incorporating microprocessor controlled and powered components.

Human gait can be divided into a sequence of repeated phases and events associated with its cyclic nature with the stance and swing phases being the two main phases of the gait cycle. In terms of events, initial contact (IC) and toe off (TO) mark the beginning of a stance and swing phase respectively and provide information about stance time, swing time, cycle duration and gait asymmetry [3]. They are thus important assessment parameters and are frequently used in clinical studies as objective measures for evaluating the efficiency of the rehabilitative processes. The timing of these events supports the analysis of temporal parameters such as stride time and periods of single and double support [4]. Accurate identification of the phases/events is thus an important feature in the control of lower limb prostheses. For instance, the C-Leg® (OttoBock; Duderstadt, Germany) is equipped with a range of sensors including strain gauges to measure the anterior/posterior bending moments and an angular position sensor measuring the angular velocity of the knee joint. These measurements are used to detect the gait phases and hence to switch between controller states to provide necessary damping resistances required during amputees' walking [5]. Event detection information is then used as the reference datum for other measurements such as knee angle [3].

Gait events can be determined using force based measurement systems, typically by means of footswitches such as force sensitive resistors (FSR) located in a shoe insole. Such footswitches are suitable for obtaining on/off information, though not appropriate for precise analog measurements [6]. However, they are susceptible to mechanical failure, have generally poor durability and low cosmetic acceptance [7]. Among wearable sensors, accelerometers and gyroscopes are being deployed at different body locations for long-term monitoring of human gait [3, 6-13]. Recent advancements allow these sensors to be miniaturized, with faster processing capability and higher memory capacities to support outdoor applications [9]. The gyroscopes, in particular, are used for ambulatory gait assessment, foot-drop correction, control of lower limb prostheses and orthoses and other related clinical applications [14-17].

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Detection algorithms based on machine learning and rule-based heuristics are used to identify gait events/phases [6, 9, 15, 17-20]. Catalfamo et al. [9] evaluated the detection of IC and end of the contact (foot off) using a single gyroscope on the shank which was validated with seven control subjects walking on level ground and an inclined surface. The rule-based algorithm produced over 98% detection rate. However, a delay of about 120 ms in foot off detection was reported for real-time implementation. Selles et al. [3] developed an algorithm based on accelerometer sensor for estimating IC and TO. The algorithm was validated with ten control subjects and eight transtibial amputees (TTA) but was only tested offline for level ground walking activities at different speeds. Gouwanda et al. [15] presented a robust real-time gait event detection algorithm and compared the performance against previous algorithms. The detection rate was found to be 100%; however, a data latency of 100 ms was recorded as an average for event detection.

To the authors' knowledge, no study has to date been carried out to evaluate event detection algorithm with transfemoral amputees using a gyroscope. The aims of this study are therefore,

- To develop a simple and reliable heuristic rule-based real-time gait event detection (R-GED) algorithm using a single gyroscope attached to the shank or prosthetic pylon.
- To evaluate the reliability of the event detection system for lower limb amputees in both level ground and inclined surface ambulation.
- To evaluate the reliability of the event detection system for TFA when different types of prostheses are used.

The resulting system should be capable of providing accurate online event detection regardless of the prosthesis type used by a particular TFA.

II. Methodology

A. Subjects

Eight control (healthy) male subjects (CS) (mean age: 29.7 ± 5 years old; mean height: 174.8 ± 4.5 cm; mean weight: 77.6 ± 7.5 kg) without any apparent gait abnormalities, one male transfemoral amputee (age: 53 years old; height: 166.1 cm; weight: 66.8 kg) and one male transtibial amputee wearing a College Park Soleus (age: 51 years old; height: 180.3 cm; weight: 71 kg) participated in this study. The amputees had no other neurological or orthopedic disorder apart from their amputation and performed all the activities without the use of an ambulation aid.

The event detection system has been tested on the transfemoral amputee using different commercial prostheses as shown in Table I. All subjects wore their normal daily shoes. The system was mounted on the dominant leg of the CS. Information sheets containing background of the research, consequences of participating and description of the experimental activities were issued to each subject and a consent form was signed by each individual. All experimental procedures carried out in this research were approved by the University of Leeds Ethical Review Board.

TABLE I
Details of Transfemoral Amputee (TFA)

Labels	Type of Prosthetic Knee	Type of Prosthetic Foot	Amputation Reason	Year of Amputation
A	Ottobock 3R80	College park Venture		
B	Ottobock C-Leg	Ottobock 1E56 Axtion		
C	Endolite Orion*	College park Venture	Trauma (Chronic infection on the knee)	2009
D	Endolite Orion*	Endolite Echelon*		
E	Ossur Rheo	College park Venture		

*Blatchford Group (Chas. A. Blatchford & Sons Ltd)

