Feasibility of A Gait Phase Identification Tool for Transfemoral Amputees using Piezoelectric-Based In-Socket Sensory System

<table>
<thead>
<tr>
<th>Journal:</th>
<th>IEEE Sensors Journal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Manuscript ID:</td>
<td>Sensors-25268-2019.R1</td>
</tr>
<tr>
<td>Manuscript Type:</td>
<td>Regular Paper</td>
</tr>
<tr>
<td>Date Submitted by the Author:</td>
<td>n/a</td>
</tr>
</tbody>
</table>
| Complete List of Authors: | Jasni, Farahiyah; International Islamic University Malaysia, Mechatronics Engineering
Hamzaid, Nur Azah; University of Malaya - City Campus, Biomedical Engineering
Al-Nusairi, Tawfik Yahya; University of Malaya - City Campus, Biomedical Engineering
Mohd Yusof, Nur Hidayah; University of Malaya - City Campus, Biomedical Engineering
Shasmin, Hanie Nadia; University of Malaya - City Campus, Center of Applied Biomechanics (CAB), Faculty of Engineering
Ng, Siew-Cheok; University of Malaya, Biomedical Engineering |
| Keywords: | SYST |
Feasibility of A Gait Phase Identification Tool for Transfemoral Amputees using Piezoelectric-Based In-Socket Sensory System

Farahiyah Jasni, Nur Azah Hamzaid*, Tawfik Yahya Al-Nusairi, Nur Hidayah Mohd Yusof, Hanie Nadia Shasmin, and Ng Siew Cheok

Abstract—Background: Gait detection is crucial especially in active prosthetic leg control mechanism. Vision system, floor sensors and wearable sensors are the popular methods proposed to collect data for gait detection. However, in active prosthetic leg control, a tool that is practical in its implementation and is able to provide rich gait information is important for effective manipulation of the prosthetic leg.

Objective: This study aims to ascertain the feasibility of the piezoelectric based in-socket sensory system that is hypothesized to be practical in implementation and provide sufficient information, as a wearable gait detection tool for transfemoral prosthetic users.

Method: Fifteen sensors were instrumented to the anterior and posterior internal wall of a quadrilateral socket. One transfemoral amputee subject donned the instrumented socket and performed two walking routines; single stride and continuous walking. The sensors’ responses from both routines were analyzed with respect to the gait phases.

Results and Conclusion: The results suggested that the sensor output signal corresponds to the force components behavior of the stump while performing gait. All sensors were seen active during the first double support period (DS1). The anterior sensors were prominent during the initial swing (Sw), while posterior sensors were active during terminal Sw. These findings correspond with the muscle activity during the respective phases. Besides, the sensors also show significant pattern during single support (SS) and the second double support (DS2) phase. Thus, it can be deduced that the proposed sensory system is feasible to be used as a gait phase identification tool.

Index Terms—Gait analysis, sensory system, prosthetic, piezoelectric, transfemoral

I. INTRODUCTION

RESEARCH in gait analysis has been a subject of interest for a long time. It is commonly used in medical and healthcare application [1-4] as it offers the necessary information by analysing human locomotion. The outcome of gait analysis includes gait phase detection, human gait kinetics and kinematics parameters identification, and evaluation of human musculoskeletal functions. The tools used for such gait analysis has evolved increasingly and thus had improved in accuracy and mobility.

One of the most important components in gait analysis is the collection of the gait information. There are few popular methods that are used in collecting gait information. The first one is camera-based/vision system. Single or multiple cameras were used to capture series of moving images in order to retrieve the gait parameters [4-7]. Camera or vision system is typically set up in the lab, in which the camera system is connected to a program that analyses the video captured by the camera. The advantage of this system is in its precision and accuracy, because the data is collected in a controlled environment. However, the controlled environment can also be a drawback in terms of its lack of practicality and restriction to represent real-life environment such as outdoor terrain. Moreover, the allowable distance for data collection is rather limited.

Another way to analyse gait is by using sensors instrumented onto the floor, also known as floor sensors [3, 8-10]. These methods are usually used to retrieve parameters related to ground reaction force (GRF), which could not be extracted using the vision system. These systems, i.e. the instrumented floor sensor and the camera system, are often used together in synchrony to complement each other in an integrated system, thus providing more complete information about the gait parameters, especially the joint kinetics. Studies have generally regarded such 3-dimensional motion analyses using camera systems incorporated with floor sensors as the ‘gold standard’ in gait and motion analysis.

However, in order to overcome the issue of limited mobility, wearable sensors, such as electromyography (EMG), Inertial Measurement Unit (IMU), and pressure sensors [11-13] are becoming a more favoured option due to its advantage over the camera system and floor sensors approach. They are more practical to be adopted outside the laboratory setting with

The first author received scholarship (SLAB) from Malaysian Ministry of Higher Education throughout her PhD studies.

F. Jasni is with the Mechatronics Engineering Department, Kulliyyah of Engineering, International Islamic University Malaysia, 53100, Selangor, MALAYSIA (email: farahiyahjasni@iiu.edu.my).

N. A. Hamzaid (Corresponding author), T. Y. Al-Nusairi, N. H. Mohd Yusof, and S. C. Ng is with Biomedical Engineering Department, University of Malaya, 50603, Kuala Lumpur, MALAYSIA (email: azah.hamzaid@um.edu.my, tawfikyahya@gmail.com, dayah90@hotmail.com, siewcng@um.edu.my).

N. A. Hamzaid (Corresponding author) and H. N. Shasmin is with Center of Applied Biomechanics (CAB), Faculty of Engineering, University of Malaya, 50603, Kuala Lumpur, MALAYSIA (email: hanie_nadia@um.edu.my).
greater movement distance and allows use in a more natural setting [14]. Wearable sensors are placed or attached on certain body parts to acquire gait parameters, such as position, force and moment, acceleration and others. Tao et al. [15] stated that there are three methods to perform gait analysis: using wearable sensors, which are electromyography methods [16, 17], and gait kinematics based method [18, 19].

Gait analysis are frequently adopted to describe and investigate prosthetics gait cases. Among the applications of motion analysis in prosthetics are gait adaptive strategies [20, 21], prosthetic socket pressure and force analysis [22]. However, the application of motion analysis is more crucial in microprocessor controlled (MPC) prostheses, because information extracted from the gait analysis (e.g. Activity identification, locomotion mode and gait phase status) become the input to the controller of the MPC prosthetic leg and used to eventually decide what is the next action to be done [23, 24].

