A NEURAL PROSTHESIS TO IMPROVE GAIT IN PEOPLE WITH MUSCLE WEAKNESS

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

ABSTRACT

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Functional Electrical Stimulation (FES) can be used to induce contractions in muscles that do not receive adequate signals from the nervous system. People who have lost function in their foot and/or ankle following a stroke may benefit from a neural prosthesis. In this project, we designed, built and tested a neural prosthesis to assist people with muscle weakness who are at risk of falling. The neural prosthesis was designed to stimulate the gastrocnemius (GN) muscle during the push off phase using a manual switch. The neural prosthesis included a FES unit for muscle stimulation, electronic circuitry, and sensors including inertial measurement units (IMUs) and foot pressure sensors to detect movement characteristics. The neural prosthesis was tested on a healthy individual who walked in a straight line while the neural prosthesis collected data. Tests were performed with and without the neural prosthesis activated. The neural prosthesis was able to successfully collect gait data and cause contraction in the GN muscle. The test results showed that the GN muscle stimulation changed the gait by inducing a plantarflexion movement on the foot and expediting the toe off event. The performance of the neural prosthesis was evaluated using a commercial camera motion capture system. The IMUs and the motion capture system had and an average correlation coefficient around 95% which was close to some

literature. This research showed that a neural prosthesis utilizing low cost IMUs was able to estimate the joint angle while walking. In addition, the device had an observable effect on the gait when tested on a healthy individual.

CHAPTER1: INTRODUCTION

Neural prostheses can be used to assist people with spinal cord injuries (SCI). In the United States, there are currently 288,000 people with SCI and there are 17,700 new SCI in the U.S.A. every year [1]. The estimated average yearly expenses for an individual with paraplegia is more than $537,271 in the first year and $71,172 each subsequent year [1]. This estimation does not include indirect costs such as losses in productivity and wages. Considering the costs of this injury, helping these individuals can be beneficial to society.

Normal muscular contractions are caused by electrical signals from the nerves attached to them. In able bodied people, these electrical signals are generated by the nervous system [2]. Some neurological disorders cause the connection between the nervous system and muscles to be interrupted. However, it is possible to generate these electrical signals artificially. The method which utilizes electrical impulses to contract the muscles and induce body movement is called Functional Electrical Stimulation (FES) [3]. This method has been employed in neural prosthetics to assist people with neurological disorders such as a spinal cord injury and strokes [3]. Examples of FES in neural prosthetics include devices that assist people in walking [3] and moving their arms [4] by stimulating the lower and upper limb muscles respectively. FES has also been utilized in the short-term therapy in an attempt to restore motor functions [5].

The overall goal of this line of research is to use FES to improve the gait. There are multiple muscles that contribute to gait but the research team chose to actuate only one muscle for simplicity. The ankle and hip joints use the majority of the energy in walking [6]. The hip joint has more degrees of freedom than the ankle which makes it more complicated to stimulate properly. Consequently, the research team chose to focus on the actuation of the ankle joint.

There are some assistive devices commercially available to prevent foot drop [7], so the research group decided to work on the muscles associated to plantarflexion motion. The soleus muscle has an internal position which makes it difficult to use surface electrodes to stimulate this muscle. As a result, the gastrocnemius was chosen to get stimulated in this project.

The purpose of this project is to help people with muscle weakness due to injury or disease improve their ability to walk. The goal of this research is to develop a neural prosthesis capable of stimulating the gastrocnemius muscle to improve gait while walking. This research is the first part of an overall goal to develop a neural prosthesis capable of contracting the gastrocnemius muscle at the appropriate time utilizing sensor feedback.

CHAPTER 2: LITERATURE REVIEW

2.1 Bipedal Locomotion

There are different gaits in human locomotion, such as walking, running and hopping. A neural prosthesis is required to work for all different gaits in order to be helpful in real life. This project will focus on the walking gait as the first step toward the development of the neural prosthesis. Walking is the first gait to be considered, because of its simplicity and being the most frequently used gait.

2.1.1 Gait Cycle

The gait cycle is considered to be the period between two strikes of the same foot (right or left) to the ground. This cycle has two main phases: swing phase and stance phase. During the swing phase, the foot is in the air and there is no contact between the foot and the ground. In the stance phase, the foot is in contact with the ground. The time duration of the stance phase is normally 60 percent of the gait cycle's time [8].

The stance phase consists of five main gait events [9]: Heel strike, foot flat, mid stance, terminal stance and toe off. The heel strike event is described as the moment when the heel strikes the ground. After the heel strike, the rest of the foot starts to contact the ground until it finishes at the toes. In this phase, the foot is flat on the ground and the body weight shifts to the stance foot. This phase is known as the foot flat. After the weight is shifted, the body balances upon the stance foot. This phase is called the mid stance. Then the heel starts rising from the ground and the foot enters the terminal stance. This is the part of the gait cycle where the heel is in the air and the toe is still in contact with the ground. The rising of the foot continues until the toe off. The toe off is the moment that the toe rises in air which is the end of the stance phase and beginning of the swing phase.

2.1.2 Muscles and Joints in the Human Locomotion

There are three main joints and their corresponding muscles that contribute to the human locomotion, the hip, knee, and ankle. Farris et.al. [6] shows that the ankle provides about 46 percent, the knee 14 percent and the hip 40 percent of the energy required for walking. Hence, the ankle provides a great portion of the required energy for walking. For this reason, the focus of this project will be on the ankle's joint.

Ankle joint can move in dorsiflexion and plantarflexion. In plantarflexion, the foot is pushed toward the ground. Dorsiflexion pulls the foot upward lifting the toes [10]. Muscles contributing to plantarflexion and dorsiflexion are located at the calf and the shin respectively. The important muscles and tendon associated with the ankle's movement are explained below [10]:

Gastrocnemius Muscle: This superficial muscle is located at the back of the leg and runs from above (superior) the knee to the heel. The gastrocnemius starts from above the knee at the distal end of the femur and attaches to the Achilles tendon which is attached to the heel (calcaneus) bone. Contraction of this muscle will push the foot to the ground while walking and will also flex the knee.

Soleus Muscle: This powerful muscle is deeper inside the calf. It is attached to the tibia and fibula just below the knee and runs towards the heel where it is attached to the Achilles tendon. The soleus muscle provides plantarflexion movement of the foot which is used while walking and while balancing the posture. Because this muscle starts below the knee and gastrocnemius starts above the knee, when the knee is bent, soleus is more efficient than gastrocnemius in plantarflexion of the foot.

Tibialis Anterior Muscle: This muscle which is located on the outside (lateral side) of the tibia on the front side of the leg (anterior). The tibialis anterior (TA) muscle helps the foot in dorsiflexing. The TA muscle lifts the toe during the swing phase to prevent any collision with the ground and also helps to stabilize the ankle in the heel strike.

Achilles Tendon: The Achilles tendon, also known as the calcaneal tendon, is the thickest tendon in the human body and located at the back of the leg. This tendon connects soleus and gastrocnemius muscles to the heel bone.

2.2 Electrical Muscle Stimulation (EMS)

An Electrical Muscle Stimulation (EMS) unit can cause a muscle contraction by applying current to the muscles via two electrodes. There are different kinds of EMS devices available which can be distinguished by the level of muscle reaction. In a Transcutaneous Electrical Nerve Stimulation (TENS) unit, the induced current is small and only cause a light stimulation, which can be used for the muscle's pain relief [11]. Passing a current threshold will cause the muscle to contract. A Functional Electrical Stimulation device induces functional muscular contraction and therefore movement in the muscle [12].

