

A Shoe Mounted System for Parkinsonian Gait Detection and Real-Time Feedback

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Abstract. Conditions like Parkinson's disease (PD) remain largely a mystery in the way that they affect individuals even under today's modern medical practices. One of the main secondary effects associated with PD can be seen in issues with the individual's gait and is referred to as Freezing of Gait (FoG). The symptom often responds poorly and sometimes paradoxically to treatment with dopaminergic medication that is traditionally used to treat the other symptoms of PD. However, a linkage found that FoG, during walking, results when the sequence effect is superimposed on a reduced step length. Prior research has focused on the development of technologies that use audio or visual feedback to help the individual adjust their gait. These systems may not be deployable in real-world environments since people rely on sight and sound for navigation. This research proposes the development of a system to measure step length in real-time and to provide haptic feedback to offset the progression of FoG episodes.

Keywords: Adaptive interfaces · Anticipatory interfaces · Mobile HCI · Context-dependent system

1 Introduction

Conditions like Parkinson's disease (PD) remain largely a mystery in the way that they affect individuals even under today's modern medical practices. According to the Parkinson's Disease Foundation, an estimated 7–10 million people worldwide are living with PD [1]. Within the United States, approximately 60,000 people are diagnosed with the disease each year and have a combined direct and in-direct cost of \$25 billion per year. Parkinson's disease is a neurodegenerative brain disorder that usually progresses over a long duration (often around 20 years) within an individual. While the disease itself is not fatal, the Center for Disease Control rated complications from the disease as the 14th top cause of death in the U.S. [2]. Because of these staggering

numbers, there has been a national push toward conducting fruitful research on PD. There are now multiple treatment methods to offset the symptoms of the disease—i.e., medication and an invasive procedure known as deep brain stimulation (DBS).

The symptoms of PD can be categorized into three main groups: primary motor symptoms, secondary motor symptoms and non-motor symptoms. Primary motor symptoms include tremor, bradykinesia, rigidity, and postural instability, which often have secondary motor symptoms. One of the main secondary effects associated with PD can be seen in issues with the individual's gait and is referred to as Freezing of Gait (FoG). It is an episodic phenomenon that is unexplained by rigidity or bradykinesia, and is more common in advanced PD. The episodes occur more frequently during high pressure or timed tasks such as walking down a narrow hallway or crossing a street in a given amount of time, and can have a large impact on an individual's quality of life since independence is diminished. The episodes are generally characterized by the individual feeling that they are unable to move or that their feet are "glued to the floor," and can often result in a fall. Because this secondary symptom is still not well understood, there does not exist effective treatment options to address it. The symptom often responds poorly and sometimes paradoxically to treatment with dopaminergic medication that is traditionally used to treat other symptoms of PD. However, a linkage found that FOG, during walking, results when the sequence effect is superimposed on a reduced step length [3].

Because of the dangers of this symptom, much research has been done in this area to offset the episodes and to prevent them from occurring as frequently. However, the majority of this research has focused on the development of technologies that use audio or visual feedback to help the individual adjust their gait. These systems may not be deployable in real-world environments since people rely on sight and sound for navigation. This research proposes the development of a system to measure step length in real-time and to provide haptic feedback to an individual to offset the progression of FOG episodes. Specifically, we are looking to extend the existing model for normal human gait to detect the abnormal cycles in Parkinsonian gait. This will involve the development of new, adaptable algorithms that can accommodate variations that are based on human-centric characteristics. Secondly, this research looks to develop a ubiquitous system that can monitor day-to-day activities and provide real-time feedback to the user to predict and offset FoG episodes.

2 Related Work

Immediately before the onset of a FOG episode, one of the main physiological events that is observed is a profound and incremental decrease in stride length [3, 4]. Thus, many researchers have attempted to identify these physiological events and to warn the individual before the actual FOG episode occurs. Prior work has found that if the user is mentally aware of the shortening of stride lengths, then they can adjust their gait accordingly. Current approaches have looked at different ways of notifying the user that their stride length has become increasingly small including through visual feedback [5] and audio feedback [6]. These devices provide cues (audio or visual) to guide the user in their step rhythm and cadence, and to make them aware of the need to take

longer steps. With these cues, the user can change their behavior pattern and often avoid the FOG episode. However, the main limitation of these approaches is that they rely on technology that would be very difficult for users to wear and use on a day-to-day basis. These technologies obstruct either the user's vision or hearing, which can be dangerous in a navigation situation. Another limitation is that the algorithms that were developed for these devices rely on making slight modifications to the models used for the normal gait cycle, and therefore, do not necessarily accurately represent Parkinsonian gait.

