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Real-time gait event detection for transfemoral amputees during ramp ascending and descending

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Abstract— **Events and phases detection of the human gait are vital for controlling prosthesis, orthosis and functional electrical stimulation (FES) systems. Wearable sensors are inexpensive, portable and have fast processing capability. They are frequently used to assess spatio-temporal, kinematic and kinetic parameters of the human gait which in turn provide more details about the human voluntary control and amputee-prosthesis interaction. This paper presents a reliable real-time gait event detection algorithm based on simple heuristics approach, applicable to signals from tri-axial gyroscope for lower limb amputees during ramp ascending and descending. Experimental validation is done by comparing the results of gyroscope signal with footswitches. For healthy subjects, the mean difference between events detected by gyroscope and footswitches is 14 ms and 10.5 ms for initial contact (IC) whereas for toe off (TO) it is -5 ms and -25 ms for ramp up and down respectively. For transfemoral amputee, the error is slightly higher either due to the placement of footswitches underneath the foot or the lack of proper knee flexion and ankle plantarflexion/dorsiflexion during ramp up and down. Finally, repeatability tests showed promising results.**

I. INTRODUCTION

Lower limb loss affects millions of people worldwide and has overwhelming effects on individuals' such as loss of function, loss of sensation and reduced performance during activities of daily living (ADLs) which in turn reduces amputees' quality of life. The main causes for the lower limb amputation are vascular diseases, diabetes, trauma, hypertension, war injuries and accidents. The number of amputees are suggested to be about 45000 from which around 4000 lower limb amputations are recorded every year in England [1]. About 34,109 lower limb amputations were carried out in 151 hospitals across UK? during 2007-2010 [2]. On average 185,000 amputations are carried out every year in the U.S. and is predicted to double by 2050 [3]. Rapid technological advancements in the prosthetic field over the few decades have caused lower limb prostheses to evolve from simple mechanical joints to more complex devices based on damping mechanisms, actuators and microprocessor control to attain stable, symmetrical and energy efficient locomotion. These prosthetic devices have certain limitations: purely passive mechanical ones require significant voluntary control effort from amputees; active-damping controlled are unable to provide positive mechanical power required during more energy consuming tasks, e.g. sit to stand manoeuvers, ramp/stair ascending; actively driven consume more power than the human muscles and are costly.

Human movement is complex and highly variable therefore, correct analysis is important for understanding the gait during different ADLs. Objective gait analysis has been employed in multiple applications: evaluation of kinematic and kinetic parameters of human gait, design and optimization of assistive devices [4-6], ambulatory monitoring method for applications to Parkinson's disease [7], rehabilitations, prosthetics and orthotics [8, 9] . Gait analysis incorporates a variety of parameters used for objective assessment of gait such as cadence, speed, stride, step length, single and double support time percentage [10]. The temporal properties of gait and evaluation of gait trajectories are entirely based on accurate detection of the key events such as initial contact (IC) and toe off (TO) for segmentation of gait cycle. Accurate detection of gait events and/or phases is important for controlling the lower limb prosthesis. The C-leg (OttoBock), the Orion (Endolite), and the Rheo (Ossur, Iceland) use feedback from mechanical sensors to detect gait phases and then adjust the joint damping level. The C-Leg for instance, is equipped with strain gauges installed in the tube adapter for measuring anterior and posterior bending moment and a knee angle sensor to measure flexion angle and angular velocity of the knee joint. These measurements are used to detect the gait phases/events and provide necessary damping resistances during stance and swing phase as required during users ambulation [11].

Optoelectronic or kinetic equipment has been used in gait laboratories for the evaluation of gait events and intervals [12], however; these systems are not suitable for long term and outdoor measurements. For instance, the kinetic (footswitches based) systems are susceptible to mechanical failure due to sensor location and have less cosmetic acceptance and poor durability. Among wearable sensors, accelerometers and gyroscopes have been used actively at different body locations for long term monitoring of ADLs. More importantly, these sensors are cheap, non-invasive, portable, have low power and a fast microprocessor, as well as high memory capacity, allowing them to work for a longer period during indoor/outdoor ambulatory applications [13]. Gyroscopes in particular have been widely used for ambulatory gait analysis systems, for foot drop correction and for control of lower limb prostheses and orthoses and several other clinical applications [13-15].