Thus, the most practical approach to get the gait information is through wearable sensors because vision system and floor sensors seldom covers the practical space in which the prosthetic users move about in.

In order to ensure the safety and reliability of the MPC prosthetic leg, the efficiency of its gait detection tool has to be ensured too. Tucker et al [25] pointed out two most important criteria that need to be optimized in selecting the sensor system for controlling prosthetic device, which are: practicality and the quality of the data provided. In this aspect, practicality can be understood as the simplicity of the sensors’ donning and doffing process in terms of time, effort and risk. On the other hand, quality of the data involves the richness of the information provided and also the cleanliness of the data signal from external factors. These two factors need to be balanced in order to make an efficient gait information source.

To date, the piezoelectric device is becoming a favourable option to be used in prostheses applications [26-28]. However, very few studies reported the use of the piezoelectric sensor in lower-limb prostheses-related research. Lorenzelli et al. [29] and Sordo and Lorenzelli [30] proposed the piezoelectric sensors to be used to measure the interface force between socket and stump for TF amputee. The purpose of the measurement is to optimize the TF socket design. Meanwhile, El-Sayed [31, 32] conducted a feasibility study to utilize bi-morph piezoelectric sensors placed inside the socket to detect knee movement. Findings from the study suggested that there is high correlation between the piezoelectric signal and knee movement, hence open the window for further investigations for its application in sensory system for MPC prosthetic leg. Motivated by the discoveries, this study was conducted to study the feasibility of use of piezoelectric-based in-socket sensory system [33], which offers easy don and doff process, to detect the gait phases’ transitions for Transfemoral (TF) amputee. It is hypothesized that a feasibility proven piezoelectric-based socket sensory system could efficiently perform as a gait identification tool for TF prosthetic leg.

II. METHODS

Experiment was done to collect sensors responses during gait from a TF amputee, and the output signals of all sensors were analysed to see the pattern and trend of the signals of each sensor for each gait phases. The signals variation across multiple trials were calculated and particular with dominant response for each phase were identified.

A. Subject and Prosthetic Leg

One healthy male, transfemoral amputee participated in this study. The description of the subject is detailed in TABLE 1. Subject provided his written and informed consent by signing the consent form. Approval for the experimental procedure was obtained from the Medical Research Ethics Committee of University of Malaya Medical Centre.

<table>
<thead>
<tr>
<th>Age</th>
<th>36 years old</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height</td>
<td>174.5 cm</td>
</tr>
<tr>
<td>Body Mass</td>
<td>80.5 kg</td>
</tr>
<tr>
<td>Condition</td>
<td>-Unilateral amputation (left leg) -Has been wearing prosthesis for 17 years</td>
</tr>
</tbody>
</table>

Since the main aim of this study is to test the feasibility of the proposed wearable in-socket sensory system in detecting gait phases and its transition, the recruitment of one subject was deemed sufficient [26].

The socket used was polypropylene quadrilateral type and was custom-made to perfectly fit the subject. The sensors were instrumented to the socket following the configuration described in Jasni et al [33].

The hydraulic knee joint and Solid Ankle Cushion Heel (SACH) foot (3R60 EBS Pro, Ottobock, US) was used by the amputee. Prior to the experiment, the subject was given time to familiarize walking with the provided prosthetic leg to ensure that the captured data was as close as possible to his normal gait.

The experiment was conducted in the motion analysis lab, which was equipped with two force plates to capture the subject’s GRF data. The sampling rate of both force plates was set to 1 kHz. A motion analysis software (VICON Nexus version 1.8.5) was used to process the force plate data.

B. Piezoelectric-based In-Socket Sensory System

Fifteen polymer-based, PVDF-type piezoelectric sensors were mounted in zig-zag orientation with its negative surface facing upwards to ensure uniform polarity of the response. Each of the sensors was mounted using cantilever with elastic foundation configuration to the anterior and posterior inner wall of the socket, so that the sensors will touch the most active area of quadriceps and hamstring muscle [33]. Fig. 1 shows the in-socket sensory system adopted in this work and Fig. 2 illustrates the relationship between the strain and the response (i.e. voltage) of the sensor when mounted in this configuration, whereby the deflecting end of the sensor was positioned at the centre of the muscle bulk, and the fixed end of the sensor placed at the outer edge of the muscle bulk.

In order to calibrate the sensors, the response during the quiet standing activity was recorded for all sensors. Fig. 3 shows the response of all sensors for quiet standing, with response range within -0.02 to +0.02V. During quiet standing, all sensors
behaved similarly with output signals not different from one to another (p = 0.163).

C. Data Acquisition and Signal Processing

The output signal from the sensors was first pre-processed, amplified, and low-pass filtered before sent to the DAQ card for analogue-digital conversion. Each sensor was connected to an active low pass filter circuit with a voltage gain of 11 and the cut-off frequency of 800 Hz. Two 3 meters-long ribbon cables were used to connect the sensors to the signal conditioning circuit. The output signal from the circuits was passed to two similar DAQ card units (NI 9221, National Instruments, USA) with sampling frequency of 1 kHz.

In order to get a smoother signal, the acquired data was again band-pass filtered in the LabVIEW software by sixth-order Butterworth Infinite Impulse Response (IIR) filter with a passing frequency of 1 to 20 Hz. This was based on the findings by Prendergast et al. [34] and Malik et al. [35] which reported that muscle activity happens in low frequency.

D. Experimental Procedure

The experiment was divided into two routines. The first one was referred as the single stride walking routine and the second one was the continuous walking routine. During the single stride walking routine, data from the sensor and the force plates were collected simultaneously. Thus, the relationship between the force and the sensors’ response can be observed. Meanwhile, the continuous walking routine recorded a natural continuous walking data. Only sensory data was collected in this routine.

The rationale for having separate routines was to eliminate the need for long wire to be used to connect the sensors to the signal conditioning circuit. Pre-investigation was conducted that determined the optimum length of the cable to reduce noise caused by the cable was 3 m [36].

1) Single stride routine

The subject performed a single stride over the force plates. The starting point was set outside of the first force plate where the subject stood quietly and one force sensor was used as the trigger switch. It was placed on the second force plate and wired to the DAQ card module. The procedure for this routine is illustrated in Fig. 4(a) and the walking procedure in Fig. 4(b). Each trial was repeated 10 times.