The frequency of the pulses sent from the FES unit is tunable. The frequency range is Having low frequency pulse will contract and relax the slow twitch muscle fibers and the high frequency pulses will contract the fast twitch muscle fibers. The slow twitch muscle fibers respond to frequencies around 30 Hz and the fast twitch muscle fibers respond to 80-150 Hz frequencies [13].

2.3 Inertial Measurement Unit (IMU)

Useful feedback on human locomotion, as a dynamical system, comes from the movement characteristics and the state of the body, more specifically the positions and velocities of the lower body segments. In this project, the lower limb joints angle will be measured to determine the current configuration of the body. For this purpose, one can use an inertial measurement unit consisting of a gyroscope coupled with an accelerometer. These sensors can measure the joint angles with two different methods, which will be discussed in more detail in the following sections. The information from these two sensors can be combined to obtain a more accurate joint angles measurement.

2.3.1 Accelerometer

A tri-axial accelerometer measures the acceleration (force divided by the mass) in three directions. The acceleration of gravity is about 9.8 m/s^2 in the downward direction and can be used to calibrate the accelerometer. The tilt angles relative to the x and y axis can be measured using the following equations [14]:

$$\theta_x = \tan^{-1}\left(\frac{a_x}{\sqrt{a_x^2 + a_z^2}}\right) \tag{2-1}$$

$$\theta_y = \tan^{-1}\left(\frac{a_y}{\sqrt{a_y^2 + a_z^2}}\right) \tag{2-2}$$

In these equations, θ_x and θ_y are the tilt angles relative to the x and y axis, respectively. a_x, a_y and a_z are the accelerations in x, y and z directions.

A problem with the accelerometer is that it is susceptible to vibrations; which could occur when the heel strikes the ground while walking. The vibration occurs because the muscles have a finite stiffness and a mechanical impact such as a heel strike will result in an oscillation in the muscles and therefore in the IMUs attached to them. The purpose of using IMUs in gait analysis is not to measure the high frequency vibrations of the muscles, but to measure the lower frequency of the changes of the angles within the joints. Therefore, one can use a low pass filter, reducing the high frequency artifacts and keeping the lower frequency accelerations unchanged. The frequency of walking is around 2.5 Hz [15], therefore a low pass filter with the cut off frequency of more than 2.5 Hz (3 Hz for example) is desirable [15].

2.3.2 Gyroscope

The triaxial gyroscope measures the angular velocity in three directions. By integrating this value over time, one can calculate the angle, relative to the initial direction. In a discrete system, the integral can be approximated by,

$$\theta(t) = \int_0^t \dot{\theta}(t)dt \approx \sum_0^t \dot{\theta}(t)T_s \qquad (2\text{-}3).$$

In this equation, $\dot{\theta}$ is the measured angular velocity, θ is the estimated angle and T_s is the sampling time. The multiplication of the angular velocity by the time step estimates the change in angle in that time step. Adding up the changes in each time step from the initial time to the current time, the current angle can be obtained. The gyroscope provides an additional way to measure the angle, but one cannot rely on the gyroscope data alone, because the estimation error will grow in time and drift the measured angle from the actual angle. As a result, a high pass filter is required after the gyroscope, reducing the long-term (low frequency) measurements and not affecting the short-term (high frequency) measurements [14].

2.3.3 Complementary filter

By combining the information obtained from the accelerometer and the gyroscope, a more accurate estimation of the actual angle can be obtained. This sensor fusion algorithm is known as a complementary filter [14]:

$$\theta_{filtered} = \alpha * \theta_{gyroscope} + (1 - \alpha) * \theta_{accelerometer} \qquad (2\text{-}4).$$

In this equation, $\theta_{gyroscope}$ is the angle measured by the gyroscope, $\theta_{accelerometer}$ is the angle measured by the accelerometer, $\theta_{filtered}$ is the filtered angle and α is a tuning parameter between 0 and 1, showing the contribution of each sensor measurement to the final estimation. The literature states that the tuning parameter α is close to the value of 0.9999 [15].

2.4 Force Sensitive Resistors (FSRs)

Force Sensitive Resistors (FSRs) can be used to measure small forces acting on them. This information can be used to find the gait events such as heel strike and toe off. The sensor's resistance changes linearly with the force applied to the sensor. Therefore, one can use the resistance's change to measure the applied force on the sensor. To measure the resistor value, one can use a voltage divider by adding a known resistor value in series to the FSR and measure the voltage drop using an Analog to Digital (A/D) converter [16].

2.5 Camera Motion Capture System

Camera motion capture system is a method of measurement for gait analysis. In this method, fluorescent markers are attached to the body and cameras track the markers location. This is possible because the cameras are equipped with infrared LEDs and cause a reflection on the markers. The cameras pick up the markers reflection and determine the markers location in 3D space utilizing triangulation [17].

The data is collected by using several high-speed cameras. The reason for having multiple cameras is that a marker may be obscured from one camera's view during movement. Having multiple cameras can reduce the possible marker loss due to obstruction. Also, more cameras result in a more accurate determination of location.

The camera motion captures are connected to a desktop computer through cables. Connecting the markers position from several cameras and combining the position data with a model of a human body, one can estimate the human motion data such as the joints' angles. The markers will be used to produce a 3D skeletal structure. An advantage of this method is that the markers are light and have the minimum effect on the gait.

The Health and Human Science Building of Western Carolina University owns a Qualisys Miqus M3 (Qualisys Americas, Chicago, IL, USA) [18], ten camera motion capture system. The Miqus M3 cameras (Figure 2.1) are equipped with invisible infrared light and have 2 Mega Pixels (1854*1088) resolution and a 340 fps frame rate [17].

Figure 2.1 Miqus M3 camera

CHAPTER3: DEVICE DEVELOPMENT

3.1 Preliminary Device

The purpose of the first revision of the neural prosthesis device is to stimulate the gastrocnemius (GN) muscle using a manual switch and collect the ankle angle and the foot pressure data while walking. The manual switch was used as a first step toward designing an automatic switch for stimulating the GN muscle. The data were postprocessed to observe the effect of the GN stimulation on the gait.

The contraction of the GN muscle will be through an EMS unit and will be triggered via a relay switch. The ankle angle will be estimated by placing one IMU on the shin and one on the foot. The horizontal angle of the IMU on the foot will be subtracted from the horizontal angle of the shin to obtain the ankle angle. Force Sensitive Resistors (FSRs) will be utilized to detect the contact between the foot and the ground. A microcontroller will be used to collect data from the IMUs and the foot switches. Overall, the preliminary device contained two IMUs, four FSRs, an Arduino microcontroller and an EMS unit. The design of each major component is described in the following sections.