Many control algorithms based on simple heuristic rules [16, 17], and machine learning techniques [5, 18, 19] have

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been implemented to identify gait phases/events. Most of these algorithms have been implemented offline. Few implemented in real-time but with poor latency and lower detection rate. Also, many works have dealt with level ground walking activities only. Furthermore, no work has been done with transfemoral amputees (TFA) to our knowledge. Therefore, this paper presents a real-time gait event detection (R-GED) algorithm based on the simple heuristics using a single gyroscope at shank. More importantly, this paper examines the robustness of the algorithm with both healthy subjects and transfemoral amputee for ramp ascending and descending at self-selected speed.

II. Method

A. Subjects

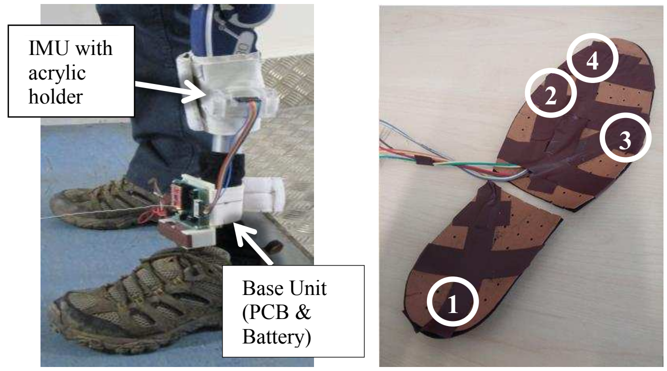
Eight healthy male subjects (age: 29.7±5 years old; weight: 77.6±7.5 kg; height: 174.8±4.5 cm) without any apparent gait abnormalities and one male transfemoral amputee (age: 52 years old; weight: 66.7 kg; height: 166.1 cm) participated in this study. The transfemoral amputee had no other neurological or orthopedic disorder apart from his amputation. All subjects wore their normal daily shoes. Subjects were briefed about the research background, consequences of participation and description of the experimental activities before obtaining their written consent. The Institution's Ethical Review Board approved all experimental procedures carried out in this research.

B. Experimental Protocol

A 6-DOF inertial measurement unit (IMU) comprising of an accelerometer and gyroscope (MPU 6050, GY-521) was mounted on an acrylic holder which stitched to a flexible Velcro strap. It was placed on the anterior side of the shank (Fig.1(a)). A base unit containing a printed circuit board (PCB) and battery was placed at the lower part of the shank with the help of the Velcro strap. A foot pressure insole consisting of four piezoresistive based FlexiForce sensors (Tekscan Inc., Boston, MA, US), was placed inside a shoe to detect the timing of the events and hence validate the proposed algorithm. The shoe insole was cut into two pieces to adjust it in different shoe sizes. The location of these foot switches is shown in Fig. 1(b). Once the suit was strapped on, subjects were requested to walk for about 2-3 minutes to familiarize with it. All the subjects were requested to walk up and down along a 5.5 m long walkway with inclination of 5^o at their self-selected speed. Subjects were given 10 minutes break in between activities. Data were collected for a transfemoral amputee on different days wearing different prosthesis types to check the consistency of the findings. Further details of the transfemoral amputee are shown in Table I.

TABLE I. Details of Transfemoral Amputee

Type of Prosthetic Knee	Type of Prosthetic foot	Amputation reason	Year of Amputation
Ottobock 3R80 (Rotary Hydraluic System)	College park Venture with multiple center of rotation	Trauma (Chronic infection on the knee)	2009
C-Leg	Ottobock 1E56 Axtion		



(a) (b)

Figure 1. Placement of equipment: (a) IMU and Base Unit (b) Footswitches placed on 1-Heel, 2-Metatarsal 1, 3-Metatarsal 5, 4-Toe

III. R-GED Algorithm

A. Description

Gait events are identified from the gyroscope signal attached at shank along the sagittal plane. Shank angular velocity signal has noticeable crest (maxima) and troughs (minima) that correspond to certain gait events. For instance, positive peak (maxima) corresponds to a mid-swing (MSw) whereas two negative peaks (minima) before and after positive peak correspond to TO and IC, respectively. An algorithm was developed using Matlab (R2014a, The Mathworks, MA, USA) by collecting small data at a sampling rate of 100 Hz from healthy subjects and transfemoral amputee. The gyroscope data were filtered using 2^{nd} order Butterworth low-pass filter. Researchers have used different cut-off frequencies for shank gyroscope signal ranging from 3 Hz to 35 Hz [13, 20-22]. For the proposed algorithm, cut-off frequency of 10 Hz was found to give the best result; keeping in view that the signal must be close to actual raw signal to avoid any phase shift and still filter the noise from the raw signal. The algorithm searches the filtered signal for the maximum value which is identified as MSw. Once MSw is marked, it searches for immediate negative trough which is marked as IC. Once IC is marked, the algorithm waits for a period of time and searches for second trough (minima) and marks this as TO. Details of the algorithm are shown is Table II.