B. Experimental Protocol

The transfemoral amputee wore each prosthetic leg for a week to familiarize himself with the device before the experiments were carried out. The TTA has been wearing the prosthetic leg for the past thirteen years. The prostheses were fitted by the professional prosthetist. A six degree of freedom inertial measurement unit (IMU) consisting of a three axis accelerometer and a three-axis gyroscope (MPU 6050, InvenSense Inc) was mounted on an acrylic holder, fixed to a flexible Velcro strap and placed on the lateral side of the shank. The gyroscope signal was used for the development of event detection algorithm with the maximum range of the gyroscope set to 500 degrees/sec. The x-axis of the gyroscope was aligned with the long axis of the tibia to record the shank angular velocity in the sagittal plane. A base unit consisting of printed circuit board (PCB) and a battery was attached to the lower part of the shank and a foot pressure insole consisting of four piezoresistive based Tekscan FlexiForce sensors (Tekscan Inc., Boston, MA, US) was placed into a shoe to provide the event timing required to validate the proposed algorithm. To adjust the insole for different shoe sizes, it was cut into two pieces. The location of the IMU, the footswitches and full experimental setup is shown in Figure 1. Once the system was attached to the subjects, they were asked to walk for approximately 2-3 minutes to familiarize themselves with the system and the test terrain to make sure they walked in a natural manner. The subjects were requested to walk along a 10 m pathway at three different self-selected walking speeds (normal, slow and fast). Five trials were carried out for each activity. Details of the participants' average walking speeds are shown in Table II. For ramp activities, the subjects were asked to walk up and down a 5.8 m long pathway with an inclination of 5° at their self-selected speed. Subjects were given a 10 minutes break between each activity.

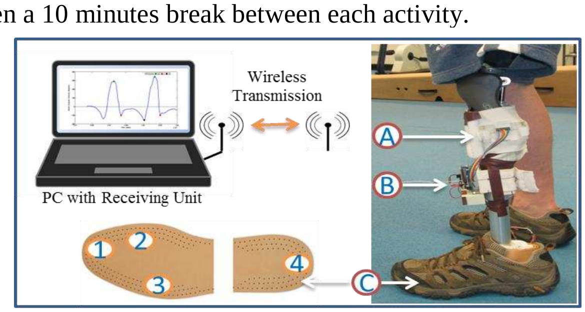

Fig. 1. Experimental Setup: Placement of — A: IMU; B: Base unit; C: Insole with footswitches; 1: Toe; 2 & 3: 1st & 5th Metatarsal; 4: Heel

TABLE II
PARTICIPANTS WALKING SPEED

MEAN ± STANDARD DEVIATION (M/S)

Subject	Slow	Normal	Fast
CS	0.92 ± 0.06	1.18 ± 0.07	1.46 ± 0.07
TFA	0.72 ± 0.04	1.03 ± 0.06	1.40 ± 0.1
TTA	0.65 ± 0.06	0.92 ± 0.05	1.45 ± 0.02

C. Real-Time gait event detection (R-GED) algorithm

A heuristic rule-based algorithm was written in Matlab (R2014a, The Mathworks, MA, USA) for the detection of gait events. The gyroscope signal on the shank in the sagittal plane has a distinct characteristic, in the form of two negative peaks (i.e. marked by triangles) each side of a large positive peak (circles) as shown in Figure 2. The positive peak corresponds to mid-swing (MSW) whereas two negative peaks before and after MSW correspond to TO and IC, respectively [8].

Preliminary data from two control subjects and a transfemoral amputee were used to develop the algorithm at a sampling rate of 100 Hz. The raw data were then filtered using a 2nd order Butterworth low pass filter. Two main aspects were taken into account while selecting filter cut-off frequency (fc): firstly, it must be low enough to attenuate the noise of the signal (that contains high-frequency oscillations such as impact spikes during IC) and thus reduce erroneous detection of the events and secondly, it must be higher than the principal frequency of the human gait [12]. Previous researchers have used cut-off frequencies ranging from 3 Hz to 35 Hz [9, 10, 12, 21]. For the proposed current algorithm, cut-off frequencies from 3 Hz to 12 Hz were tested on the signal and 10 Hz was selected; considering that the filtered signal must have as little latency as possible and emulate actual raw signal to reduce any phase shift or delay associated with event detection.

A flowchart of the algorithm and its implementation with the gyroscope signal is shown in Figure 2. The threshold values and rules were determined empirically based on preliminary data. The algorithm evaluates each sample sequentially in a prescribed time denoted by T_{given}. Once the signal is filtered, the algorithm computes the difference (Dn) between two consecutive samples (w_n and w_{n-1}). Initially, the algorithm searches for positive peak MSW meeting the following two conditions; 1) the slope must be positive and 2) the angular velocity must be greater than a threshold value of 100 degree/sec. Once the MSW is identified, the algorithm searches for the first negative minima for IC associated with a negative slope. During the detection of IC, it might be possible to have more than one negative peak closer to each other due to jitter in the gyroscope signal. To avoid such a situation a further condition was set that if in a window of 80 ms, there is any maxima closer to the already detected minima with a magnitude difference of ≤ 10 degree/sec, search for the next immediate minima and mark it. Otherwise select the previous minima as IC. Finally, the algorithm will search for TO after a 300 ms time-counter to avoid any false detection of TO. Once the counter finishes and the magnitude of the angular velocity is less than a threshold value of -20 degree/sec, the algorithm searches the local minima and marks it as TO.

D. Experimental Validation

For the validation of the algorithm, an insole containing four footswitches was placed inside the participants' shoe as shown in Figure 1. The data from eight control subjects, one TTA and one TFA amputee from both gyroscope and footswitches were captured in real-time using the wireless communication system. Figure 3 shows a sample of R-GED for the prosthetic side of TFA.

