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Real-time gait event detection for lower limb amputees using a single wearable sensor*

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***Abstract*—This paper presents a rule-based real-time gait event/phase detection system (R-GEDS) using a shank mounted inertial measurement unit (IMU) for lower limb amputees during the level ground walking. Development of the algorithm is based on the shank angular velocity in the sagittal plane and linear acceleration signal in the shank longitudinal direction. System performance was evaluated with four control subjects (CS) and one transfemoral amputee (TFA) and the results were validated with four FlexiForce footswitches (FSW). The results showed a data latency for initial contact (IC) and toe off (TO) within a range of ± 40 ms for both CS and TFA. A delay of about 3.7 ± 62 ms for a foot-flat start (FFS) and an early detection of -9.4 ± 66 ms for heel-off (HO) was found for CS. Prosthetic side showed an early detection of -105 ± 95 ms for FFS whereas intact side showed a delay of 141 ±73 ms for HO. The difference in the kinematics of the TFA and CS is one of the potential reasons for high variations in the time difference. Overall, detection accuracy was 99.78% for all the events in both groups. Based on the validated results, the proposed system can be used to accurately detect the temporal gait events in real-time that leads to the detection of gait phase system and therefore, can be utilized in gait analysis applications and the control of lower limb prostheses.**

I. INTRODUCTION

The recovery of functional attributes of human gait is one of the key objectives of lower limb rehabilitation. Real-time information of gait obtained during activities of daily living (ADLs) can be exploited to enhance the efficiency of gait interventions. Objective gait analysis is useful for understanding the gait during different ADLs. Gait can be described in terms of spatial (position and orientation) and temporal (events related) components [1]. Identification of the gait events facilitates the analysis of gait, design and development of assistive technologies such as functional electrical stimulation systems, prosthetics and foot-drop stimulators [2, 3]. In clinical applications, these events are used to evaluate the progress of patients with cerebral palsy (CP), Parkinson's disease, to improve alignment or fitting of the prostheses/orthoses [4, 5]. Most of the microprocessor based prostheses use the feedback from mechanical sensors to detect the phases/events and provide necessary damping resistance by switching between controller states.

Gait cycle (GC) can be segmented into two broad phases stance and swing which can be evaluated by detecting initial contact (IC) and toe off (TO). Stance phase can be segmented into different sub-phases such as loading response (LR), mid-stance (MST), terminal stance (TST), pre-swing (PSW) [6]. Methods for detecting these phases/events include the utilization of kinetic methods such as force sensitive resistors (FSRs) and kinematic methods using motion capture system. However, FSRs are less comfortable and prone to mechanical failure whereas, motion capture system are expensive, require large space and limited to laboratories only [7]. Wearable sensors such as gyroscope and accelerometer can be used at different body locations to detect the temporal gait events [1]. Additionally, they can be used for longer period of time and for both indoor/outdoor ambulatory applications [8].

Many control algorithms have been implemented using wearable sensors for accurate detection of gait events/phases based on threshold values, wavelet transformation and machine learning techniques [8-15]. The processing time for rule-based (threshold based approach) is faster than machine learning techniques. In most previous studies, it is common practice to divide the GC into two main phases by detecting IC and TO. There are very few authors investigated the inner-stance phase events. Mariani et al. [14] detected the inner-stance phase gait events, namely heel-strike (HS), toe-strike (TS), heel-off (HO) and toe-off (TO) using one inertial measurement unit (IMU) mounted on the foot. Detection of these four events yielded three inner stance phases, loading response, foot-flat and push-off system was evaluated with 42 subjects including both control subjects and patients with ankle osteoarthritis while walking for 50 meters pathway. Pressure insoles (Pedar-X) were used as a reference system. Good accuracy and precision were found when compared with the reference system, however, tested offline. Muller et al. [15] developed a novel gait phase detection algorithm in real-time based to detect four gait events namely, IC, complete contact (heel + toe), HO and TO. IMU was placed dorsally on the instep of each foot and data were recorded at 50 Hz while motion capture system was used as a reference. 14 control subjects and 5 transfemoral amputees participated in this study during level ground walk at slow, medium and fast pace. However, the results showed high time difference delays for both healthy subjects and amputees.