3.1.1 Foot Switches

The first step in estimating location within the gait cycle is to detect the initiation of the gait cycle. The beginning of the gait cycle (0%) is defined by the heel strike. This event can be determined using a sensor placed on the ground (force plates) [8] or on the foot (FSR) [16]. Because the neural prosthesis must be self-contained, the foot sensor is the only option. The force sensitive resistor (FSR) 402 model (Interlink Electronics, Camarillo, California, USA) [19] was chosen to detect the contact between the foot and the ground because it is cost efficient, small enough to detect the contact locally, and easy to use. The cost of a single FSR was $7. The specifications of FSR-402 are summarized in Table 3.1:

Table 3.1 Specifications of FSR-402 [19]

Specification	**Value**
Force Range	0 to 20 lb.
Resistance Change	Infinity (no pressure) to 200 Ω (maximum pressure)

In addition to the sensor detecting the heel strike, another FSR was added to configure the occurrence of the toe-off and the swing phase. Also, two more FSRs were placed on the first and the fifth metatarsal to detect weight shifts in the forefoot. In order to make the experiment convenient for the participant, all FSRs were attached below a shoe insole as demonstrated in the picture below (Figure 3.1):

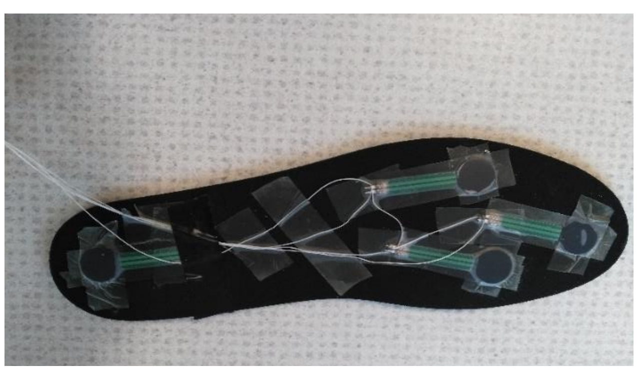

Figure 3.1 She insole equipped with four FSRs.

The FSR sensors were connected to the microcontroller as shown in Figure 3.2. The sensor value was collected by using the analog to digital converter on the Arduino Uno. The FSRs were connected to pull-down resistors in order to shift the output value to zero when there was no load on the sensor.

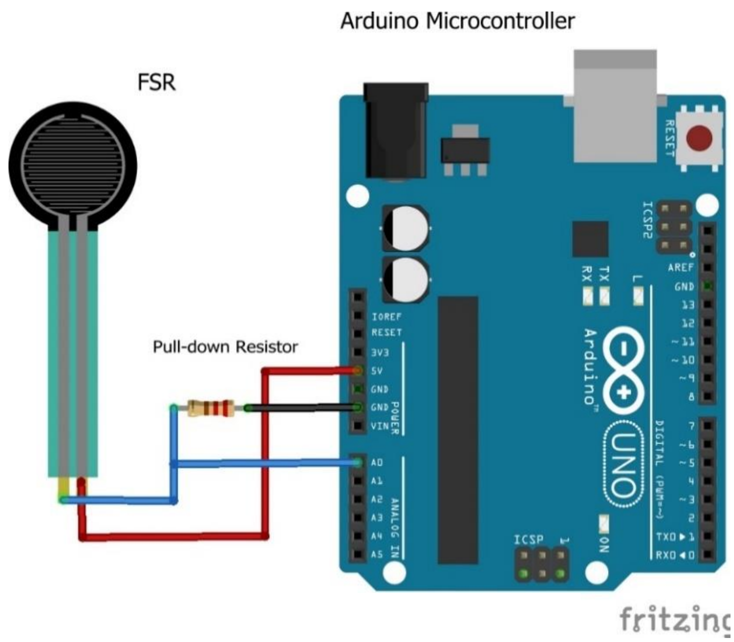

Figure 3.2 Voltage divider for measuring the FSR's output [20]

3.1.2 Inertia Measurement Units

In this project, it was assumed that the patient is walking in a straight line; therefore, it was decided not to use an IMU equipped with a magnetometer. Because the magnetometer is used to measure the absolute angle in the transverse plane which does not vary much when the participant is walking in a straight line. Following a thorough review of available IMUs, it was determined that MPU-6050 (InvenSense, San Jose, California, USA) [21] was the best fit due to its low cost and suitable performance characteristics. The cost of each MPU-6050 was about $3. The specifications of the MPU-6050 are summarized in Table 3.2:

Table 3.2 Specifications of MPU-6050 [21]

Parameter	Accelerometer	Gyroscope
Full-Scale Range	±2 g, ±4 g, ±8 g, ±16 g	±250˚/s, 500 ˚/s, ±1000 ˚/s, ±2000 ˚/s
Sensitivity Scale Factor	16384 LSB/g, 8192 LSB/g, 4096 LSB/g, 2048 LSB/g	131 LSB/(˚/s), 65.5 LSB/(˚/s), 32.8 LSB/(˚/s), 16.4 LSB/(˚/s)
Zero offset	X and Y: ±50 mg, Z: ±80 mg	±20 ˚/s

Two IMUs were connected to the microcontroller via I^2C protocol (Figure 3.3). This protocol is capable of transmitting data in series using only two wires by calling an address associated with the sensor [22]. The Arduino code for reading the sensor's output via this protocol is included in Appendix A [23]. Each sensor's output is a two-bytes (16-bit) signed integer. For having symmetric data around zero, the most significant bit was used for determining the sign of the output, thus the range of the received integer was -2^{15} to 2^{15}-1 instead of 0 to 2^{16}-1.

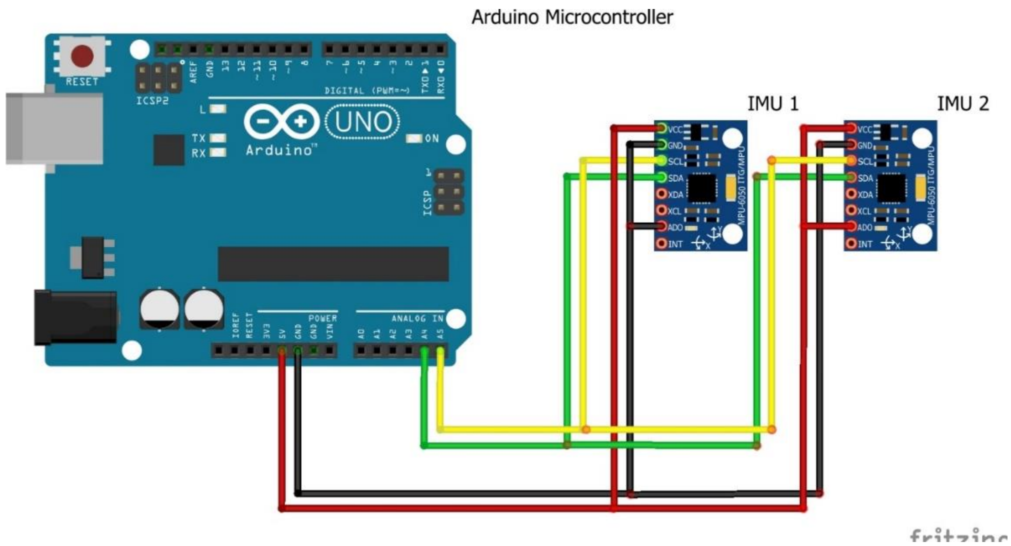

Figure 3.3 Arduino and two MPU6050 circuit

The sensors' outputs do not represent the accelerations and angular velocities in SI units. The outputs are an affine function (linear function plus a constant number, e.g. y = mx + b) of the actual values. As a result, constants in the affine function need to be determined in order to approximate the acceleration and angular velocity. The sensor's datasheet provided an approximation of these constants, but these parameters were also derived in order to generate a more accurate approximation. The methods utilized are described below.