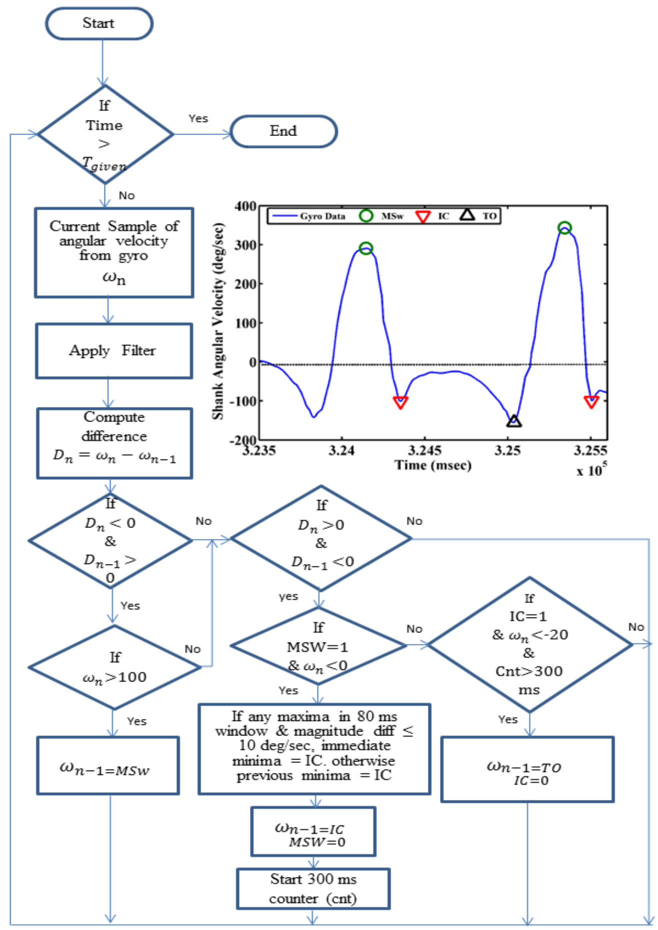

Fig. 2. Flowchart of the event detection algorithm based on gyroscope signal on the shank

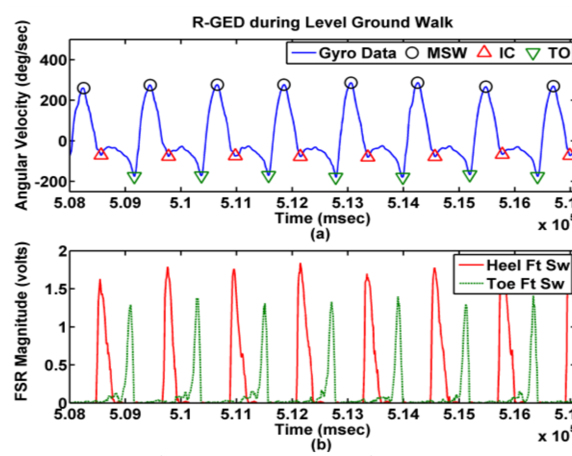

Fig. 3. Sample of R-GED using signals of (a) Gyroscope (b) heel and toe footswitches of prosthetic side of an amputee during LGW; MSW: Mid-Swing; IC: Initial Contact; TO: Toe Off; Ft Sw: Footswitches

E. Data Analysis

The comparison was performed by calculating the time difference between the gyroscope signal and the heel and toe footswitches (switches 1 and 4 in Figure 1). After careful calibration of FSRs, a suitable threshold (T) was considered such as for IC, T >= 0.1 volt and for TO, T <= 0.1 volt. The time difference (TD) was calculated for each activity using the following equation:

$$TD = T_{Gyro} - T_{Ft\ sw} \tag{1}$$

where T_{Gyro} and $T_{Ft\ sw}$ denote the timings of the detected events (IC and TO) from the gyroscope and the footswitches respectively. The TD for all the subjects during level ground walking (LGW), ramp ascending (RA) and ramp descending (RD) were then averaged to find the mean difference (MD). Other parameters such as the standard deviation (Std) and 95% confidence interval (CI) were calculated separately for each activity. The CI was calculated as follows [22].

$$CI = \bar{X} \pm t_{\alpha/2,n-1} s/\sqrt{n} \tag{2}$$

where $\bar{X}$ is the estimated mean, s is the estimated standard deviation, n is the total number of observations, and $t_{\alpha/2,n-1}$ is the "t-intervals", the value of which depends on the probability value and the degree of freedom [22]. The distributions of the TD were presented graphically using boxplots.

III. RESULTS

The accuracy of the results in terms of MD, standard deviation and CI for both IC and TO detection all expressed in milliseconds (ms) for eight control subjects, one TTA and one TFA amputee during level ground walking and ramp activities are shown in Table III. The current data contains both starting and stopping positions along with variation in speed within a trial whereas data with incomplete steps were excluded. Each trial covers about 10-15 strides for slow, normal and fast and 3-5 strides for RA and RD. The results of Table III and Figure 4 show that in general the proposed algorithm detects IC later and TO earlier than the footswitches for control subjects and intact side of both amputees for all activities. Little variation in terms of earlier or later detection exists for the prosthetic side in the case of prosthetics A, D, E and TTA.