TABLE II. R-GED Rules

Events	Rules
MSw	i. If the angular velocity is greater than 100 degree/sec, find maxima ii. Mark this maxima as MSw
IC	i. MSw is identified ii. Find the immediate local minima iii. Wait for 80 ms and if there is any maxima within this time period with the difference of angular velocity of 10 degree/sec, mark later minima as IC otherwise select the previous minima as IC.
TO	i. IC is identified ii. Wait for 300 ms iii. If angular velocity is smaller than -20 degree/sec search for local minima iv. Mark this minima a as TO

The threshold values and the rules were determined empirically based on the preliminary data from healthy subjects and transfemoral amputee.

B. Experimental Validation

For validation purposes, an insole consisting of four footswitches was placed inside the participants' shoe. The data of eight healthy subjects and one amputee from the gyroscope and the four foot switches were recorded in real-time through wireless communication and saved to a computer for data analysis. Fig. 2 shows the samples of real-time gait event detection (R-GED) for a transfemoral amputee during ramp ascending and descending respectively.

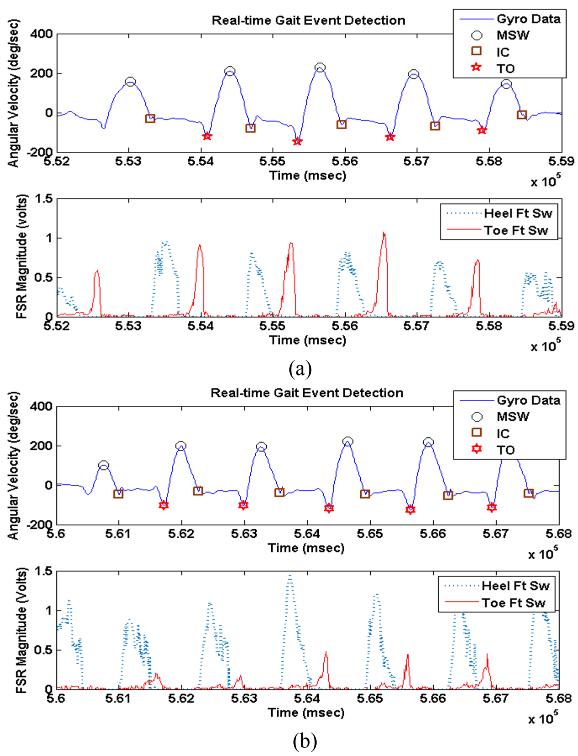


Figure 2. Samples of real time event detection based on shank angular velocity during (a) Ramp Ascending (b) Ramp Descending of Transfemoral Amputee (Prosthetic side). Note: MSw= Mid-Swing; IC= Initial Contact; TO= Toe Off; Ft Sw= Foot Switch

IV. Data Analysis

Differences in error were calculated using (1), where T_{gyro} and T_{ftsw} denote the timings of the events (IC or TO) detected from the gyroscope and the footswitches. Threshold values for initial contact and toe off foot switches, were (T >= 0.1 Volt) and (T <= 0.05 Volt) respectively. The time index values for heel and toe off foot switches were taken based on these threshold values.

$$Error = T_{gyro} - T_{ftsw} \qquad (1)$$

The mean differences for all the subjects for ramp trials were averaged and other parameters such as standard deviation and 95 % confidence interval (CI) were determined to compare the results with previously reported data. In addition, repeatability was calculated from Analysis of Variance (ANOVA) to check the consistency of detection errors between the subjects. Repeatability, R, was calculated based on the following equation [23].

$$R = S_A^2 / (S^2 + S_A^2) \qquad (2)$$

Where S_A^2 corresponds to the variance among groups and S^2 corresponds to the variance between groups.