According to a previous study, the placement of IMU on shank has some advantages over placing on thigh and foot. For example in the former case there will be less soft tissue movement and less signal variability [8]. However, no study in the literature has revealed the inner-stance phase gait event detection using shank attached IMU for lower limb amputees to our knowledge. The current study, therefore, presents a novel rule-based algorithm to detect stance phase gait events in real-time using one IMU placed on the shank. By detecting

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these events, gait cycle can be segmented into different phases such as loading-response, foot-flat and push-off. The reliability of the system was evaluated with four control subjects (CS) and one transfemoral amputee (TFA) while walking at slow, medium and fast pace. The proposed system can be a useful tool for rehabilitation and in the control system development for lower limb prostheses/orthoses.

II. METHODOLOGY

A. Subjects

A pilot study was conducted from four healthy male subjects (age: 29.3 ± 1.7 years; weight: 80.3 ± 22.5 kg; height: 172 ± 6.2 cm) and had no apparent physical or cognitive abnormalities that affect gait. One male transfemoral amputee (age: 53 years old; weight: 66 kg; height: 166.1 cm) without any neurological or orthopedic disorder except his amputation also took part in the study. He was wearing Ottobock 3R80 prosthetic knee and Odyssey K2 College Park Venture prosthetic foot. Participants were provided with a brief description regarding the purpose of the study and experimental activities before obtaining their written consent. The experimental procedures involving human subjects described in this study were approved by the Institutional Review Board.

B. Experimental Protocol

Participants were equipped with a single inertial measurement unit (MPU 6050, InvenSense) consisting of a 3D accelerometer (range ±4g) and a 3D gyroscope (±500 deg/s), microcontroller, battery and other circuitry, placed on the anterior side of the shank (Fig.1) with the help of a flexible Velcro strap. The IMU was aligned along the long axis of the tibia to measure shank angular velocity in the sagittal plane (x-axis). Also, acceleration along the longitudinal axis of the shank (z-axis) was recorded by an accelerometer. Four piezoresistive FlexiForce sensors (Tekscan Inc., Boston, MA, US), were used as a reference system and were directly placed underneath the foot at different locations shown in Fig. 1. After getting familiarization with the equipment, participants were requested to walk along a 6 meter walkway at their self-selected pace with three variations (slow, medium and fast). Control Subjects (CS) walked barefoot whereas TFA walked with his normal shoes. A sufficient break was given to participants in between activities. All the data were recorded at a sampling rate of 100 Hz and transmitted to a personal computer (PC) through wireless communication.

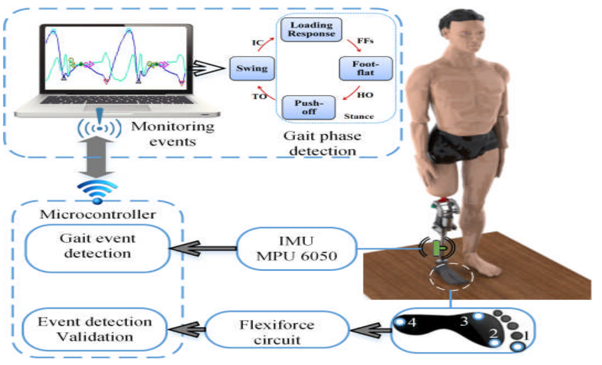

Figure 1. Experimental Setup: FlexiForce Footswitch (FSW) placed on 1-Toe; 2-Metatarsal 1; 3-Metatarsal 5; 4-Heel