TABLE III
DETECTION OF TIME DIFFERENCES (MS) OF INITIAL CONTACT (IC) AND (TO) BETWEEN GYROSCOPE AND FOOTSWITCH BASED METHOD.
MEAN DIFFERENCE ± STANDARD DEVIATION AND 95% CONFIDENCE INTERVAL

		Level Ground Walk		Ramp Ascending		Ramp Descending	
		IC	**TO**	**IC**	**TO**	**IC**	**TO**
Control Subjects		10.7 ± 17.9 [10, 12]	-7.6 ± 35.2 [-9, -6]	14 ± 21 [11, 17]	-5 ± 32 [-10, 0.2]	10.5 ± 17 [8, 13]	-25 ± 36 [-31, -20]
Prosthetic Side	A	13 ± 34 [9, 18]	13 ± 10 [12, 15]	37 ± 28 [25, 49]	23.5 ± 7.7 [20, 27]	-12.8 ± 15 [-19, -7]	17 ± 11 [12, 21]
	B	34.5 ± 30 [31, 38]	-11.7 ± 13 [-13, -10]	18.6 ± 12 [14, 24]	-34.6 ± 10 [-39, -30]	10 ± 25 [0.7, 19]	-122 ± 44 [-141, -105]
	C	29 ± 44 [23, 35]	-28.7 ± 18 [-31, -26]	27 ± 26 [16, 38]	-44 ± 21 [-54, -34]	-27 ± 8.8 [-31, -24]	-205 ± 38 [-221, -189]
	D	48 ± 19 [45, 51]	1.3 ± 29 [-2.8, 5]	39 ± 18 [30, 47]	51± 12 [44, 58]	9 ± 40 [-7, 25]	23.7 ± 13 [17, 30]
	E	20 ± 31 [16, 24]	31.5 ± 19 [29, 34]	23 ± 12 [17, 29]	-19 ± 6.7 [-22, -16]	-6 ± 42 [-23, 12]	-41 ± 5 [-43, -39]
	TTA	-5.7 ± 16 [-9, -2]	-12.8 ± 6.7 [-14, -11]	-10 ± 14.7 [-17, -2]	-11.6 ± 7.6 [-16, -7]	-11.8 ± 16.4 [-19, -4]	-22.8 ± 10 [-27, -18]
Intact Side	A	11 ± 13 [9, 13]	-44.6 ± 11.7 [-46, -43]	13.4 ± 13 [8, 19]	-40.6 ± 6 [-43, -38]	11.5 ± 12 [6, 17]	-41.5 ± 7 [-44, -38]
	B	2.5 ± 30 [-1, 6]	-32 ± 15 [-34, -30]	14.8 ± 7.7 [12, 18]	-20 ± 11 [-25, -16]	5.6 ± 14 [-0.25, 11]	-32.5 ± 14 [-38, -26]
	C	15 ± 29 [12, 19]	-36 ± 14 [-38, -35]	18.6 ± 28 [8, 29]	-21 ± 13 [-27, -16]	55 ± 51 [37, 73]	-42 ± 12 [-46, -37]
	D	28.7 ± 34 [24, 33]	-17 ± 26.5 [-1.2, 8]	20 ± 17 [13, 27]	21.4 ± 9.5 [17, 25]	20.3 ± 26 [9, 31]	-3.9 ± 19 [-11, 4]
	E	20 ± 20.4 [18, 23]	-12.6 ± 14 [-14, -11]	13 ± 10 [9, 17]	-13 ± 11 [-17, -8]	14 ± 13.7 [9, 19]	-15 ± 14 [-21, -10]
	TTA	5.7 ± 6.7 [4, 7]	-4 ± 9.5 [-6, -2]	1.9 ± 7.5 [-2, 6]	-3 ± 11 [-9, 3]	6 ± 7.3 [2.5, 9.5]	-11.6 ± 8 [-15.6, -7.6]

*Positive values indicate the delay in detection whereas negative values indicate early detection when compared against footswitch approach.

A. Distribution of time differences

The distributions of time differences (TDs) for IC and TO for all the activities are presented graphically in Figure 4. As there was variation in the number of events detected for each subject, the maximum number of available events for all subjects was considered to avoid any bias in the boxplots. For level ground walking 25 events of both IC and TO were considered for each subject and for each walking speed (slow, normal and fast). The overall IC and TO events for the eight control subjects and the amputees were 1200 and 1500 respectively. For ramp ascending and descending 16 IC and 17 TO events were considered for each control subject and each amputee. The overall variation in TDs showed a positive and negative values for IC and TO about the zero reference line. Intact side showed small variability in IC and TO detection. The prosthetic side of TFA, on the other hand, showed higher variability during ramp descending in particular due to the lack of proper TO.

B. Reliability

Reliability is defined in relation to the study as the ability of the system to detect events across the full range of subjects and their activities and is calculated as the ratio of total gait events detected by the gyroscope to the footswitch method. The total number of events detected (including both IC and TO) by footswitches during LGW was 2,793 for all the control subjects and 5,041 for all the amputees (considering both prosthetic and intact side). The total events detected during RA and RD was 323 and 357 for all the control subjects and 495 and 564 for all the amputees. In total, 9,573 events were detected across all subjects and all activities from which the gyroscope missing none of the events, resulting in 100% detection rate regardless of the prosthesis being worn.