For calibrating, IMUs were placed on a smooth leveled surface for a few seconds while measurements were recorded (Figure 3.4). By averaging the measured output, one can estimate the mean value. The mean of the acceleration in the downward position when the IMU is on a smooth surface is equal to 9.8 $\frac{m}{s^2}$ (Figure 3.5). When the sensor is in the upward position, this value is $-9.8\ \frac{m}{s^2}$ (Figure 3.4). Using the acceleration in these two positions and assuming the sensor is linear, one can estimate the sensitivity (c) and the offset (d) of the accelerometer:

$$c = 2 * g/(a_{downward} + a_{upward}) \tag{3-1}$$

$$d = g * (a_{upward} - a_{downward})/(a_{downward} + a_{upward}) \tag{3-2}$$

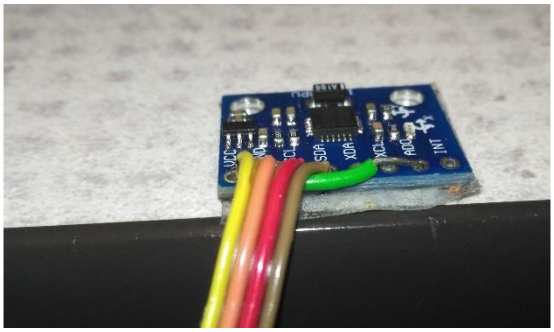

Figure 3.4 IMU in the downward orientation

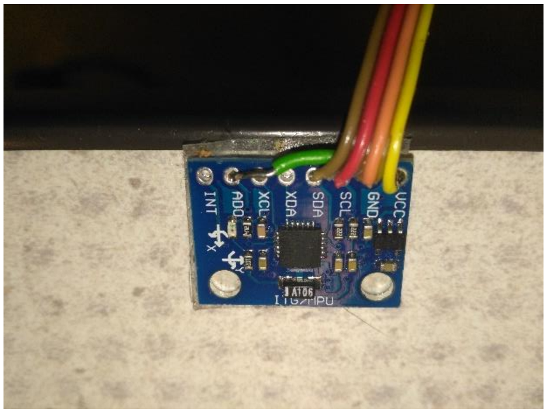

Figure 3.5 IMU in the upward orientation

Knowing c and d, one can calculate the acceleration:

$$Accelration = c * accelerometer\ output + d \tag{3-3}$$

According to the IMU's data sheet, the sensitivity of the gyroscope is 131 in all three directions [19]. As a result:

$$Angular\ velocity = gyroscope\ output/131 \tag{3-4}$$

3.1.3 Electrical Muscle Stimulation (EMS) unit

The EMS 5000 (EMSI, Tampa, Florida, USA) [24] was chosen because it is affordable, it is approved by the FDA, and the researchers had prior experience using this device [12]. The features of the EMS 5000 are mentioned in Table 3.3:

Table 3.3 Specifications of EMS 5000 [24].

Specification	Value
Pulse Amplitude	0-80 mA
Pulse Frequency	5, 30, 100 Hz
Contraction Time	1-30 Seconds
Relaxation Time	1-45 Seconds
Power Source	9-Volt Battery

A relay module was utilized to switch the EMS on and off. The relay module was powered up through the microcontroller. A voltage divider was used to record the state of the switch (Figure 3.6).

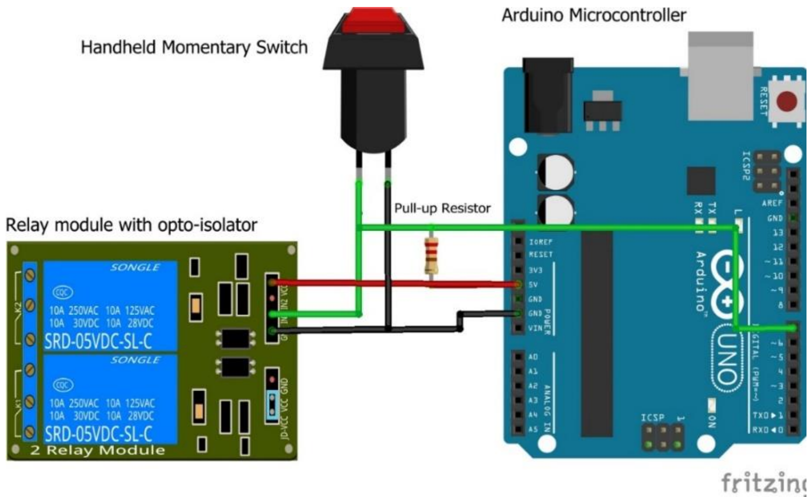

Figure 3.6 Relay and handheld switch diagram

The EMS were connected to the muscle via two electrode pads. The electrode pads were available in different sizes. The larger the size of the electrode pads, the more contraction we will have on the muscle and the less accuracy we will have on contracting the target muscle. The 1.75 inch x 3.75 inch electrode pads were utilized, because the size was large enough to have a sufficient contraction on the GN muscle and compact enough to not contract the other muscles nearby. The placement of the electro pads on the GN muscle were represented in Figure 3.7:

Figure 3.7 Electrode pads placement on the GN muscle

3.1.4 Microcontroller

An Arduino Uno (Arduino, Italy) [25] microcontroller was utilized due to its popularity as a platform for prototyping of the embedded systems. The Arduino Uno has enough input pins to read 9 sensors and sufficient sampling speed in the I^2C protocol to capture IMUs' data [14].

As described in the Arduino code in Appendix A, to read the data from the IMUs, the frequency of the I^2C protocol was set up utilizing the "Wire.setClock" function [20]. Then, the sensor's reading was initialized by calling the MPU6050 I^2C address using the "Wire.beginTransmission" function. After that, byte by byte of the sensor's output was obtained by applying the "Wire.write" function. To read the FSR's analog values from the Arduino Uno, the "analogRead" function was used, which output a value proportional to the force acting on the sensor.

Parameters including time in milliseconds, state of the switch, FSRs data, three-dimensional acceleration and angular velocity were recorded during the experiments. A Bluetooth module was connected to the serial port of the microcontroller. The Bluetooth module was paired with the researcher's laptop. The recorded data was sent via the Bluetooth module to the researcher's laptop using the "Serial.print" function. The HC-05 Bluetooth module [14] was chosen over the other wireless modules, due to its cost efficiency and its sufficient capability for the one to one communication.

3.1.5. System Design

For this iteration of the design, IMUs were attached to the right shin and the right foot as shown in Figure 3.8. The IMUs were calibrated prior to the experiment. The shin's IMU was placed on the skin close to the tibia bone and was tightened by wrapping physical therapy tape around the shin (Figure 3.8). The foot's IMU was fixed on the top of the participant's shoe. The participant also wore a shoe insole containing four FSRs on the heel, toe, as well as the first and fifth metatarsal. The IMUs and the FSRs were connected to the Arduino microcontroller via jumper wires. The microcontroller was placed on the top of the foot. The frequency of the data collection was set to 50 Hz. The microcontroller was powered by a 9V battery. The battery was placed on the top of the microcontroller.