III. ALGORITHM DESCRIPTION AND IMPLEMENTATION

Temporal gait events are identified from the gyroscope signal (rotation about the x-axis) and accelerometer signal (acceleration z-axis). TO and IC correspond to the two negative peaks occurred before and after a maximum peak known as Mid-Swing (MSW) in a shank angular velocity signal and has been detected accurately in our previous work [13]. Shank angular velocity also showed a maximum peak in the stance phase and is known as Mid-Stance (MST) [16]. MST occurs when the shank angular velocity is approximately zero [16]. Two more gait events called Foot-flat Start (FFS) and Heel-Off (HO) are identified before and after the MST using acceleration signal. The first instant when the foot is flat on the ground during stance is termed as FFS in this study. Two points are considered as a potential candidate for both FFS and HO (see Fig. 2). Acceleration signal during IC produces some peaks and then shows approximately a flat signal. Later, it started to increase during HO, whereas shank angular velocity decreases during dorsiflexion until TO occur. Identification of all these events will segment the stance phase into three phases termed as, loading response, foot-flat and push-off as shown in the state machine in Fig.1.

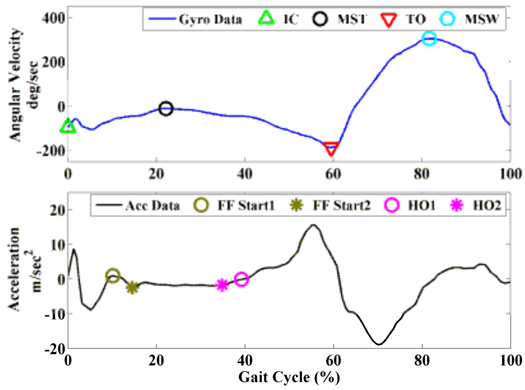

Figure 2. Detection of potential points for FFS and HO

Preliminary data were collected from control subjects and one transfemoral amputee. The IMU data were filtered using a 2nd order Butterworth low-pass filter with a cut-off frequency of 10 Hz. After IC detection, the algorithm will search for a maximum and minimum peak in the acceleration signal after a certain period of time (counter=40 ms). Once the two potential points for FFS were detected, the algorithm will search for the maximum peak of angular velocity in the stance phase. Since there are more noise artifacts in the stance phase than the swing phase, therefore, an automatic adjustment of the counter was incorporated to detect the actual maximum peak (MST). This counter is set based on the magnitude of MSW for each cycle (see details in Table I). Once MST is marked, two possibilities were considered for HO after a counter set to 30 ms: one based on the threshold value of acceleration and second when the signal crosses the zero. The proposed rules and threshold values were determined empirically based on preliminary data and found reliable when evaluated with four CS and one TFA. Details of the algorithm are shown in Table I. Fig. 3 shows the samples of real-time gait event detection system (R-GEDS) for a TFA during the normal walk.

TABLE I. TEMPORAL EVENTS DETECTION USING ACCELERATION SIGNAL (ACC) AND GYROSCOPE SIGNAL (GYRO), AN, AN-1: CURRENT AND PREVIOUS SAMPLES OF ACC

Events	Signal	Rules
FFS	Acc	a. IC is identified b. Counter = 40 ms c. Mark the maximum peak (FFS1) and minimum peak (FFS2)
MST	$Gyro$	a. FFS2 is identified b. Find the immediate local maxima c. Counter adjustment based on the magnitude of MSW Default value : counter = 50 ms If MSW < 260 ; counter = 90 ms If 320 < MSW > 260; counter = 70 ms
HO	Acc	a. MST is identified b. Counter = 30 ms c. Zero crossing, mark HO1 d. If $\|A_N - A_{N-1}\| \geq 0.1$ m/sec^2, mark HO2