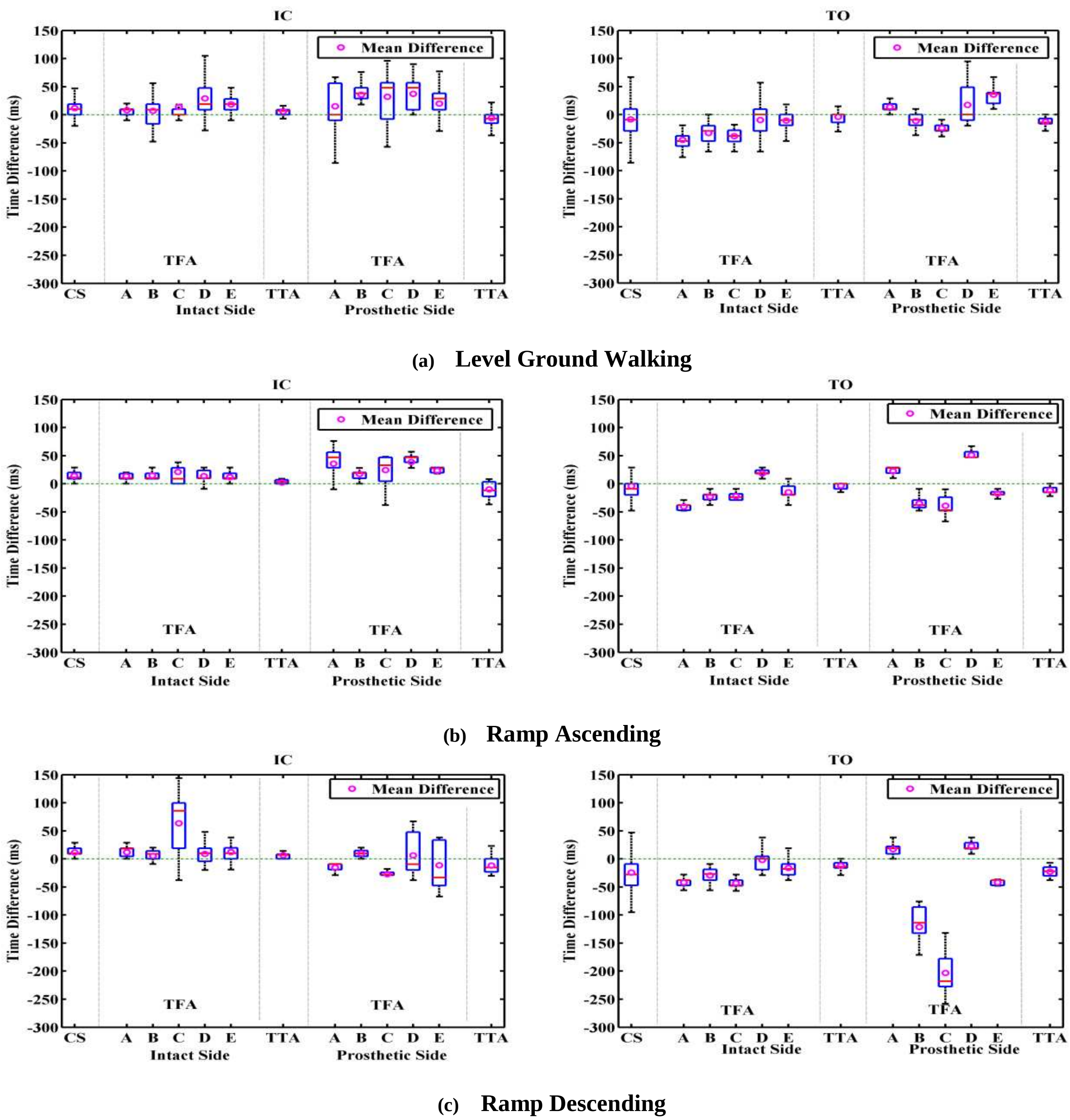

Fig. 4. Distribution of time differences based on box plot using normal distribution fit for all subjects during (a) Level Ground Walking (b) Ramp Ascending (c) Ramp Descending, CS: Control Subjects, TTA: Transtibial Amputee, (For A,B,C,D and E see Table I)

A pilot study was also carried out to investigate the sensitivity of the gyroscope to its placement on the shank. One control subject participated in this study at his self-selected walking speed of 1.2 m/s while placing a gyroscope at three different positions on the lateral side of the shank (namely 16 cm, 21.6 cm and 27 cm away from the ankle joint for a subject of height 1.66 m). The experimental protocol included a straight walking and turning. Three repetitions were carried out with the gyroscope at each position. Results showed 100% detection accuracy regardless of the location of the gyroscope.

C. Lack of Proper TO

The MD of more than 100 ms was found in the case of TO detection during ramp descending while the amputee was wearing prosthetics B and C as shown in Table III and Fig. 4. To further address the issue of high MD for TO during RD, the mean differences between gyro signal and 5th metatarsal (location 3 in Figure 1) were calculated in order to detect the Foot Off in the case of unnatural TO. MD was found to be much smaller for both the prostheses, particularly when compared with the 5th metatarsal, establishing that the transfemoral amputee was doing foot-off instead of toe off (see Table IV).