3 Proposed Approach

The proposed approach has two main objectives:

1. Create a new model for Parkinsonian gait that models the cycles better than the traditional approach of modifying a normal gait model.
2. Develop a minimally intrusive device that can be worn by an individual during daily activities.

3.1 A New Model for Parkinsonian Gait

We begin by developing an accurate model to represent Parkinsonian gait characteristics. This is a challenging problem in that Parkinsonian cycles, immediately before and during a FOG event, do not follow regular gait cycles, and thus distinct features and events will need to be identified within this new model. Traditional detection of heel-strike and toe-off events that occur in normal gait should be avoided because these characteristics are often degraded beyond recognition in patients with PD. We therefore consider analysis on the frequency domain as the main method of step detection, and use pressure shifts between the toe and heel as a validation method, rather than detection, to develop a model for start and stop events. We will apply this model to an assistive device that the user can wear everyday with minimal intrusion.

3.2 A Minimally Intrusive Device

To achieve the goal of developing a device that can be used in everyday environments, the device needs to be relatively small and should be easily embeddable into clothing that the user already wears. Furthermore, it should not be significantly heavy since it will be adding weight to each of the user's feet, and fatigue while walking can be a major concern. We propose the development of a device that can be worn on the user's sock and can be easily hidden using pockets over the sock. The batteries for the device will be worn over the shank to decrease the amount of weight carried by the foot since these are the heaviest pieces.

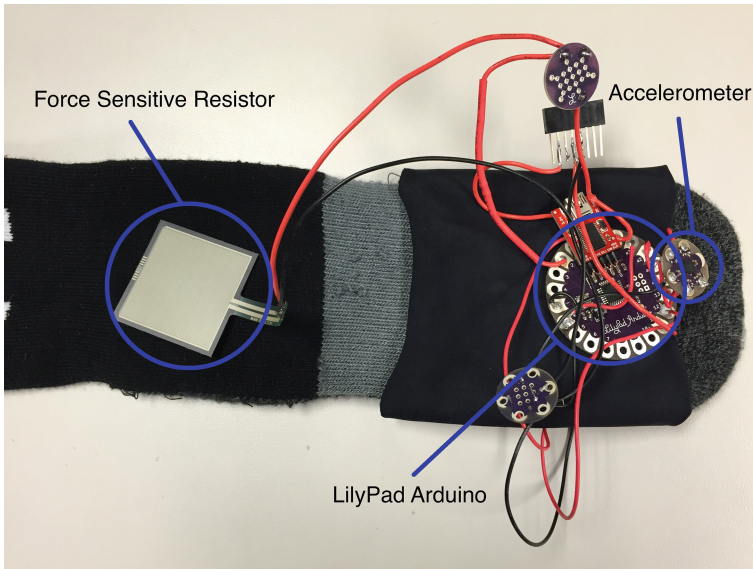


Fig. 1. Components of wearable system

4 Implementation

A wearable system has been built that consists of two small devices that can be worn over each of the user's socks (Fig. 1). The devices consist only of a LilyPad Arduino 328, a LilyPad accelerometer, an XBee module, and a force sensitive resistor. Processing is currently being done offline; however, in future iterations, these devices will integrate with the user's smartphone as a user interface and as a means for actuation. The device was built using the Arduino LilyPad board for easy embedding into the sock. The accelerometer is worn above the toe since this was the location that was found to have the least angular displacement during toe-off events, and reduced the distortion of the signal during those events. The pressure sensor is worn under the heel and is used as a method of validating steps taken by the user. The pressure data relies on a threshold value that will be person-centric and easily adjustable depending on the user. The device streams data for both acceleration and pressure that is annotated with timestamps to a computer for storage. Once the gait data is collected, it is processed to first identify which events were steps and then to determine the length of each of those steps.

To reconstruct the signal to counter the distortion caused by toe-off acceleration, the following method was implemented. It was assumed that only the toe-off portion of the signal was significantly distorted, so the toe-off portion of the signal was reconstructed based on the rest of the signal. The signal was approximated as a sinusoid with one crossing of the time axis between toe-off and heel-strike (one full step). Given that the acceleration and velocity of the foot at toe-off and heel-strike are known to be zero, the area under the acceleration curve over the course of one step is also known to be zero.