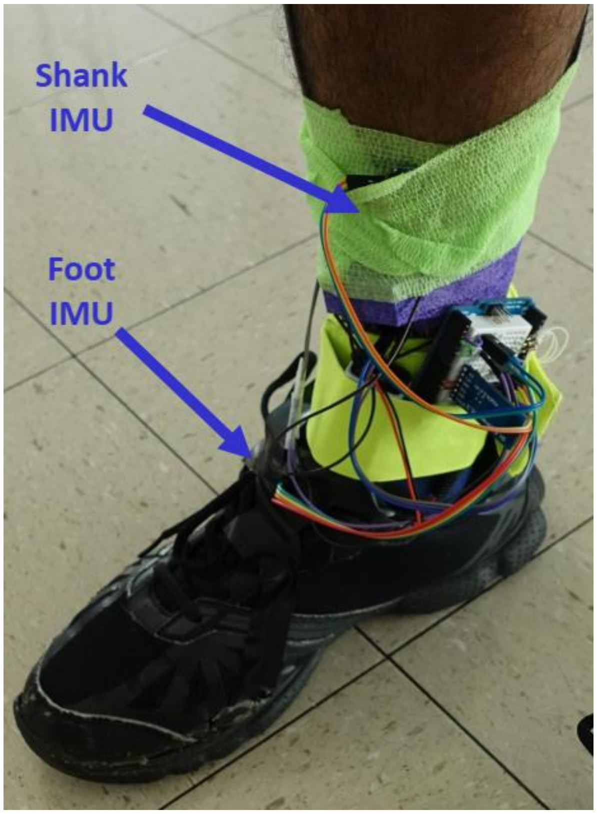

Figure 3.8 IMUs positions

The FES unit was placed at the waist. One electrode was connected directly to a pad attached on top of the gastrocnemius muscle. Another electrode went through a relay module and then to a pad attached below the gastrocnemius muscle and at the top of the Achilles tendon as shown in Fig 3.7. The relay module could be switched on and off by pushing and releasing a handheld switch, respectively. The state of the switch was recorded by the microcontroller. The schematic of the first revision of the device is shown below in Fig 3.9.

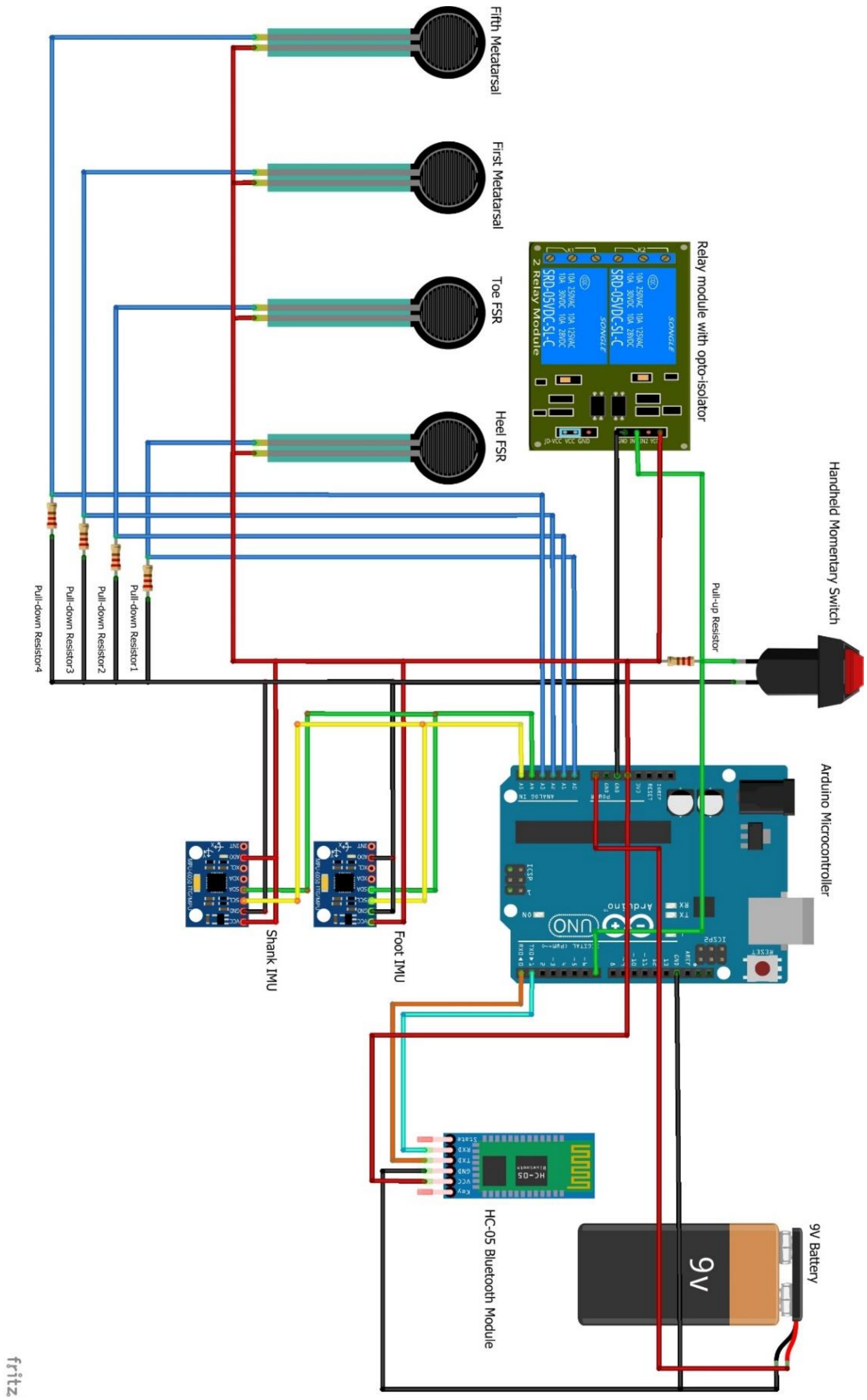

Figure 3.9 The preliminary device's schematic diagram

3.2 The Testing of the Preliminary Device

In this section, the process of testing the neural prosthesis device will be discussed. Two different sets of tests were performed on a participant:

1) The participant walked normally while the sensors recorded the motion data.

2) The participant walked normally while he switched the EMS module to contract the GN muscle in the push off phase.

The first test was designed to observe the performance of the device in recording the gait data and the second test was designed to observe the effect of the GN muscle stimulation on the gait, by comparing the gait data with and without the muscle stimulation.

The test preparation was started by explaining the purpose of the project and the process of the experiment to the participant. Prior to beginning the study, each participant read and signed the informed consent form approved by the IRB at Western Carolina University. The shoe insole equipped with the FSR sensors was placed in the right shoe of the participant. As shown in Figure 3.8, one IMU was attached to the shin and another one on the top of the shoe using physical therapy tape. The microcontroller and the battery were attached to the participant's shoe utilizing physical therapy tape. The electro pads were placed on the gastrocnemius muscle and were connected to the EMS module. As shown in Figure 3.10, the EMS unit was placed on the waist of the participant using a clip.

Figure 3.10 EMS placement on the waist

In the next stage of the setup, the microcontroller was turned on, and the Bluetooth module was paired with the researcher's laptop. Then, the researcher tuned the EMS unit to have enough amount of contraction to be able to be captured by a camera and also affect the gait. The tuning was performed by turning the knob on the EMS while observing the GN muscle contraction on the participant's muscle. In order to test the device, some sample data was collected from the IMUs and the FSRs prior to the experiment. After preparing the equipment, the research team started the two sets of trials. Each set was repeated 10 times. In each trial, the participant walked 20 meters in a straight line while an observer was walking beside him. One researcher collected the data and another one recorded video from the experiment with a camera.

3.2.1 Test results

The collected data from the sensors was postprocessed utilizing MATLAB [26]. The foot, shank and ankle angle were calculated by implementing a complementary filter on the IMUs' data as explained in Section 2.3. The foot, shank, and ankle angle are shown in Fig 3.11-13. The blue solid lines show the data with the FES. The green solid lines show the data without the FES. The red square wave represents when the FES was switched on or off. The dotted lines are located two standard deviations from the mean value. The deviation from the mean value can be in the positive and negative direction. Hence, there are lines above and below each mean value curve.