TABLE IV
MEAN DIFFERENCE ± STANDARD DEVIATION (MS) FOR TOE OFF AND FOOT OFF DETECTION DURING RAMP DESCENDING BETWEEN GYROSCOPE AND FOOTSWITCHES (TOE AND 5TH METATARSAL, SEE 1 AND 3 IN FIGURE 1)

Type of Prosthetics	Gyro - Toe	Gyro – 5th Metatarsal
B	-122 ± 44	17 ± 40
C	-205 ± 38	15 ± 25

Figure 5 shows a sample of one gait cycle (2nd stride) during ramp descending of the prosthetic side for three prostheses (A=3R80, B= C-Leg and C=Orion) and one of the control subjects (CS-1). To show the same effect for both groups, gyroscope magnitude was normalized between -1 and 1 and the time was normalized in the percentage of gait cycle i.e. from 0 to 100%. The area between the two vertical dotted lines shows the kinematic variation of CS-1 and the rest of the prostheses at the time of TO. With the 3R80, the knee has more flexion than the C-Leg and Orion. CS-1 has, even more flexion, prior to TO when compared to all the prostheses. Transfemoral amputees suffer from a noticeable asymmetrical gait of the intact leg and the prosthetic leg compared to the control subjects [23]. Figure 5 shows this asymmetrical effect in terms of stance and swing phase duration. Overall, the prosthetic side showed less stance phase duration for A, B and C compared to CS-1. Prosthetics B and C showed even less stance duration than A as the TO happened earlier due to lack of push off which affects the TO timing. Consequently, the prosthetic swing duration is longer compared to the control subject. The variation in the kinematics of the gait cycle in particular for TO during ramp descending may also be due to the lack of adaptability of the prosthetic knee and prosthetic foot.

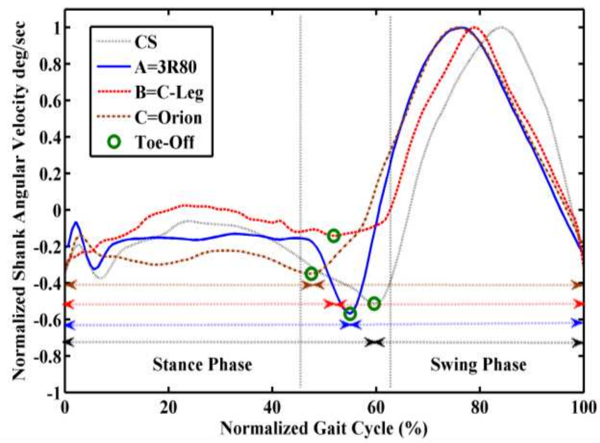

Fig. 5. A sample of Normalized gait cycle during ramp descending, CS-1: Control Subject 1

IV. DISCUSSION

This study presents a heuristic rule-based algorithm to identify IC and TO in both control subjects and lower limb amputees based on a single gyroscope attached to the shank. According to Salarian et al. [24], a rule-based algorithm generally performs 9 times faster than algorithms based on wavelet analysis [8], which indicates an advantage for real-time systems. Lesser soft tissue movement on the anterior side of the shank for the able-bodied control subjects and on the intact side of TFA than on the thigh and less signal variability between the subjects when compared with a gyroscope attached to the foot are the main advantages of attaching gyroscope to the shank [25].

Evaluating the time difference between the proposed system against the reference system for IC and TO detection in eight control subjects indicated MD ± Std of 10.7 ± 17.9 ms, 14 ± 21 ms and 10.5 ± 17 ms for IC and -7.6 ± 35 ms, -5 ± 32 ms and -25 ± 36 ms for TO during LGW, RA and RD respectively. The results for the TFA using five different prostheses showed a MD of ± 50 ms on average for all activities. The results for TTA were also found promising during all activities.

To assess the reliability of the proposed system, detection accuracy was found to be 100% for all control subjects, TTA and TFA using five different prostheses. Catalfamo et al. [9] reported a detection rate of 99.5% and Salarian el at. [24] reported the sensitivity in gait event detection as 99.6% in control subjects and 96.4% in patients with Parkinson's disease. Lee et al. [12] and Gouwanda et al.[15] also reported a detection rate of 100% for subjects with normal and altered gaits, however, they evaluated for LGW activities only.

A comprehensive literature review has revealed no previous study on event detection with TFA for LGW, RA and RD; hence, a direct comparison with other research cannot be made in relation to this study. Only one study [3] validated the system on transtibial amputees for event detection using two uni-axial piezoresistive accelerometers (range: ±5g) located on the lower leg, for LGW at different walking speeds. The results in the present study show that the overall trend for the intact side exhibits behaviour in terms of early and late detection to that of control subjects with relatively higher TD values. The prosthetic side showed higher values, particularly in the case of TO detection during RD as shown in Table III. This could be due to system errors such as improper placement of the footswitches, movement of the insole inside the shoe during ambulation, variation in the kinematics of the transfemoral amputee compared to able-bodied subjects and also prosthesis performance [26]. In addition, subject based constraints such as knee-ankle adaptability, particularly in the case of ramp activities and the use of these prosthetics for relatively short periods of time (each prosthetic leg was used for a week or two before the experiment) could also be potential explanations for the observed high TD values.