The first trigger switch was used to synchronize the starting point of the routine for both systems. The sensor switch, placed on the force plate, produced instantaneous and simultaneous response with the force plate when knocked, thus was used as the signal synchronization marker. Fig. 4(c) illustrates the response for both systems during the start of the routine. Both systems (i.e. force plate and sensors data collection) data were synchronized by identifying the trigger point impulse. The x-axes of that two systems’ plot were considered synchronized after the starting point was determined.
Thereafter the force plate response, which actually represents the Ground Reaction Force (GRF) for the subject, were used as the indicator to determine the gait cycle. The point where the first force plate data started to raise was marked as the first heel strike. Meanwhile, the point of abrupt changes in the second force plate was considered as the second heel strike point, thus marked 100% gait cycle of the prosthetic leg. The phases were then further identified based on the GRF signals. The process was illustrated in Fig. 4(d). The data for GRF and sensory data for single stride was cropped accordingly for all 10 trials and the mean and standard deviation was calculated.

2) Continuous walking routine

One sensor was used as the trigger switch to mark the start and end of the routine. The procedure of the routine is illustrated in Fig. 5(a). A semi-circle path was used to compensate the wire-length limitation. However, it was assumed that the subject walked in a straight path manner, because the ratio of the path circumference to the stride length was larger than 18 (18.85:1), which led to the assumption that the curved path did not affect the subject walking pattern. Five trials were performed with an average of 6 complete continuous strides performed in each trial.

To mark the gait events during the gait, a video throughout the trial was recorded and analysed using Kinovea [37] to confirm and adjust the gait phase’s identification. Data for single stride was cropped accordingly. The mean and standard deviation of 30 complete gait cycles (5 trials × 6 strides/trial) for each sensor were calculated. Data from both routines were normalized to 100% gait cycle.

The significant sensors were identified based on the response of the sensor during the respective gait phase. Two criteria that the sensor must possess to be significant are; 1) at least one slope sign change, and 2) at least one peak with amplitude $>0.02V$, or Peak $>0.02V$, during the respective phase.
The equation to calculate the mean and standard deviation was described in (1).

\[
\text{Mean}(t) = \frac{\sum \text{SensorResponse}(t)}{n}
\]

\[
\text{StdDev}(t) = \sqrt{\frac{\sum (\text{SensorResponse}(t) - \text{Mean}(t))^2}{n}}
\]

(1)

where:

n=number of trials, and t=gait percentage

III. RESULTS

Based on the motion analysis and the average GRF curves, the average gait phase percentage distribution for this subject was determined for both walking types (Table 2). The graphs that represent the results for both routines are shown in Fig. 3.

Table 2

<table>
<thead>
<tr>
<th>Gait phase</th>
<th>Single stride walking</th>
<th>Continuous walking</th>
</tr>
</thead>
<tbody>
<tr>
<td>LR (DS1)</td>
<td>0% - 23%</td>
<td>0% - 19%</td>
</tr>
<tr>
<td>MST (SS)</td>
<td>23% - 35%</td>
<td>19% - 33%</td>
</tr>
<tr>
<td>TSt (SS)</td>
<td>35% - 46%</td>
<td>33% - 48%</td>
</tr>
<tr>
<td>PSw (DS2)</td>
<td>46% - 65%</td>
<td>48% - 62%</td>
</tr>
<tr>
<td>Sw</td>
<td>65% - 100%</td>
<td>62% - 100%</td>
</tr>
</tbody>
</table>

Note: HS-Heel Strike, LR-Loading Response, DS1-1st Double Support, MST-Midstance, SS-Single Support, TSt-Terminal Stance (TSt), PSw-Pre-Swing, Sw-Swing.

C. Single Stride Walking

Fig. 6 shows the mean response and standard deviation of the in-socket sensors with respect to the vertical ground reaction force (GRV) and Anterior-Posterior components over the complete gait cycle for all anterior and posterior sensors. In general, it can be observed that all sensors show low SD values for the Loading Response (LR), Midstance (MSt) and Terminal Stance (TSt) phase, and higher deviation was noted during the Pre-Swing (PSw) and swing (Sw) phase, except for sensor A3, which displayed a consistent SD for all gait phases. However, for the most proximal anterior sensors (A1 and A2) and mid- and posterior sensors (P4 and P5) displayed a greater data variation throughout the gait cycle as compared to the other sensors.

In terms of the intensity of the response, during the LR phase (0-23%), which is the first double support phase, sensors A3, A4, A5 and A6, and all posterior sensors except for P2, P4 and P6 showed a significant response. For the second section (23%-35%) and third section (35%-46%) which is the single support phase (SS), sensors A1, A2, and A3 for anterior sensors, and all posterior sensors except for proximal posterior sensors (P6 and P7) showed a prominent pattern compared to the others. During the second double support phase (PSw phase), all anterior sensors were active except for A5, A6 and A7, and P2, P3, P4, P5 and P6 were active in the posterior part. Finally, for Sw phase, almost all sensors were active except for P1.

B. Continuous Walking

Fig. 7 presents anterior and posterior sensors response signal for the continuous walking routine with its respective standard deviation. Overall, it can be seen that the standard deviation for continuous walking signals are more uniform throughout the phases. Nevertheless, sensor A2, A4, A6 (for anterior part), and P2, P4 and P6 (for posterior part) showed a lower deviation as compared to the other sensors in almost all phases.

During the first double support (LR phase), it can be seen that almost all sensors behaved actively. For the single support phases (MSt and TSt), the proximal and middle anterior sensors (A1, A2, A3 and A4), and P3, P5 and P7 (for posterior sensors) displayed a more prominent behaviour. During the second double support phase (i.e. PSw phase), A1, A2, A5 and A6 (for anterior sensors) and some posterior sensors (P2, P4 and P6). Finally, during Sw phase, all sensors showed strong responses.

C. Continuous vs Single stride

Fig. 8(a) and (b) illustrate the comparison of continuous walking and single stride walking response for all anterior and posterior sensors, respectively. Three main observations were noted. First, the amplitude of the single stride walking response was smaller than the continuous walking for almost all sensors. Second, the response pattern of most of the sensors showed similarity for both types of walking, but, shifted in time domain. For instance, for sensor P2, the positive peak for single stride walking occurred at ~10% of the gait cycle, but for continuous walking, it happened at ~20% of the gait cycle. Finally, for continuous walking had larger signal standard deviation, but more uniform throughout the gait cycle as compared to single stride walking signals. For instance, the calculated SD at 9% (LR phase) of the gait cycle is ~1.5times smaller than the SD calculated at 83% (Sw phase) for Sensor A5 for single stride, but ~0.9times smaller for continuous walking which indicates, that the SD for continuous walking is more uniform.