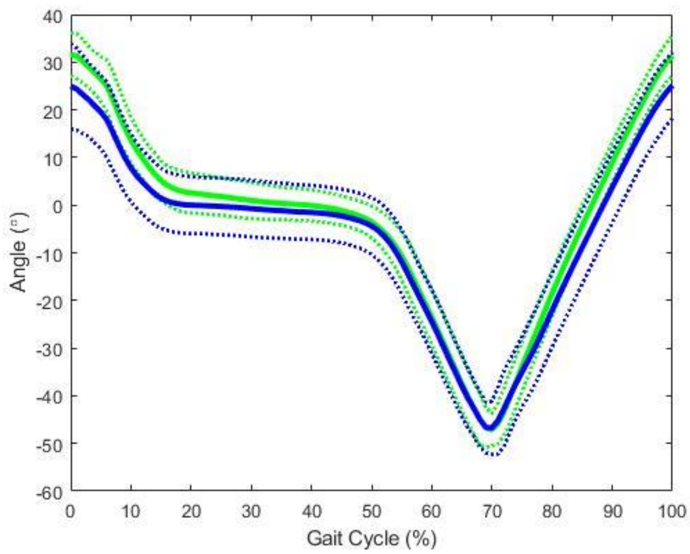

Figure 3.11 Foot angle vs gait cycle

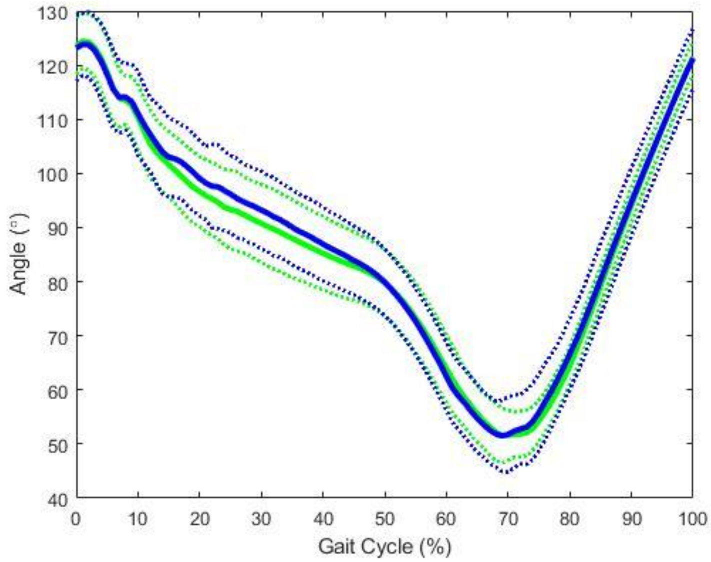

Figure 3.12 Shank angle vs gait cycle

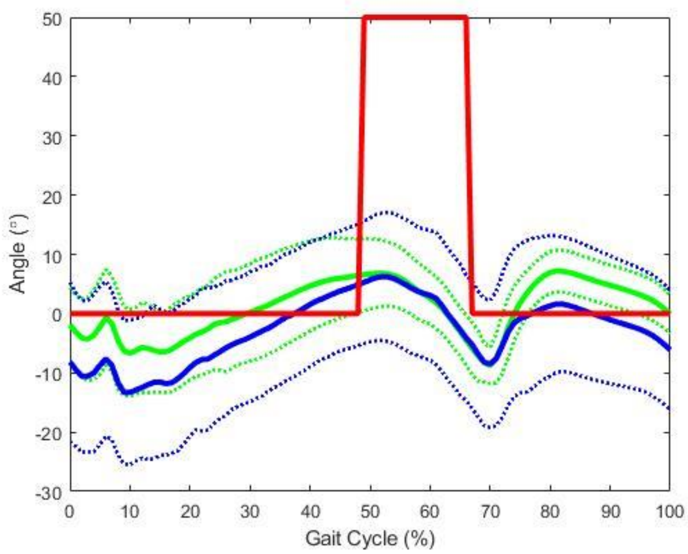

Figure 3.13 Ankle angle vs gait cycle

Figure 3.11 shows the deviation of the blue and green curves from about 0 to 50% of the gait cycle and from 70% to 100% of the gait cycle. This represents that during this period, the foot is more plantarflexion when using the FES. Figure 3.12 shows that the blue and green curves are close to each other, stating that the GN stimulation did not change the shank angle.

Figure 3.12 is a summation of the curves in the Figure 3.11 and 3.12. As mentioned, shank angle remains unchanged with and without the FES but the foot angle changes with and without the FES. Therefore, the difference in the blue and the green curve in the Figure 3.13 is a result of change in the foot angle. Comparing the data with and without the FES, one can conclude that when the GN muscle stimulates in the push off phase, the foot will remain plantarflexion during the swing phase. The foot plantarflexion did not cause the foot to hit the ground during the swing phase.

Furthermore, stimulating the calf muscle did not make the participant’s gait unstable or cause any visual tendency to fall. Therefore, there is no need to have a researcher walking beside the participant to protect him from falling.

3.3 The Design of the Final Device

There were several problems with the initial design, which were resolved in the second revision of the design. The first problem was that the connections between the IMUs, foot pressures, and the microcontroller were via jumper wires inserted into a breadboard. During the experiment, the connections came loose several times and the trial had to be repeated. This problem was addresses by soldering the connections and replacing the breadboard with a routed proto-board.

The second problem was that the microcontroller and the battery were packaged together. The cables were short, and the package need to be placed on the ankle. Placing the package on the ankle can change the gait due to the increased mass added to the foot. This problem was addresses by using longer wires and placing the microcontroller on the waist.

The third problem was that the first revision of the device had only two IMUs which is not enough for measuring the knee and hip angle. This problem was addressed by replacing the Arduino by a Teensy 3.6 microcontroller (PJRC, Sherwood, Oregon, USA) [27], having more I^2C protocol sets of pins, capable of collecting data from more IMUs (Figure 3.14). Two IMUs were added to record the thigh and torso angle. As a consequence of these changes, 12 more variables (acceleration and angular velocity in 3D for two sensors) were added to the collected data. Unfortunately, the Bluetooth connection was not fast enough to transmit this much sensor data. This problem was resolved by replacing the wireless connection with a wired USB.