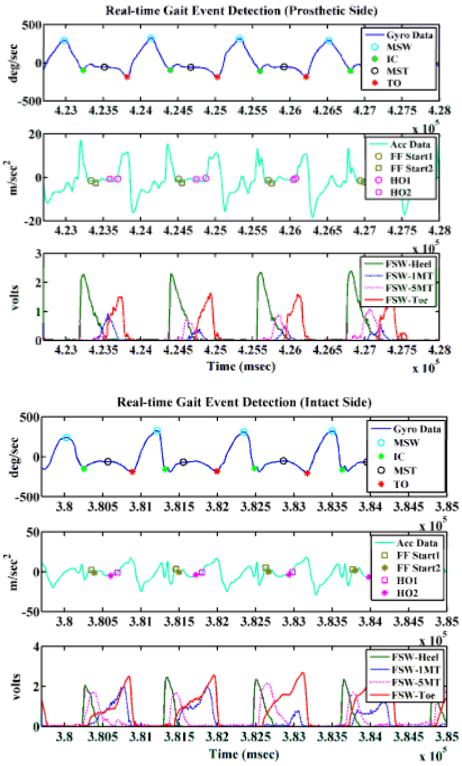

Figure 3. Samples of real-time gait event detection during normal walk of Transfemoral Amputee, (top) Prosthetic Side (bottom) Intact side; Note: (MSW) Mid-Swing; (IC) Initial Contact; (MST) Mid-Stance; (TO) Toe Off; (FF) Foot-Flat; (HO) Heel-Off; (FSW) Footswitch; (MT) Metatarsal

IV. DATA ANALYSIS

The timing difference (TD) between the kinematic and the kinetic sources were computed using (1), where T_{IMU} and T_{FSW} indicate the timings of the temporal events detected from the IMU and the FSW. For IC and FSW threshold values (T) were set to (T >=0.1 volts) whereas for HO and TO (T<= 0.1 volts) respectively.

$$TD = T_{IMU} - T_{FSW} \quad (1)$$

The mean difference (MD) and standard deviation (STD) were then calculated for all the subjects. The distributions of the TD are also shown statistically in Fig 4.

V. RESULTS AND DISCUSSION

Detection accuracy in terms of MD and STD for events detection (expressed in milliseconds, ms) are shown in Table II. Averaged measurements for IC and TO showed the similar trend for CS and TFA when compared against FlexiForce footswitches (FSW). Positive values indicate the mean detection was delayed by 17 ms for CS, 12 ms and 21.8 ms for TFA intact and prosthetic side, respectively. Negative values indicate the early detection of TO by -15.5 ms for CS, -23.8 ms and -7.5 ms for TFA. Evaluation of FFS was done by comparing the two potential points with the initiation of FSW at 1st and 5th Metatarsals. Wafai et al. [17] presented a study of plantar pressure distribution during the whole stance phase and the outcome showed low-pressure distribution at the metatarsal region indicate the start of foot-flat for CS. Based on statistical results, Foot-Flat Start2 (FFS2) was found to be a suitable candidate as MD was 3.7 ms for CS and -18.5 ms for the intact side of TFA. However, prosthetic side showed high MD of -105 ms, may be due to improper placement of FSW at 1st Metatarsal. MD of 16 ms was calculated when compared with FSW at 5th Metatarsal. FFS2 was also compared with the initiation of FSW at Toe termed as toe start (TS). MD of -21.5 ms and 20.7 ms was found for CS and TFA (intact-side). Prosthetic side, however, showed an early detection of -236 ms. For HO2, MD of -9.4 ms and 1.7 ms for CS and TFA (prosthetic side) were found respectively. However, 141 ms delay was found for amputee's intact side. It was observed that in order to push the body forward, TFA was exerting more pressure on his intact side to compensate for his amputated side. The transfemoral amputee does vaulting to complete the stride and to provide prosthetic foot clearance, he performed early HO and spending most of the stance phase time on his forefoot as shown by FSW signals in Fig. 3.

Two studies investigated the stance and inner-stance phase gait events, however, using foot-mounted IMU. Muller et al. [15] reported the MD of IC and TO approximately 50 ms and 100 ms for CS while walking barefoot. They did not report the delays for amputees' both sides separately. Overall, IC showed a delay of more than 100 ms for the prosthesis. Complete contact showed a delay of more than 200 ms for both CS and transfemoral amputees, whereas the delay for HO was between -20 ms to 60 ms for all the subjects [15]. Mariani et al. [14] evaluated four stance phase gait events with 42 subjects including healthy participants and patients with ankle osteoarthritis. They reported an overall MD and STD of 1 ± 8 ms, -4 ± 37 ms, 4 ± 54 ms and -3 ± 13 ms for HS, TS, HO, and TO respectively [14]. The current algorithm results in higher accuracy when compared to [15].