The significant sensors for single stride and continuous walking routines in the respective gait phases are summarised in Fig. 9.

IV. DISCUSSION AND CONCLUSION

Piezoelectric is a type of dynamic force sensor that detects the changes in force or pressure on the measured area with respect to time. In this application, an array of piezoelectric sensors was used to capture the force profile of the stump while it is inside the prosthetic socket and doing gait. It was hypothesized that there is certain force profile that is consistent every time the prosthetic user performs gait and piezoelectric sensor can translate it in terms of voltage signal. By analysing the signal and studying its behaviour for the respective gait phases, the gait phases can be characterized and eventually
become a tool to detect the gait phases. If it is proven efficient, it can be implemented as the input source of MPC prosthetic leg.

Based on the findings of this study, it can be deduced that the proposed in-socket sensory system could produce consistent signal for most of the gait phases, especially for the stance phase (LR, MSt, TSt). It is believed that higher deviation was noted on the swing phase for single stride routine because of naturally inconsistent speed during performing gait for single stride routine for every trial. Other than that, the termination for single stride routine might also be the reason for the higher deviation in the swing phase. During single stride routine, the subject was asked to perform only one stride while walking and stop. Thus, the swing might not be as natural as that should because the subject is already ready to stop. On the other hand, the deviation in continuous walking routine was seen to be more uniform throughout the gait cycle as the signal captured while the subject performed multiple strides. Thus, the motion is more natural and repetitive.

The possible factor for the difference between the sensors' response for the single stride walking and the continuous walking was due to the speed and momentum during walking motion. For continuous walking, the subject walked at a higher speed than the single stride walking, potentially due to the momentum gained throughout the continuous walking. Thus, higher amplitude was detected for continuous walking due to the momentum effect. The difference in the walking speed from one step to another might also contribute to the higher standard deviation in some phases of the continuous walking. The difference between the two walking types led to the deduction that the in-socket sensory system can also detect the difference in the walking type or speed by the amputee. However, more studies have to be planned and conducted to further verify this new hypothesis.

From the single stride routine results, it was observed that some sensors’ behaviour actually corresponded to the GRF components during the gait especially during LR and MSt phase. Some similarities were observed in sensors response and the AP-GRF component. This agrees with this study hypothesis that there are three major force components that contribute to the force profile of the stump, which are; the GRF, the interface force between the socket and the stump, and also the muscle force. Thus, some similar behaviour with the GRF components are expected. In addition, it was also observed that the pattern of the signal response was consistent with the muscle activity for quadriceps and hamstring muscle group [35, 38] in which two muscles are seen to be more active during LR phase, quadriceps muscle group is active during PSw and initial swing phase, and hamstring muscle group is active during the terminal swing phase for a normal person. The sensors signal showed agreement with this fact, where both anterior and posterior sensors were seen active during LR phase, and anterior sensors were prominent during the early stage of swing (60-80% of gait cycle) and posterior sensors were more active at the later stage of swing phase (80-100%). However, during PSw phase, it can be seen that some posterior sensors also displayed prominent response. One of the reasons that might contribute to this outcome is the effect of amputation. The muscle activity of a TF amputee, might be different from the one displayed by the normal person, and can also be different between two TF amputees [20]. Because of that fact, the authors deduced the pattern of output signal of this sensory system might vary from one user to another user, depending on the user’s stump’s length and condition, and muscle strength of the user. Therefore, the sensory system needs to be calibrated for each user and there is no one general template that can be used for all users. The calibration process includes Maximum Voluntary Contraction (MVC) evaluation so that the active muscle area can be identified, and sensors are placed on the active area. This is only needed once for each user. The same placement can be used for the same user if the socket must be changed. The calibration process can be done during the training sessions with the physician at the beginning of the prosthesis use.

As a limitation, this study did not explicitly measure muscle activation which would require EMG electrode placement as part of the data collection. However, EMG placement would present conflicting space requirement with the in-socket sensors. This space limitation was addressed by performing the experiment with the same transfemoral amputee participant in the earlier study [33] in which his gait muscle activation was documented and referred to in the study design of this work [33].

![Fig. 6. In-socket sensors response with respect to the ground reaction force and its AP components over a complete gait cycle for anterior sensors and posterior sensors for single stride walk. (Note: HS: Heel Strike; LR: Loading Response, MSt: MidStance, TSt: Terminal Stance, PSw: Pre-Swing and TO: Toe Off).](image-url)
Fig. 7. In-socket sensors response for continuous walking over a complete gait cycle. (Note: HS: Heel Strike, LR: Loading Response, MST: MidStance, TSt: Terminal Stance, PSw: Pre-Swing and TO: Toe Off).

Fig. 8(a). Anterior sensors response for continuous walking and single stride walking.

Although separate studies are needed to confirm the relationship between each force components and the sensors’ output signal quantitatively, this study has proven the feasibility of extracting the information. Besides, this study also shows that this proposed wearable sensory system has the potential to provide more information (i.e. muscle and GRF information), which cannot be achieved by gait kinetic based only method [39, 40] or EMG based only method [16] in one sensory system.

Fig. 8(b): Posterior sensors response for continuous walking and single stride walking.

Fig. 9. The summary of prominent sensors for each gait phase. The black boxes represent the single stride walking routine, the grey boxes represent the continuous walking routine. (Note: DS1: Double Support 1, SS: Single Support, DS2: Double Support 2, Sw: Swing)

ACKNOWLEDGMENT

The authors thank Mr Suparjo for his contribution as the subject in this study and also Performance of Body and Analysis of Movement Lab, University of Malaya for their permission to use the facilities for the experiment in this project.

REFERENCES


Yang, V. Stankovic, L. Stanko—detection of Prosthetic Knee Movement Phases via In... for Footstep


Reviewer: 1

Recommendations: This resubmitted version is much improved than the first version. The conclusion that the proposed sensory system is feasible to be used as a gait phase identification tool is well supported. The reviewer has some questions and concerns which hopefully can be addressed by the authors.

The authors thank the reviewer for this recommendation.