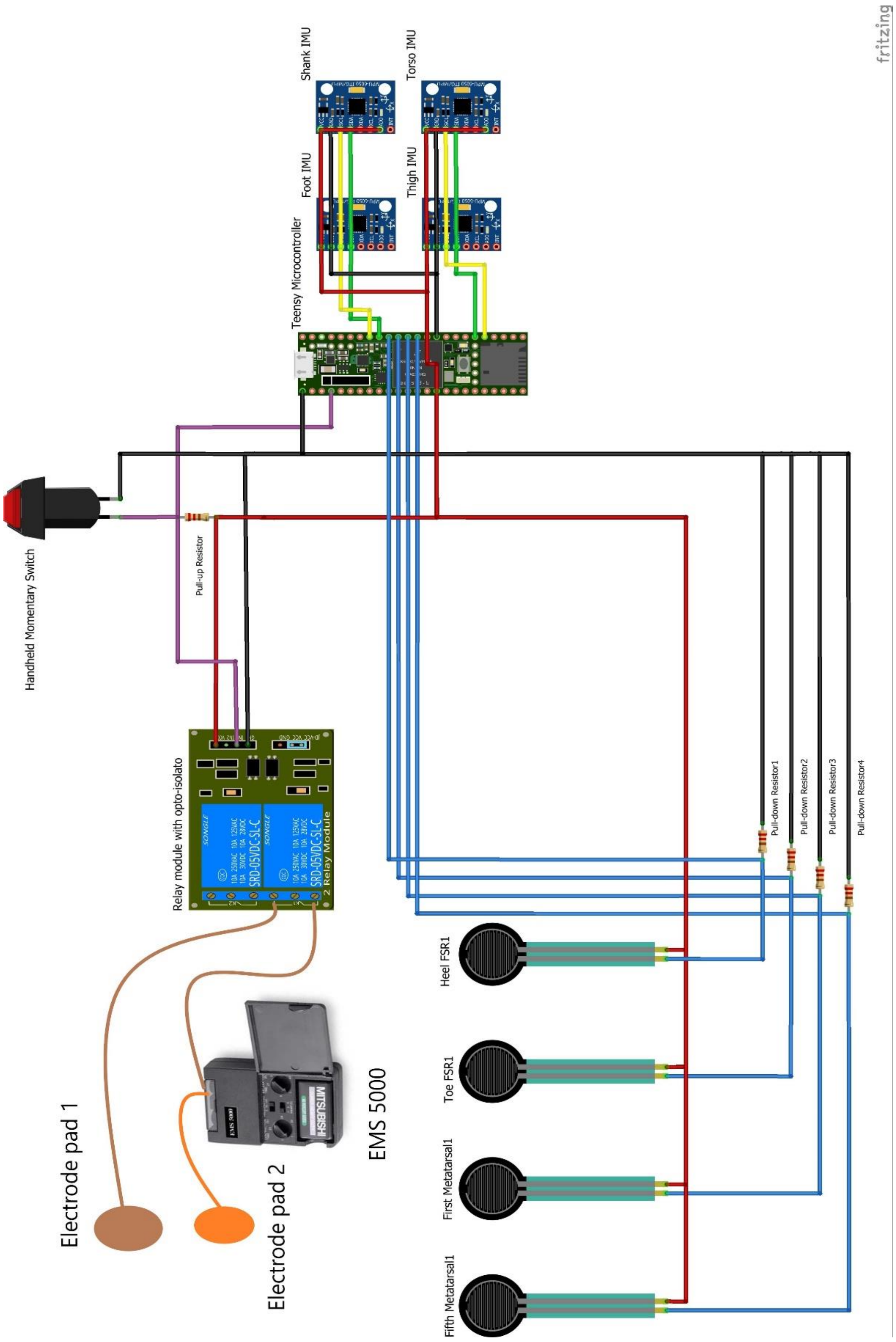

Figure 3.14 The final device's schematic diagram

3.4 Data Analysis Methods

The camera motion capture system's data was filtered with a Butterworth filter using a cutoff frequency of 6 Hz. The data from the accelerometers and the gyroscopes were fused using a complementary filter.

After post-processing the data, the ankle angle was calculated by subtracting the foot angle from the shank angle. Knee angle was calculated by subtracting the shank angle from the thigh angle. Hip angle was calculated by subtracting the thigh angle from the torso angle.

In this project, different computational methods were used to analyze the collected data. These methods are described in following:

Forward difference: Forward difference method was used to estimate the derivative at one point when analyzing discrete data. In this project, the joint angles were obtained at short time intervals (2 milliseconds) utilizing the IMUs which produce discrete data. To calculate the angular velocity from the IMU data, the forward difference method was used [28]:

$$Angular\ velocity\ (t_1) \cong \frac{Angle\ (t_2) - Angle(t_1)}{(t_2 - t_1)} \qquad (3\text{-}5).$$

In equation (3-5), $Angle\ (t_1)$ and $Angle\ (t_2)$ were the angles at time t_1 and t_2. $Angular\ velocity\ (t_1)$ was the estimated angular velocity at time t_1

Correlation Coefficient: This coefficient can be used to measure the correlation between two series of data using a single number. The correlation coefficient between two series of data (X and Y) was defined as [29]:

$$r = \frac{cov(X,Y)}{\sigma_X \sigma_Y} \quad (3\text{-}6).$$

In this equation, $cov(X,Y)$ was the covariance of X and Y. σ_X was the standard deviation of X and σ_Y was the standard deviation of Y. r was the correlation coefficient which was a value between (-1) and 1. Positive correlation coefficient means that as X increases, Y increases. Negative correlation coefficient means that as X increases, Y decreases. The closer r to 1 and -1, the more correlated X is to Y. Zero correlation coefficient means that there was no correlation between X and Y. One can use the "corrcoef" function in MATLAB to find the correlation coefficient.

Root Mean Square Error (RMSE): This value is a measure of accuracy between two series of data (X and Y). The RMSE was defined as [28]:

$$\text{RMSE} = \sqrt{\frac{\sum_1^N (X_i - Y_i)^2}{N}} \quad (3\text{-}7).$$

In this equation, $\sum_1^N (X_i - Y_i)^2$ was the summation of squared error between the components of X and Y. N was the number of data points in X and Y. The RMSE had the same dimension as the data points in X and Y.

CHAPTER 4: TESTING

The purpose of the device testing was to examine the neural prosthesis device ability to record gait data and to be able to observe the effect of the GN muscle contraction on the human gait. Over ground tests on a healthy individual were performed in the Health and Human Science building at Western Carolina University. To perform this test, the second revision of the neural prosthesis device was used. The collected data from the IMUs was compared to the data collected from the camera motion capture system to determine the accuracy of the IMUs. The camera motion capture system is considered the “Gold Standard” for this research.

4.1 Test preparation

4.1.1 Participant’s preparation

The test protocols were approved by the Institutional Review Board (IRB) at Western Carolina University and the participant signed a consent form prior to participating in the experiment. The participant was asked to wear shorts to allow the researcher better access to the GN muscles. The attire allowed the visibility of the muscle contractions for video data collecting. The purpose of the test and the project were explained to the participant before commencing the data collection. In one of the tests, the participant was instructed to trigger the FES unit by himself. The participant was given an explanation and demonstration of when the trigger should be pressed during the gait phase. The participant was allowed several dry runs before the data was collected.

4.1.2 Equipment preparation

The neural prosthesis device was tested prior to the experiment to prevent any delay during the experiment due to technical issues. This test was performed by instructing the participant to walk across a flat level surface while the research team observed the collected data. The FES unit was tuned to contract the GN muscle as explained in 3.2. The FES unit was set to have the maximum time of contraction (30 second) and minimum relaxation time (1 second) as shown in Figure 4.1. This setup was chosen in order to have the FES device “on” for the maximum amount of time so that the manual switch could be used to trigger the GN muscle contraction.

Figure 4.1 FES on the participant

4.1.3 Coordination for Data Collection

The School of Engineering and Technology of Western Carolina University does not currently own a camera motion capture system; therefore, an appointment was scheduled with the Director of the Human Movement Science Lab to use their camera motion capture system available in the Health and Human Science Building.

4.1.4 Marker placement

A complete set of markers was attached on the participant’s body as shown in Figure 4.2 and clothes using double sided tape.

Figure 4.2 Participant with a set of markers and IMU attached

4.1.5 Foot sensor placement

The shoe insole had four FSRs placed in the participant's right shoe. The participants were asked for their shoe size in order to purchase insoles for their shoes. The insoles were modified by placing four sensors in the bottom of the insole. All four sensors were tested by asking the participant to place pressure on the sensors separately and observe the measured values.

