Estimating Gait Asymmetry Using Wearable Sensors

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***Abstract*— Below the knee prosthetics are expensive, especially in growing children. A majority of this cost is due to the expertise of the prosthetist. By developing a lost cost wearable sensor, we hope to democratize this knowledge, reducing the total cost for prosthetic patients by allowing them to make their own adjustments. In this paper, we used two IMU sensors to estimate three walking parameters: step length, walking frequency, and contact time. We use these three parameters to determine gait asymmetry and to compare our results with a pressure sensitive treadmill.**

I. Introduction

Current below-the-knee prosthetic limbs are expensive for children because as the child grows, they must visit the prosthetist every six months to evaluate their prosthetic limb. Designing an affordable prosthetic solution while accounting for the variance in children's sizes is a challenging problem.

Children's prosthetics must be changed frequently as the child matures because children change size and weight very rapidly as they are developing. From age 6 -16, a childs height can range from 112 to 178 cm and can weigh from 15 to 66 kg [1]. If the prosthetic is not changed frequent enough, the poor fitting prosthetic will lead to gait asymmetry, producing pain in the lower back and other parts of the body. In addition, there is an increased metabolic energy cost associated with gait asymmetry.

Parents want their child to have an active lifestyle similar to other children but may have limited finances to account for frequent visits to the prosthetist, therefore, a low cost solution is needed. Prosthetic legs can cost between \$5,000 to \$50,000 [2], which puts them out of reach for many patients. Recent developments in 3D printing have demonstrated a promise for manufacturing functional low-cost prosthetic hands [3]. However, little work has been done for developing low-cost solutions for below-the-knee amputees.

While many have attempted to make functional low-cost prosthetics, they fall short in addressing the true cost of prosthetics: the knowledge and experience of a prosthetist. The research done on attaching a sensor on a human body to analyze human walking gait has been done [4][5] where IMUs were placed on three areas: thigh, calf and feet. The sensor data taken from each legs can be compared to adjust the prosthetic leg and achieve gait symmetry, which can improve comfort and mobility.

Here we present a low cost wearable set of sensors to detect and understand gait asymmetry to make informed decisions when developing a fully adjustable prosthetic limb. From this data, we are able to quantify gait asymmetry and make informed decisions when developing an adjustable prosthetic limb.

Fig. 1: The IMU placement, the measurement axis, and the spacer.

II. Hardware and Sensor

We used Arduino Yun and two IMUs (Adafruit BNO055 9-DoF IMU), which were attached on top of the toes as shown in Fig. 1. We chose this IMU placement because the toe tends to have a fixed orientation when in contact with the ground and the IMUs can be more rigidly attached to the shoe. The Arduino is used to communicate and collect the IMU data which was sampled at a rate of 100 Hz. The IMU were used to collect the orientation and acceleration of the toe which can be used to determine three important gait parameters: step frequency, step length and foot contact time.

A. Step Frequency

The step frequency is the number of steps done in one second. Since walking is a periodic behavior, we can estimate the step frequency by performing spectral analysis on the data and observing the power spectrum. We found that this method is very robust and works on angle, acceleration, and gyro data parameters. Fig. 2 shows the raw angular rotation data of the IMU's x-axis. Fig. 3 shows the power spectrum of the gyro data. In this example, we can see the dominant frequency occurs at 0.8291 Hz.

B. Step Length

The step length is estimated by double integrating the acceleration data about the fixed coordinate frame (lab coordinate) in the x-direction [6]. We can calculate this by doing the following:

- Filter the data by using a moving average filter ($N = 4$).
- Eliminate gravity vector by substracting the data with the calibration value.
- Rotate the acceleration vector with the orientation angle from Fig. 2.
- Double integrate the acceleration in the x direction.

C. Contact Time

The contact time measures how long the feet contacted the ground. We can measure this by calculating the time

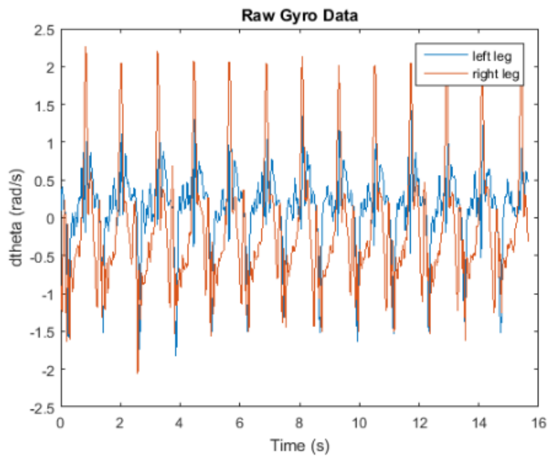

Fig. 2: Angle measurement of the toes

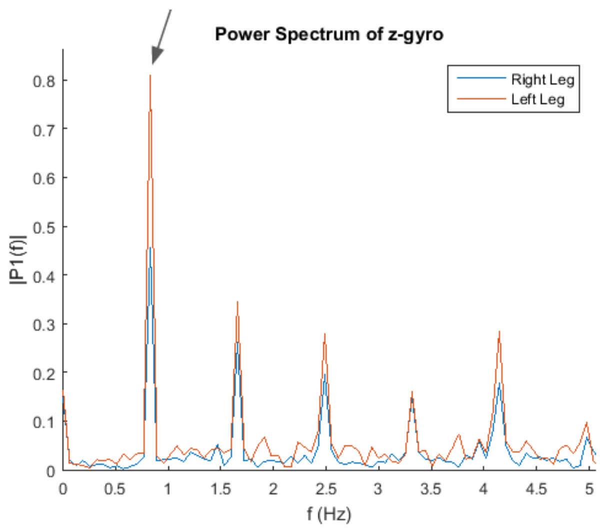

Fig. 3: FFT of the angle data in Fig. 2

between the lowest peak and the highest peak of the angle measurement data as shown in Fig. 4.