Comment 1: Bluetooth wireless DAQ is widely used in state-of-the-art wearable sensors [1-2]. However, the proposed sensory system still has a 3 meter cable. What blocked the usage of Bluetooth? With a wireless DAQ, the subject will be able to walk freely. This will definitely increase the potential application of the proposed In-Socket Sensory system.

Thank you for the suggestion. Yes, we are aware of the availability of the wireless DAQ technology and we do agree the utilization of that technology will increase the potential application.

However, since this study is still in feasibility phase, we would like to reduce the ambiguities in the response caused by external factor, e.g. transmission loss and etc. In this study, we used the 3-m cables which we already examined the performance, and noted the loss and noise caused by the cables (i.e. response of direct connection using jumper, and connection using 3-m cables were compared). As we are moving forward in the research, we definitely will consider to utilize the wireless DAQ used in [1,2] and recommended by the reviewer.

Comment 2: What is the next step of this study? Are you going to use the signals from the in-socket sensors to control the artificial limb? Or as mentioned above, moving to wireless?

Thank you for the suggestion. The bigger aim of this research work is to use the sensors’ signal to control the movement of the microprocessor-controlled (MPC) prosthetic leg. The next step is to develop a classifier to classify the signals and use it as control input for the MPC prosthetic leg. Of course, a parallel study can be conducted to improve current wired sensory system to wireless platform.

Comment 3: Fig. 9 shows that not all the sensors are needed. Is there a plan to eliminate redundant sensors? Also, it is not clear why the sensor arrays are designed in this way.

Yes, the future work involves correlation analysis between number of sensors with classifier performance in detecting gait phases to optimize the number of sensors.

The idea is to let the deflecting end of the sensor be positioned at the centre of the muscle bulk, and the fixed end of the sensor placed at the outer edge of the muscle bulk. (Line 129-131, second column)

We have conducted a study to determine the area of the muscle bulk which was already reported in [3].


Comment 4: The format is messed up in Line 282 to 283.

Thank you for the comment. We have fixed the format in Line 282 (Line 276 in the new document).

Comment 5: Please make sure to have necessary legends for all the figures. For example, what is the x axis in

The legend for Fig. 6 and Fig. 7 has been added, and the borders of all figures are checked and amended accordingly.
**Reviewer: 2**

**Recommendation:** Publish Unaltered

**Comments:**
- The authors responded to my comments clearly.
- The current version of the paper can be published in the IEEE sensor.

The authors thank the reviewer for this recommendation. Thank you for the paper summary and suggestions.

**Reviewer: 3**

**Recommendation:** Publish in Minor, Required Changes (as noted in the Comments section. This rating may not be assigned for Sensors Letters.)

**Comments:**
- Fundamental information and pieces of evidence related to muscle activities of active muscles for prosthetic leg control mechanism are missing as highlighted in the commented article attached herewith. Authors are required to address them and include appropriate justification in this article in order to be different to their similar article published as reported in [33].

The authors thank the reviewer for this recommendation. Thank you for the paper summary and suggestions.

---

**Title and Abstract Explanation:** In the abstract, authors do not provide substantial evidence for muscle activities relevant to the active prosthetic leg control mechanism. Active leg control is dependent on muscle activities too not solely based on force analysis. Authors would have done the comparative analysis of the active area of muscles chosen from amputee's leg for the sensor placement with the identical muscles of the healthy leg in order to verify the validity of muscle activities of relevant active muscles chosen.

This paper did not explicitly measure muscle activation which would require EMG electrode placement as part of the data collection and conflicting placement with the in-socket sensors and space would be an issue. We however have documented the muscle activation in our earlier and related study.


“As a limitation, this paper did not explicitly measure muscle activation which would require EMG electrode placement as part of the data collection. However, EMG placement would present conflicting space requirement with the in-socket sensors. This space limitation was addresses by performing the experiment with the same transfemoral amputee participant in the earlier study [33] in which his gait muscle activation was documented and referred to in the study design of this paper [33].” (Line 370-379 second column).

“Active leg control is dependent on muscle activities too not solely based on force analysis. Authors would have done the comparative analysis of the active area of muscles chosen from amputee's leg for the sensor placement with the identical muscles of the healthy leg in order to verify the validity of muscle activities of relevant active muscles chosen. There are considerable research carried out during past decades on multi-sensor...”

Thank you for the comment.

We have studied the recommended article and another article written by the same author.

fusion for lower limb activities such as the below article which provides the insight into the effort of assessment of muscle activity jointly with motion sensors; “An Intelligent Recovery Progress Evaluation System for ACL Reconstructed Subjects Using Integrated 3-D Kinematics and EMG Features”, IEEE Journal of Biomedical and Health Informatics (IEEE J-BHI), ISSN 2168-2194, Vol. 19, Issue 2, pp. 453-463, March 2015; http://ieeexplore.ieee.org/document/6805568/”

However, we found out that, these articles did not report the muscle activity of all muscles which were considered in the study (vastus medialis, vastus lateralis, semitendinosus and biceps Femoris), but only reveals the pattern for vastus medialis and vastus lateralis across the gait cycle extensively, while in our study, we focused more on the muscle belly i.e. rectus femoris and bicep femoris. Therefore, the comparison might not be directly compatible but rather just an association (Figure 1).

![Figure 1: Muscle activation sequence during gait.](image)

While authors have realized the importance of relevant muscles in their own article published before, authors failed to verify the validity of muscle activity of the amputee’s leg. The significant improvement for this article would have been the inclusion of muscle activity validation.

The article [34] might be a good reference to justify this claim. But, the experimental claim on the same might require further pieces of evidences using the article below published;


As a limitation, this paper did not explicitly measure muscle activation which would require EMG electrode placement as part of the data collection. However, EMG placement would present conflicting space requirement with the in-socket sensors. This space limitation was addressed by performing the experiment with the same transfemoral amputee participant in the earlier study [33] in which his gait muscle activation was documented and referred to in the study design of this paper [33].” (Line 370-379 second column)

Page 3, Line 202-204:
“This was based on the findings by Prendergast et al. [34] and Malik et al. [35] which reported that muscle activity happens in low frequency.”

The article [37] is about children's muscle activities. Can authors conclude the normal gait of children, in particular, muscle activity is similar while authors conclude the muscle activity depends on the person to person. Are children a good reference for muscle activity?

The article [37] is about children's muscle activities. Can authors conclude the normal gait of children, in particular, muscle activity is similar while authors conclude the muscle activity depends on the person to person. Are children a good reference for muscle activity?