TABLE II. DETECTION OF TIMING DIFFERENCES (MS) OF GAIT EVENTS BETWEEN IMU AND FSW: MEAN DIFFERENCE ± STANDARD DEVIATION

	CS	TFA	
Events		**Intact**	**Prosthetic**
IC	17 ± 11.4	12 ± 9.5	21.8 ± 20
FFS1	-33 ± 61	-54.5 ± 75	-153 ± 103
FFS2	3.7 ± 62	-18.5 ± 75	-105 ± 95
HO1	78 ± 64	262 ± 100	114 ± 60
HO2	-9.4 ± 66	141 ± 73	1.7 ± 53
TO	-15.5 ± 22	-23.8 ± 8	-7.5 ± 15.5

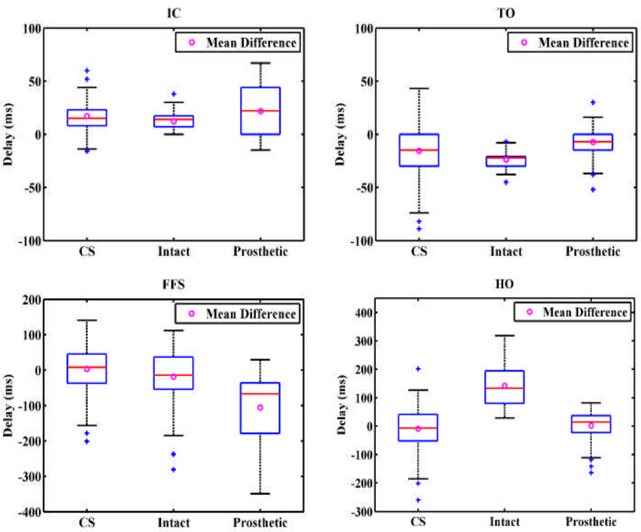

Figure 4. Time distribution of temporal gait events

Fig. 4 shows the time distributions of the gait events detected for all the subjects using IMU and validated with FSW. In terms of reliability of the system, 2706 events were detected by FSW for all the subjects and the proposed system missed MST, HO1 and HO2, three times overall in case of TFA. MSW, IC, FFS1, FFS2 and TO shows 100 % detection accuracy in both groups. Overall, detection accuracy was found to be 99.78 % in both groups. Fig. 5 shows a detection accuracy for all the gait events.

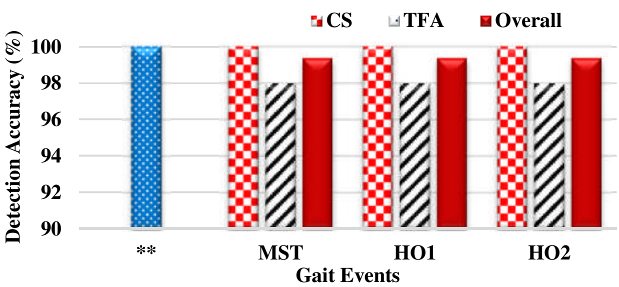

Figure 5. R-GEDS Detection Accuracy, **: MSW, IC, FFS1, FFS2, TO

FFS and HO events were detected by using the accelerometer that must be positioned accurately. High variation of these events may be due to the positioning of the accelerometer. The difference between the kinematics of a TFA and CS was another aspect to be considered. Moreover, asymmetry behavior may also affect the detection accuracy. Although we placed the footswitches directly underneath the foot and CS walked barefoot, still the variation in timing difference may also be due to the sensor placement. One of the limitations of the proposed study is the prior detection of MSW, as IC will not be detected until MSW is identified.

VI. Conclusion

This study presented a real-time gait event detection system for stance phase using a single IMU placed on the shank. Based on the events detected, stance phase can be segmented into loading response, foot-flat and push-off and provides additional insight in gait analysis applications. The efficacy of the proposed system will be evaluated with larger participant pool and its implementation on other terrains such as ramp ascent/descent in future.

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