V. Results and Discussion

The mean difference and other statistical parameters such as standard deviation and confidence interval (CI) were expressed in milliseconds (ms) for all subjects during ramp ascending and descending as shown in Table III. Positive values indicate the delay in detection, whereas negative values indicate early detection, when compared against the foot switch approach. Results for healthy subjects (overall mean difference) showed that the algorithm detected TO earlier and IC with few milliseconds of delay. From the previous reported data only two investigations were found that evaluated the events detection using gyroscope (one attached at shank and one attached on foot) for ramp activities. the mean difference in [13] for IC was found to be 21 ms and 9 ms for ramp ascending and descending respectively, whereas for TO it was -43 ms and -73 ms. Ghoussayni [21] mentioned the overall mean difference of -11 ms for IC and 69 ms for TO. The results of eight healthy subjects in this study showed the mean difference of 14 ms and 10.5 ms for IC and -5 ms and -25 ms for TO during ramp ascending and descending respectively. The results indicate the algorithm performs more accurately when compared against previously reported data. A thorough study in literature has revealed no previous study on TFA for ramp activity; hence a direct comparison with the results in this study is not possible. For transfemoral amputee data were recorded with two different prosthetics on two different days. Overall trend for the intact side was found to be the same as that of the able-bodied subjects with relatively higher values, however, prosthetic side showed variable results. The prosthetic side TO during ramp descending using the C-Leg microprocessor knee and IE56 Axtion foot noted a large delay. The mean difference found to be more than 100 ms as shown in Table III. According to Vrieling et al.[24], transfemoral amputees tend not to increase knee flexion in both ramp up and down and hence require some adjustments during ramp descending. Furthermore, the amputee wore the device for two weeks and may have not become entirely familiar to the device which may have altered his gait during performing ADLs. While collecting data from transfemoral amputee, it was observed that the amputee was exerting more pressure on his intact side to compensate for the prosthetic side in order to push his body forward. However, this asymmetry behavior did not affect the detection accuracy. This variation in timing difference may also be due to the sensor placement underneath the prosthetic foot. Repeatability results for both healthy and transfemoral amputee participants are listed in Table IV. Repeatability, R ranges from 0 to 1 [23] and the test shows all the values lies within an acceptable range. For example, for transfemoral amputee during ramp ascending, R = 0.46 means that the error is repeatable, but 46 % of variation is due to the differences among individuals.

TABLE III. DETECTION OF TIMING DIFFERENCES (MS) OF IC AND TO BETWEEN GYROSCOPE AND FOOT-SWITCH BASED METHOD: MEAN ERROR ± STANDARD DEVIATION AND [95 % CONFIDENCE INTERVAL]

		Ramp Ascending		**Ramp Descending**	
		IC	***TO***	***IC***	***TO***
Healthy Subjects (Overall)		14 ± 21 [11, 17]	-5 ± 32 [-10, 0.2]	10.5 ± 17 [8, 13]	-25 ± 36 [-31, -20]
Prosthetic Side	Day 1 3R80	37 ± 28 [25, 49]	23 ± 7.7 [20, 27]	-13 ± 15 [-19, -7]	17 ± 11 [12, 21]
	Day 2 C-Leg	19 ± 12 [14, 24]	-35 ± 10 [-39, -30]	10 ± 25 [0.7, 19]	-122 ± 44 [-141, -105]
Intact Side	Day 1 3R80	13 ± 13 [8, 19]	-41 ± 6 [-43, -38]	11.5 ± 12 [6, 17]	-41 ± 7 [-44, -38]
	Day 2 C-Leg	15 ± 7.7 [12, 18]	-20 ± 11 [-25, -16]	6 ± 14 [-0.3, 11]	-32 ± 14 [-38, -26]

TABLE IV. REPEATABILITY TEST ON ERROR DATA

Repeatability R	**Ramp Ascending**		**Ramp Descending**	
	IC	***TO***	***IC***	***TO***
Healthy Subjects	0.54	0.65	0.35	0.33
Prosthetic Side	0.46	0.94	0.29	0.89
Intact Side	0.33	0.90	0.10	0.86

The proposed algorithm could possibly miss the very first TO during first step if someone starts walking with the leg without the gyroscope attached, since MSw is being detected initially followed by IC and TO. A notable challenge is in preventing the footswitches from moving inside the insole during participants' walking activity. Customized insoles might improve this. In addition, using an array of footswitches might improve the timing differences compared to discrete sensor locations used in this study.

VI. CONCLUSION

This study presents a reliable and robust real-time gait event detection algorithm using a single gyroscope attached at shank. Despite having only 8 healthy subjects and one amputee, the methodology showed promising results and can be potentially used to detect main gait events required for controlling lower limb prosthesis. Future works will include a larger participant pool, implementation on other terrains such as stair ascent/descent for effective functional assessment of the gait.

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