While authors have realized the importance of relevant muscles in their own article published before, authors failed to verify the validity of muscle activity of the amputee's leg. The significant improvement for this article would have been the inclusion of muscle activity validation.

We thank the reviewer for this important note. We have included this key point in our discussion section as a limitation of the study and concern for future work.

Thank you.

Page 3, Line 202-204:
“This was based on the findings by Prendergast et al. [34] and Malik et al. [35] which reported that muscle activity happens in low frequency.”

Thank you for the suggestion. We have added the mentioned article as our reference in justifying our claim.

Page 3, Line 202-204:
“This was based on the findings by Prendergast et al. [34] and Malik et al. [35] which reported that muscle activity happens in low frequency.”

Thank you for the suggestion. We have added the mentioned article as our reference in justifying our claim.
Probably authors would have used as the reference the below article which is at least relevance to muscle activities of young adults during the normal gait;


rather just give some insight of which muscles are active during each gait phases.

However, we accepted the suggestion to add the recommended article to support our discussion.

Page 6, Line 405-411:
“In addition, it was also observed that the pattern of the signal response was consistent with the muscle activity for quadriceps and hamstring muscle group [35, 38] in which these two muscles are seen to be more active during LR phase, quadriceps muscle group is active during PSw and initial swing phase, and hamstring muscle group is active during the terminal swing phase for a normal person.”
Feasibility of A Gait Phase Identification Tool for Transfemoral Amputees using Piezoelectric-Based In-Socket Sensory System

Farahiyah Jasni, Nur Azah Hamzaid*, Tawfik Yahya Al-Nusairi, Nur Hidayah Mohd Yusof, Hanie Nadia Shasmin, and Ng Siew Cheok

I. Introduction

Abstract — Background: Gait detection is crucial especially in active prosthetic leg control mechanism. Vision system, floor sensors and wearable sensors are the popular methods proposed to collect data for gait detection. However, in active prosthetic leg control, a tool that is practical in its implementation and is able to provide rich gait information is important for effective manipulation of the prosthetic leg. Objective: This study aims to ascertain the feasibility of the piezoelectric based in-socket sensory system that is hypothesized to be practical in implementation and provide sufficient information, as a wearable gait detection tool for transfemoral prosthetic users.

Method: Fifteen sensors were instrumented to the anterior and posterior internal wall of a quadrilateral socket. One transfemoral amputee subject donned the instrumented socket and performed two walking routines; single stride and continuous walking. The sensors’ responses from both routines were analyzed with respect to the gait phases.

Results and Conclusion: The results suggested that the sensors’ output signal corresponds to the force components behavior of the stump while performing gait. All sensors were seen active during the first double support period (DS1). The anterior sensors were prominent during the initial swing (Sw), while posterior sensors were active during the terminal Sw. These findings correspond with the muscle activity during the respective phases. Besides, the sensors also show significant pattern during single support (SS1) and the second double support (DS2) phase. Thus, it can be deduced that the proposed sensory system is feasible to be used as a gait phase identification tool.

Index Terms — Gait analysis, sensory system, prosthetic piezoelectric, transfemoral

ESEARCH in gait analysis has been a subject of interest for a long time. It is commonly used in medical and healthcare application [1-4] as it offers the necessary information by analysing human locomotion. The outcome of gait analysis includes gait phase detection, human gait kinetics and kinematics parameters identification, and evaluation of human musculoskeletal functions. The tools used for such gait analysis has evolved increasingly and thus had improved in accuracy and mobility.

One of the most important components in gait analysis is the collection of the gait information. There are few popular methods that are used in collecting gait information. The first one is camera-based/vision system. Single or multiple cameras were used to capture series of moving images in order to retrieve the gait parameters [4-7]. Camera or vision system is typically set up in the lab, in which the camera system is connected to a program that analyses the video captured by the camera. The advantage of this system is in its precision and accuracy, because the data is collected in a controlled environment. However, the controlled environment can also be a drawback in terms of its lack of practicality and restriction to represent real-life environment such as outdoor terrain. Moreover, the allowable distance for data collection is rather limited.

Another way to analyse gait is by using sensors instrumented onto the floor, also known as floor sensors [3, 8-10]. These methods are usually used to retrieve parameters related to ground reaction force (GRF), which could not be extracted using the vision system. These systems, i.e. the instrumented floor sensor and the camera system, are often used together in synchrony to complement each other in an integrated system, thus providing more complete information about the gait parameters, especially the joint kinetics. Studies have generally regarded such 3-dimensional motion analyses using camera systems incorporated with floor sensors as the ‘gold standard’ in gait and motion analysis.

However, in order to overcome the issue of limited mobility, wearable sensors, such as electromyography (EMG), Inertial Measurement Unit (IMU), and pressure sensors [11-13] are becoming a more favoured option due to its advantage over the camera system and floor sensors approach. They are more practical to be adopted outside the laboratory setting with greater movement distance and allows use in a more natural setting [14]. Wearable sensors are placed or attached onto
A. Subject and Prosthetic Leg

One healthy male, transfemoral amputee participated in this study. The description of the subject is detailed in TABLE 1. Subject provided his written and informed consent by signing the consent form. Approval for the experimental procedure was obtained from the Medical Research Ethics Committee of University of Malaya Medical Centre. Since the main aim of this study is to test the feasibility of the proposed wearable in-socket sensory system in detecting gait phases and its transition, the recruitment of one subject was deemed sufficient [26]. The socket used was polypropylene quadrilateral type and was custom-made to perfectly fit the subject. The sensors were instrumented to the socket following the configuration described in Jasni et al [33]. The hydraulic knee joint and Solid Ankle Cushion Heel (SACH) foot (3R60 EBS Pro, Ottobock, US) was used by the amputee. Prior to the experiment, the subject was given time to familiarize walking with the provided prosthetic leg to ensure that the captured data was as close as possible to his normal gait.

The experiment was conducted in the motion analysis lab, which was equipped with two force plates to capture the subject’s GRF data. The sampling rate of both force plates was set to 1 kHz. A motion analysis software (VICON Nexus version 1.8.5) was used to process the force plate data.