III. Experimental Results

We used a C-Mill 3N treadmill to validate our wearable sensors. The C-Mill treadmill contains a large pressure sensor plate underneath the treadmill belt which is used to detect the force and location of each step. From these it calculates several gait parameters including step frequency, step length, contact time, which is used to determine gait asymmetry. We collected C-Mill data and wearable IMU data simultaneously to compare our wearable sensors accuracy.

We conducted two experiments: normal walking, and walking with leg length discrepancy. To simulate leg length discrepancy, we attached a 3mm spacer below the right shoe, as shown in Fig. 1.

A. Normal Walking Experiment

Results from normal walking experiment can be seen in table I. Looking at our treadmill's asymmetry values, we

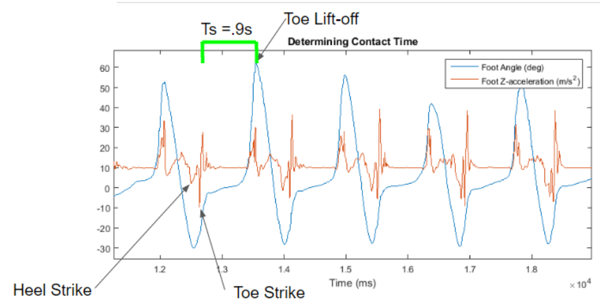

Fig. 4: How to measure the feet contact time.

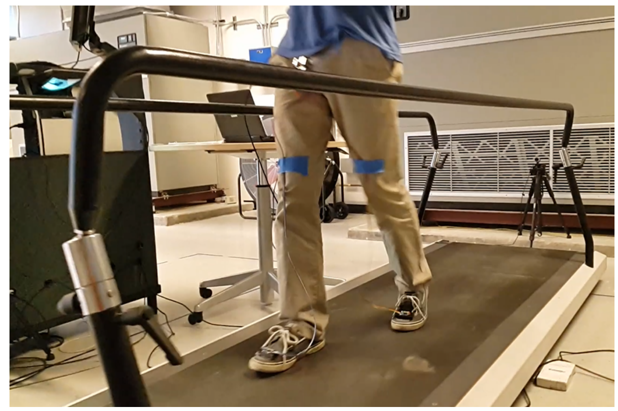

Fig. 5: The treadmill experiment

can see that our test subject exhibits a nearly symmetrical walking gait. In addition we see good agreement between treadmill and wearable sensors for contact time and step frequency. However, our step length estimate was way off.

B. Leg Length Discrepancy Experiment

Results from our leg length discrepancy experiment can be seen in table II. It appears that the greatest asymmetry induced by the 3mm spacer is step length: a 12.2% difference. Again, our wearable sensor was unable to measure the step length accurately. However, it was still able to correctly estimate the step frequency and contact time.

IV. Conclusion and Future Work

Through our experiments we were able to show that our wearable IMU sensors placed on the toe of the shoe are sufficient to estimate the contact time and step frequency of normal and slightly asymmetric walking gaits. Our "peak to peak" contact time estimate error was less than 6%, and our

Normal Walking			
	IMU Result	*Treadmill Result*	*% Difference*
Step Frequency	0.925 Hz	0.952 Hz	2.96%
Step Length (L)	-0.730 m	0.540 m	xxx
Step Length (R)	0.293 m	0.524 m	xxx
Length Asymmetry	140%	-2.8%	
Contact Time (L)	1.40 s	1.33 s	-5%
Contact Time (R)	1.44 s	1.41 s	-2.1%
Time Asymmetry	2.93%	6.2%	

TABLE I: Normal Walking Experiment Results

3cm Spacer Walking			
	IMU Result	*Treadmill Result*	*% Difference*
Step Frequency	1.4 Hz	1.4 Hz	0%
Step Length (L)	0.0196 m	0.563 m	xxx
Step Length (R)	0.223 m	0.641 m	xxx
Length Asymmetry	1030%	12.2%	
Contact Time (L)	1.01 s	0.95 s	-5.94%
Contact Time (R)	0.973 s	0.95 s	-2.36%
Time Asymmetry	-3.80%	0%	

TABLE II: Leg Discrepancy Experiment Results

spectral analysis for determine the walking frequency was extremely accurate and robust. We also discovered that step length is the most useful metric to assess gait asymmetry. Unfortunately, our double integration method to estimate step length failed. We believe that our acceleration data was too noisy and too excited in order to accurately double integrate. Even if we address those issues, we have low confidence that this simple method will provide us an accurate enough estimate. Regardless, we believe that our initial results have given us good insight and intuition on low cost wearable sensors. We believe that this was a good first approach towards this problem, and that with better sensor placement and by developing a kinematic model of the leg, we will be able to improve our estimates.

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