For each step, the point at which acceleration crossed from positive to negative was labeled as t_m , thus defining the portion of the signal which must be reconstructed to be between t_0 (toe-off) and t_m with the signal from t_m to t_f (heel strike) assumed to be accurate. The region between t_0 and t_m was thus reconstructed as a sine signal with t_0 mapped to zero and t_m mapped to $\Delta t = (t_m - t_0)$. Thus, given the signal $f(t')$ represents the reconstruction of the signal, it can be defined as follows:

$$f(t') = C \sin\left(\frac{\pi}{\Delta t} t'\right), t' \in [0, \Delta t]$$

where C is the amplitude. Given that the area under the acceleration for one step must be zero, the magnitude of the area under the curve from t_0 to t_m must be equivalent to that from t_m to t_f . Thus, letting A represent the area under the curve from t_m to t_f ,

$$-A = \int_{t_0}^{t_m} C \sin\left(\frac{\pi}{\Delta t} t'\right) dt' = -C \left[\frac{\Delta t}{\pi} \cos\left(\frac{\pi}{\Delta t} t'\right) \right]_{t_0}^{t_m}$$

Solving for C and substituting this equation into $f(t')$ and mapping t_0 to zero and t_m to Δt ,

$$f(t') = \frac{A\pi \sin\left(\frac{\pi}{\Delta t} t'\right)}{2\Delta t}, t' \in [0, \Delta t]$$

To implement this, the area under the curve from t_m to t_f for each step was calculated and substituted for A , and $f(t)$ was calculated to replace the signal from t_0 to t_m for each step, evaluating the function for $t' = t - t_0$ to correctly map the values.

5 Early Evaluation

Two major components have been assessed: the accuracy of detecting when a step has been taken and the accuracy in step length calculation for that step. These components were assessed for an individual with regular gait so that a baseline could be determined for the validity of distance calculation using a single accelerometer. Ten trails were conducted where a user walked 8 steps forward, turned around, and then walked 8 steps back. Each step was manually measured and logged for distance. Step length was measured using the distance between heel-strike events on a single foot. As shown in Fig. 2, the pressure values validated the starting and stopping points for each step. The acceleration curve follows what we would expect to see for normal gait and allows us to visually identify where the heel-strike and toe-off events are occurring as well as adequately showing the mid-swing. The system was able to accurately detect every step event within the data, however, was limited in accuracy with respect to the distance calculated. This was due to the posture of the foot during mid swing that we do not account for with a single accelerometer approach. We will be modifying our proposed system to use an IMU to solve for this issue in accuracy since we will be able to get angular acceleration from the foot and shank.

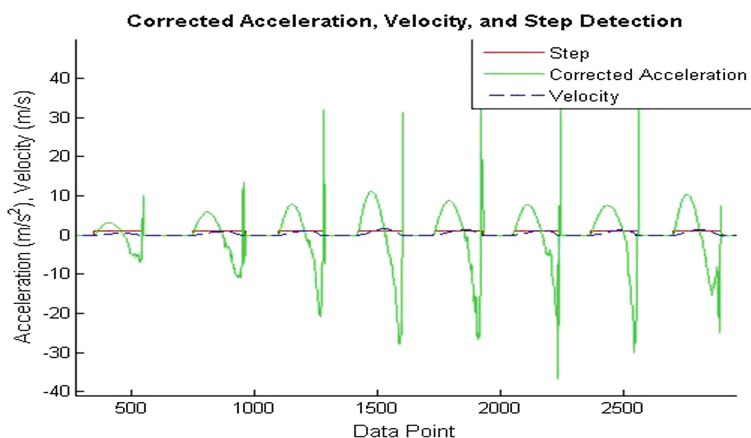


Fig. 2. Corrected acceleration, velocity and step detection graph

6 Conclusion and Future Work

Although much more evaluation needs to be done, preliminary results have shown that this system does have promise for being able to detect asynchronous steps. Future work will augment the single-accelerometer approach to include an IMU to more accurately determine angular acceleration of the foot and thus make the distance calculation much more accurate. We will test the effectiveness of site-specific haptic stimulation as well once step length is more accurately determined since recent research has shown its effectiveness [7]. We hope that this system will become an effective tool for the anticipation and prevention of FOG episodes.

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