B. Piezoelectric-based In-Socket Sensory System

Fifteen polymer-based, PVDF-type piezoelectric sensors were mounted in zig-zag orientation with its negative surface facing upwards to ensure uniform polarity of the response. Each of the sensors was mounted using cantilever with elastic foundation to the anterior and posterior inner wall of the socket, so that the sensors will touch the most active area of quadriceps and hamstring muscle [33]. Fig. 1 shows the in-socket sensory system adopted in this work and Fig. 2 illustrates the relationship between the strain and the response (i.e. voltage) of the sensor when mounted in this configuration, whereby the deflecting end of the sensor was positioned at the centre of the muscle bulk, and the fixed end of the sensor placed at the outer edge of the muscle bulk.

In order to calibrate the sensors, the response during the quiet standing activity was recorded for all sensors. Fig. 3 shows the response of all sensors for quiet standing, with response range within -0.02 to +0.02V. During quiet standing, all sensors behaved similarly with output signals not different from one to another (p = 0.163).

<table>
<thead>
<tr>
<th>Age</th>
<th>36 years old</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height</td>
<td>174.5 cm</td>
</tr>
<tr>
<td>Body Mass</td>
<td>80.5 kg</td>
</tr>
<tr>
<td>Condition</td>
<td>Unilateral amputation (left leg)</td>
</tr>
</tbody>
</table>

- Has Been wearing prosthesis for 17 years

II. METHODS

Experiment was done to collect sensors responses during quiet standing from a TF amputee, and the output signals of all sensors were analysed to see the pattern and trend of the signals of each sensor for each gait phases. The signals variation across multiple trials were calculated and particular sensors with dominant response for each phase were identified.
C. Data Acquisition and Signal Processing

The output signal from the sensors was first pre-processed, amplified, and low-pass filtered before sent to the DAQ card for analogue-digital conversion. Each sensor was connected to an active low pass filter circuit with a voltage gain of 11 and the cut-off frequency of 800 Hz. Two 3 meters-long ribbon cables were used to connect the sensors to the signal conditioning circuit. The output signal from the circuits was passed to two similar DAQ card units (NI 9221, National Instruments, USA) with sampling frequency of 1 kHz.

In order to get a smoother signal, the acquired data was again band-pass filtered in the LabVIEW software by sixth-order Butterworth Infinite Impulse Response (IIR) filter with a passing frequency of 1 to 20 Hz. This was based on the findings by Prendergast et al. [34] and Malik et al. [35] which reported that muscle activity happens in low frequency.

D. Experimental Procedure

The experiment was divided into two routines. The first one was referred as the single stride walking routine and the second one was the continuous walking routine. During the single stride walking routine, data from the sensor and the force plates were collected simultaneously. Thus, the relationship between the force and the sensors’ response can be observed. Meanwhile, the continuous walking routine recorded a natural continuous walking data. Only sensory data was collected in this routine.

The rationale for having separate routines was to eliminate the need for long wire to be used to connect the sensors to the signal conditioning circuit. Pre-investigation was conducted that determined the optimum length of the cable to reduce noise caused by the cable was 3 m [36].

1) Single stride routine

The subject performed a single stride over the force plates. The starting point was set outside of the first force plate where the subject stood quietly and one force sensor was used as the trigger switch. It was placed on the second force plate and wired to the DAQ card module. The procedure for this routine is illustrated in Fig. 4(a) and the walking procedure in Fig. 4(b). Each trial was repeated 10 times.

The first trigger switch was used to synchronize the starting point of the routine for both systems. The sensor switch, placed on the force plate, produced instantaneous and simultaneous response with the force plate when knocked, thus was used as the signal synchronization marker. Fig. 4(c) illustrates the response for both systems during the start of the routine. Both systems (i.e. force plate and sensors data collection) data were synchronized by identifying the trigger point impulse. The x-axes of that two systems’ plot were considered synchronized after the starting point was determined.
Thereafter the force plate response, which actually represents the Ground Reaction Force (GRF) for the subject, were used as the indicator to determine the gait cycle. The point where the first force plate data started to raise was marked as the first heel strike. Meanwhile, the point of abrupt changes in the second force plate was considered as the second heel strike point, thus marked 100% gait cycle of the prosthetic leg. The phases were then further identified based on the GRF signals. The process was illustrated in Fig. 4(d). The data for GRF and sensory data for single stride was cropped accordingly for all 10 trials and the mean and standard deviation was calculated.

2) Continuous walking routine

One sensor was used as the trigger switch to mark the start and end of the routine. The procedure of the routine is illustrated in Fig. 5(a). A semi-circle path was used to compensate the wire-length limitation. However, it was assumed that the subject walked in a straight path manner, because the ratio of the path circumference to the stride length was larger than 18 (18.85:1), which led to the assumption that the curved path did not affect the subject walking pattern. Five trials were performed with an average of 6 complete continuous strides performed in each trial.

To mark the gait events during the gait, a video throughout the trial was recorded and analysed using Kinovea [37] to confirm and adjust the gait phase’s identification. Data for single stride was cropped accordingly. The mean and standard deviation of 30 complete gait cycles (5 trials × 6 strides/trial) for each sensor were calculated. Data from both routines were normalized to 100% gait cycle.

The significant sensors were identified based on the response of the sensor during the respective gait phase. Two criteria that the sensor must possess to be significant are; 1) at least one slope sign change, and 2) at least one peak with amplitude > 0.02V>Peak, or Peak>0.02V, during the respective phase.
A. Single Stride Walking

The equation to calculate the mean and standard deviation was described in (1).

\[
\text{Mean}(t) = \frac{\sum \text{SensorResponse}(t)}{n}
\]

\[
\text{StdDev}(t) = \sqrt{\frac{\sum | \text{SensorResponse}(t) - \text{Mean}(t) |^2}{n}}
\]

where:
- \( n \) = number of trials, and \( t \) = gait percentage

![Fig. 5: Experimental setup and procedure for continuous walking routine.](image)

**TABLE 2**

<table>
<thead>
<tr>
<th>Gait phase</th>
<th>Single stride walking</th>
<th>Continuous walking</th>
</tr>
</thead>
<tbody>
<tr>
<td>LR (DS1)</td>
<td>0% - 23%</td>
<td>0% - 19%</td>
</tr>
<tr>
<td>MST (SS)</td>
<td>23% - 35%</td>
<td>19% - 33%</td>
</tr>
<tr>
<td>TSt (SS)</td>
<td>35% - 46%</td>
<td>33% - 48%</td>
</tr>
<tr>
<td>PSw (DS2)</td>
<td>46% - 65%</td>
<td>48% - 62%</td>
</tr>
<tr>
<td>Sw</td>
<td>65% - 100%</td>
<td>62% - 100%</td>
</tr>
</tbody>
</table>

Note: HS-Heel Strike, LR-Loading Response, DS1-1st Double Support, MSE-Midstance, SS-Single Support, TSt-Terminal Stance (TSt), PSw-Pre-Swing, Sw-Swing.