A high MD was also observed in the case of TO detection for some of the transfemoral prostheses. This is because transfemoral amputees tend not to increase the knee flexion during ramp ascending and descending and, therefore, make physical adjustments, particularly for RD, such as a shorter step length as a result of the smaller hip flexion with the prosthetic limb during the swing phase [27]. Also, the limitation of the rotation of the prosthetic ankle and compliance to the ground inclination in addition to improper controlling of the prosthetic knee especially during RD produce unnatural TO. While recording data from the TFA, it was observed that the subject was exerting more pressure on the intact side to compensate for the amputation side in pushing the body forward, especially during ramp based activities. However, based on the results, this asymmetric behaviour did not affect event detection.

Wearable sensors and the associated algorithms deployed in other studies have been reported to have been successfully validated with reference systems. Most of the studies were implemented offline with very few real-time data either from the healthy subjects or subjects with motor-control and functional disorders [3, 4, 9, 12, 13, 17]. Sellas et al. [3] reported a MD ± Std of 34 ± 25 ms and 19 ± 36 ms for IC and TO respectively in control subjects and 33 ± 41 ms and 13 ± 38 ms for IC and TO respectively in transtibial amputees. They also reported that separate algorithms for slower and faster walking speeds and the adjustment of the cutoff values of some of the low pass filters are required. Lee et al. [12] developed a quasi real-time event detection algorithm using a gyroscope attached to the shank and compared the results with four footswitches attached directly to the foot. The system was evaluated on 5 healthy subjects for LGW only. The mean error was 19 ms and -8 ms for IC and TO respectively. Gouwanda et al. [15] recently developed a rule-based algorithm and compared this with two previous algorithms and reported a latency with an average of 100 ms for event detection in real-time. For ramp activities, two studies were carried out into event detection using the gyroscopes, one attached to the shank and one on the foot.

TABLE V
SUMMARY OF DIFFERENT EVENT DETECTION STUDIES FOR CONTROL SUBJECTS (CS); PRESENT STUDY (PS); NA: NOT AVAILABLE

Ref.	IC Mean ± Std	TO Mean ± Std	Sensor and Location	Activities	Offline/Online	CS
[3]	34 ± 25	19 ± 36	Two uniaxial accelerometers below knee	LGW (different speeds)	Offline	15
[9]	-8 ± 9 -21 ± 15 -9 ± 20	50 ± 14 43 ± 10 73 ± 12	Gyroscope on the shank	LGW RA RD	Offline	7
[17]	-4.5 ± 14.4	43.4 ± 6	Gyroscope on the shank	LGW Mimicked shuffling gait	Offline	9
[12]	19 ± (NA)	-8 ± (NA)	Gyroscope on the shank	LGW (different speeds)	Online	5
[13]	-16.6 ± 11.9	3.7 ± 26.5	Gyroscope on the shank	LGW	Offline	9
[15]	NA	NA	Gyroscope on the shank	LGW LGW with knee and ankle braces	Online	16
PS	10.7 ± 17.9 14 ± 21 10.5 ± 17	-7.6 ± 35.2 -5 ± 32 -25 ± 36	Gyroscope on the shank	LGW (different speeds) RA RD	Online	8

A MD of -21 ± 15 ms and -9 ± 20 ms for IC and 43 ± 10 ms and 73 ± 12 ms for TO during RA and RD respectively have been reported by Catalfamo et al. [9] whereas Ghoussayni [10] reported an overall MD of -11 ms and 69 ms for IC and TO. Table V provides a comparison between the proposed system and previous approaches based on gyroscopes attached to the shank for control subjects only. Overall, the proposed system shows an improvement for IC and TO. The MD ± Std and percentage increase/decrease (% I/D) for IC was found to be slightly higher for LGW and RD, however, results of TO showed significant improvement when compared with [9] as shown in Table VI. Percentage I/D was calculated using this formulation by using absolute values.

$$Percentage\ I/D = \frac{\textit{Obtained value} - \textit{Reference value}}{\textit{Reference value}} * 100 \quad (3)$$

where obtained and reference values correspond to the values obtained in this study and the previous study respectively [9].

TABLE VI
PERCENTAGE I/D OF AVERAGE MEAN DIFFERENCE BETWEEN THIS WORK AND PREVIOUS STUDY [9]

	Level Ground Walk	**Ramp Ascending**	**Ramp Descending**
IC	34% (I)	33% (D)	17% (I)
TO	85% (D)	88% (D)	65% (D)

Accelerometers could provide an alternative approach to event detection. However, their output is affected by gravity and also require optimal placement on the segments of the human body to achieve a consistent performance. It should be noted that gyroscopes have issues of sensitivity to temperature variations. However, they offer numerous advantages when compared to accelerometers and magnetometers as enumerated below:

I. They can be placed anywhere along the same plane on the same segment to produce identical signals [21].
II. Their data are not subjected to gravity and/or linear acceleration, which may contain high frequency components [15].
III. Gyroscopic data is not affected by the local magnetic field [15].
IV. They are the most suitable device to monitor human gait over longer periods of time [6, 8].
V. A single axis gyroscope is sufficient to detect both IC and TO.