B. Continuous Walking

![Image](image)

The graph represents the results for both routines are shown in Fig. 8. The gait phases can be characterized and eventually described in (Table 2). The graph that represent the results for both routines are shown in Fig. 8.

For instance, for sensor P2, the positive peak for single stride walking occurred at ~10% of the gait cycle, but for continuous walking, it happened at ~20% of the gait cycle. For instance, the calculated SD at 9% (LR phase) of the gait cycle is ~1.5 times smaller than the SD calculated at 83% (Sw Phase) for Sensor A5 for single stride, but ~0.9 times smaller for continuous walking which indicates, that the SD for continuous walking is more uniform.

Second, the response pattern of most of the sensors showed similarity for both types of walking, but, shifted in time domain. For instance, for sensor P2, the positive peak for single stride walking was smaller than the continuous walking for almost all sensors. For example, for sensor P2, the positive peak for single stride walking was higher than the response pattern of most of the sensors showed similarity for both types of walking, but, shifted in time domain. For instance, for sensor P2, the positive peak for single stride walking was higher than the continuous walking routine with its respective standard deviation.

C. Continuous vs Single stride

Fig. 7 presents anterior and posterior sensors response signal for the continuous walking routine with its respective standard deviation. Overall, it can be seen that the standard deviation for continuous walking signals are more uniform throughout the phases. Nevertheless, during the first double support (LR phase), it can be seen that almost all sensors behaved actively. For the single support phases (MSt and TSt), the proximal and middle anterior sensors (A1, A2, A3 and A4), and P3, P5 and P7 (for posterior sensors) displayed a more prominent behaviour. During the second double support phase (i.e. PSw phase), A1, A2, A5 and A6 (for anterior sensors) and some posterior sensors (P2, P4 and P6). Finally, during Sw phase, all sensors showed strong responses.

**IV. DISCUSSION AND CONCLUSION**

Piezoelectric is a type of dynamic force sensor that detects the changes in force or pressure on the measured area with respect to time. In this application, an array of piezoelectric sensors was used to capture the force profile of the stump while it is inside the prosthetic socket and doing gait. It was hypothesized that there is certain force profile that is consistent every time the prosthetic user performs gait and piezoelectric sensor can translate it in terms of voltage signal. By analysing the signal and studying its behaviour for the respective gait phases, the gait phases can be characterized and eventually
Based on the findings of this study, it can be deduced that the proposed in-socket sensory system could produce consistent signal for most of the gait phases, especially the stance phase (LR, MSt and TSt). It is believed that higher deviation was noted on the swing phase for single stride routine because of the higher deviation in the swing phase. During single stride routine, the subject was asked to perform only one stride and walking and stop. Thus, the swing might not be as natural as it should because the subject is already ready to stop. On the other hand, the deviation in continuous walking routine was seen be more uniform throughout the gait cycle as the signal was captured while the subject performed multiple strides. Thus, motion is more natural and repetitive.

The responsibility for the difference between the sensors response for the single stride walking and the continuous walking was due to the speed and momentum during walking motion. For continuous walking, the subject walked a higher speed than the single stride walking, potentially due to the momentum gained throughout the continuous walking. Thus, higher amplitude was detected for continuous walking due to the momentum effect. The difference in the walking speed from one step to another might also contribute to the higher standard deviation in some phases of the continuous walking. The difference between the two walking types led to the deduction that the in-socket sensory system can also detect the difference in the walking type or speed by the amputee. However, more studies have to be planned and conducted further to verify this new hypothesis.

From the single stride routine results, it was observed that some sensors’ behaviour actually corresponded to the GRF components during the gait especially during LR and MSt phase. Some similarities were observed in sensors response and the AP-GRF component. This agrees with this study hypothesis that there are three major force components that contribute to the force profile of the stump, which are: the GRF, the interface force between the socket and the stump, and also the muscle force. Thus, some similar behaviour with the GRF components are expected.

In addition, it was also observed that the pattern of the signal response was consistent with the muscle activity for quadriceps and hamstring muscle group [35, 38] in which these two muscles are seen to be more active during LR phase, quadriceps muscle group is active during PSw and initial swing phase, and hamstring muscle group is active during the terminal swing phase for a normal person. The sensors signal showed agreement with this fact, where both anterior and posterior sensors were seen active during LR phase, and anterior sensors were prominent during the early stage of swing (60-80% of gait cycle) and posterior sensors were more active at the later stage of swing phase (80-100%). However, during PSw phase, it can be seen that some posterior sensors also displayed prominent response. One of the reasons that might contribute to this outcome is the effect of amputation. The muscle activity of a TF amputee, might be different from the one displayed by the normal person, and can also be different between two TF amputees [20]. Because of that fact, the authors deduced the pattern of output signal of this sensory system might vary from one user to another user, depending on the user’s stump’s length and condition, and muscle strength of the user. Therefore, the sensory system needs to be calibrated for each user and there is no one general template that can be used for all users. The calibration process includes Maximum Voluntary Contraction (MVC) evaluation so that the active muscle area can be identified, and sensors are placed on the active area. This is only needed once for each user. The same placement can be used for the same user if the socket must be changed. The calibration process can be done during the training sessions with the physician at the beginning of the prosthesis use.

As a limitation, this study did not explicitly measure muscle activation which would require EMG electrode placement as part of the data collection. However, EMG placement would present conflicting space requirement with the in-socket sensors. This space limitation was addresses by performing the experiment with the same transfemoral amputee participant in the earlier study [33] in which his gait muscle activation was documented and referred to in the study design of this work [33].
Although separate studies are needed to confirm the relationship between each force component and the sensors' output signal quantitatively, this study has proven the feasibility of extracting the information. Besides, this study also shows that this proposed wearable sensory system has the potential to provide more information (i.e. muscle and GRF information), which cannot be achieved by gait kinetic based only method [39, 40] or EMG based only method [16] in one sensory system.

ACKNOWLEDGMENT

The authors thank Mr Suparjo for his contribution as the subject in this study and also Performance of Body and Analysis of Movement Lab, University of Malaya for their permission to use the facilities for the experiment in this project.

REFERENCES