In this study footswitches were used as a validating system. These must be placed at optimal locations to accurately detect the gait events. An important challenge was to prevent the footswitches from moving inside shoe, during the participants' gait activities. A customized insole might improve this problem. The footswitch system was chosen over force platform in this study since the latter is limited to recording isolated steps. Further, a previous study [28] has shown high accuracy and minimal delay (± 10 ms for IC and ± 22 ms for TO) between a footswitch system and a force platform. The authors have concluded that the footswitch system can be a useful tool as it is a low-cost option for extending laboratory-based studies.

The detection of IC and TO cannot be possible without prior detection of MSW, and is one of the drawbacks of the proposed algorithm. For the algorithm to function properly the first step should be taken by the instrumented leg otherwise, the first TO will not be detected. Another issue is the adjustment of the threshold values for various subjects, including patients with functional dexterity to increase the continuous and reliable event detection with 100% accuracy. For this study, however, the same threshold values were used for eight control subjects, one TTA and one TFA with all prosthesis types during different ADLs.

The detection of IC and TO events are based on finding local minima on the gyroscope signal. A condition of 80 ms window to detect the actual IC was not used as a waiting delay in the algorithm rather it was used as a time constraint (i.e. either IC detected before or the timer continues until 80 ms) to ensure that IC detection happens within 80 ms window. It was observed that in most cases IC detection took place between 20-30 ms within that 80 ms window and did not require the entire 80 ms to find the peak in the signal. Furthermore, the condition that terminates the 80 ms timer is the magnitude difference between the first minima (detected as IC) and the next sample. It was observed that the first minima (i.e. when $MSW = 1$ and $w_n < 0$) was marked as a true IC for more than 98% of the entire IC events detected and only 2% fell in 20-30 ms time interval. In general, the algorithm requires the current sample to be compared with the previous sample and therefore, there is at least one sample delay (10 ms) in detecting IC and TO.

In this study, data latency, with a small number of exceptions, lies within a range of ± 50 ms. Kotiadis et al. [29] reported that a temporal tolerance of ± 0.05 s, at a sampling rate of 100 Hz is suitable for many biomechanics applications. The early detection of TO can be a pre-indicator to the prosthetic controller to take action to ensure proper toe clearance during the swing phase [30].

Lack of the knee joint in TFA leads to a larger variability between amputated and intact side. TTA data showed better results as TTA has the knee joints and a longer residual limb, and therefore they conceive a better proprioceptive feedback about the location and orientation of the limb, resulting in more control over the prosthetic foot/ankle joint during both the stance and the swing phases similar to the control group.

Accurate identification of gait events using wearable sensors placed at suitable locations and supported by a robust and reliable algorithm would be beneficial for the rehabilitation of lower limb prosthetic users and patients with disorders. These gait events (IC and TO) provide useful information about clinical parameters such as stance time, swing time and stride cycle duration. The information will be utilized in switching controller states using a finite state machine to provide the necessary damping resistances or actuation while amputees' are in ambulatory action. In addition, these gait events provide further information about asymmetrical timing behavior between the intact (sound) and the amputated side of lower limb amputees. Hence, the information from the gait events can be utilized to control and evaluate the performance of the prosthesis.

This study presented a simple heuristic rule-based gait event detection system for healthy control subjects and lower limb amputees. The system is based on a single gyroscope attached to the shank and is capable of identifying gait events (IC and TO) in real-time. Because of its portability, the system can be used in both indoor and outdoor environments to reflect the actual performance of the subjects. It could also be implemented in clinical applications such as with functional electrical stimulation (FES) devices to provide locomotion in paraplegic subjects [31], as a monitoring tool to evaluate progress through rehabilitation [8] and other ambulatory gait analysis requirements.

TFA and TTA gaits are different, however the aim of this work was to validate the system on both transfemoral and transtibial amputees for potential use in controlling and evaluating the prostheses as a case study. Limitation of this study is the small number of amputee participants. Future work will include the evaluation with more amputees from both groups.

V. CONCLUSION

This study presents the following outcomes:

- A reliable real-time gait event detection algorithm using a single gyroscope attached to the shank.
- Evaluation with eight control subjects, one transfemoral amputee and one transtibial amputee.
- Experimental results showing 100% detection accuracy of IC and TO across five different prostheses which ensured the robustness of the proposed system.

The proposed system could be used in the following applications:

- The development of control systems for lower limb prostheses to switch between control states based on phases and events.
- Outcome evaluation after hip, knee, or ankle replacement.
- A diagnostic tool for abnormal and pathological gait in relation to activities of daily living (ADLs).

Future work will include evaluation on a larger participant pool, on varying terrains and environments such as stair ascent/descent and during different manoeuvers such as acceleration and/or deceleration to make it more effective in the functional assessment of the gait and to utilize its outputs in prosthetics control.

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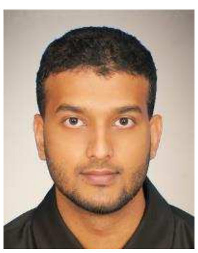

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