4.1.6 IMU placement

The IMUs were attached to the person's shoe, shank, thigh and torso, utilizing double sided tape. The foot IMU was placed on the top of the shoe and secured by the shoelaces (Figure 4.3). The shank IMU was attached to the skin on the top of tibia (Figure 4.4). As explained in 2.3.1, the muscle oscillates more than the bones during an impact. Therefore, the sensor was not attached to the muscle to diminish the effect of sensor noise resulting from the muscle's vibration. Because of the muscles surrounding the bone in thigh, the IMU could not be attach to the femur, therefore an IMU attached to the skin on the thigh (Figure 4.5). Another IMU was attached to the t-shirt on the top of the hip bone (Figure 4.6).

Figure 4.3 Foot IMU

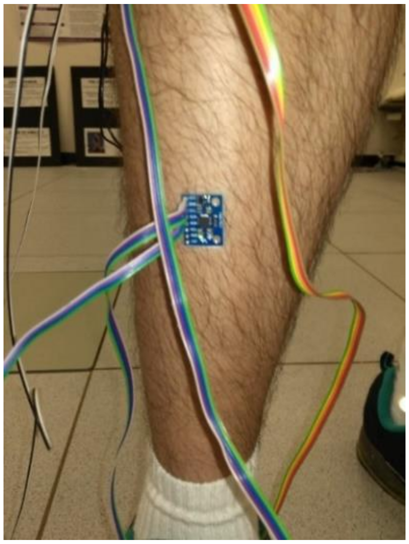

Figure 4.4 Shank IMU

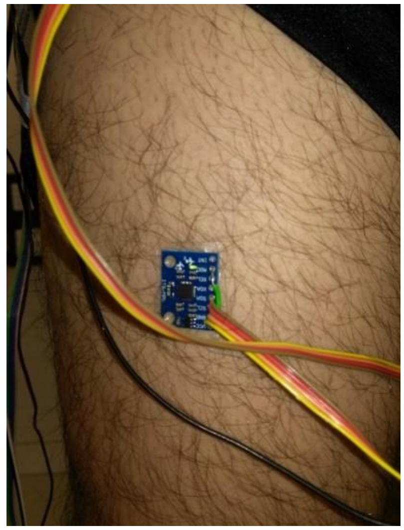

Figure 4.5 Thigh IMU

Figure 4.6 Torso IMU

4.2 Testing Procedure

Each trial was initiated by one researcher calling the number of the trial and the type of test that was going to be performed. There were three different testing configurations: normal walking, walking while the participant is triggering the FES unit and walking while a researcher is stimulating the calf muscle. The same researcher calling the number of the trial and the type of trial, made sure the researcher collecting the IMU data was ready to record the data; then made sure the researcher switching the FES was ready to commence. After this stage, the participant was asked to start walking in a straight line

4.2.1 Over ground walking

For the over ground test, the research team asked the participant to walk about 10 meters in a straight line. Two pieces of tape were placed on the floor as a target to help the participant walk in a straight line. The tape also served as a start and stop indicator. Three different conditions of the test were recorded:

1) The participant walked normally while the sensors recorded the motion data.

2) The participant walked normally and was asked to switch the FES module to contract the GN muscle in the push off phase.

3) The participant walked normally while a researcher switched the FES module to contract the GN muscle in the push off phase.

The first condition was needed to examine the accuracy of the IMUs when capturing the body movement. For comparison, the body motion was recorded using a standard camera motion capture system this allowed verification of the joint angles' data collected from the IMUs.

The second and third conditions were performed to observe the effect of the GN stimulation on the gait. To observe the differences between the switching time, first the switch was triggered by the participant and then by the researcher. The switch state was also recorded to observe when it was pressed during the gait cycle. The difference between the researcher and the participant can be due to difference in the feedback from the gait. The examples of the feedback are visual (which is different between the researcher and the participant) and feeling the contact between the foot and the ground (which is not available for the researcher). Also, the researcher was more experienced in this field than the participant.

4.2.2 Data collection

The data from the IMUs, FSRs and the FES switch was recorded using the Teensy microcontroller. The data was transmitted to a laptop via a USB cable and later stored in a text file. Acceleration and angular velocities in three dimensions were captured from four IMUs. The state of the FES switch, FSR data and time were also collected. Also, the camera motion capture system's data was captured using Qualisys Miqus M3 (Qualisys Americas, Chicago, IL, USA) [18]. The data was obtained and post processed in the Qualisys Track Manager software (Qualisys Americas, Chicago, IL, USA) [31] provided by the camera motion capture system's manufacturing company (Figure 4.7).

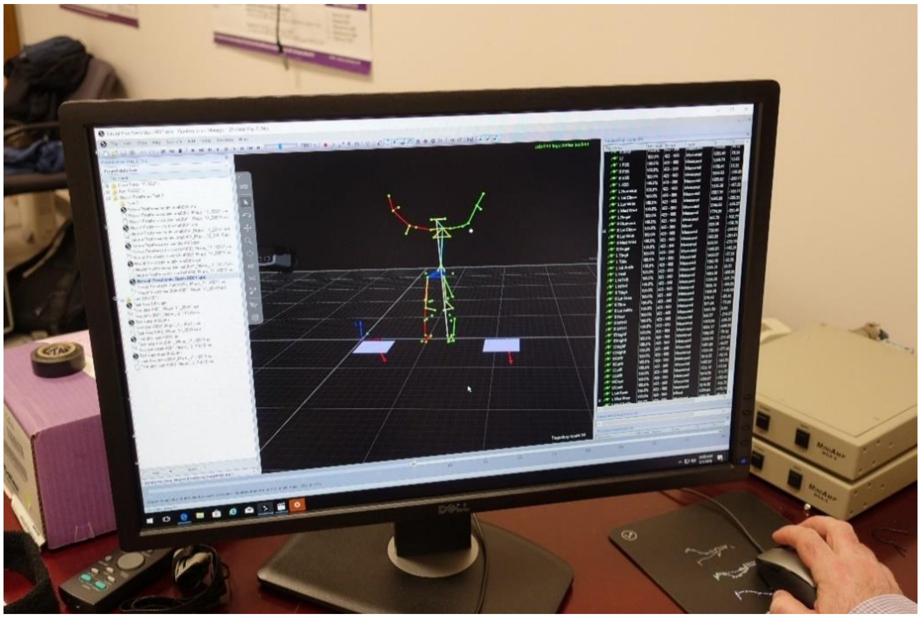

Figure 4.7 Qualisys Track Manager software

CHAPTER 5: RESULTS

In this chapter, the test results from the second revision of the device will be reviewed. The ankle, knee and hip angle were measured using the IMUs and the camera motion capture system. The chapter has three sections: movement data collected with the camera motion capture system, movement data collected with the IMUs in the neural prosthesis, and comparison of the two measurement systems to determine the accuracy of the IMU measurement system.

5.1 Camera motion capture system's data

The data of the angle of the joints collected using the camera motion capture system is shown in Figures 5.1, 5.2, and 5.3. Figure 5.1 shows the foot, knee and hip angle for the three different conditions of the test in the gait cycle. Each curve consists of 101 data points representing the joint angle in degree verses gait cycle percentage. The curves represent the average of three to four trials of the experiment. The heel strike was extracted from a video recorded of the experiment to find the 0% gait cycle.

In Figure 5.1, the ankle angle without stimulation and with the researcher switching the FES are so close to each other that the green line covers the black. This much similarity in the result indicates that at least one of these series of data is incorrect. The ankle angle with stimulation starts to deviate from the no stimulation condition around 70% gait cycle (around toe off). Another observation is that during the swing phase the knee and hip angles with the FES deviate from normal walking.

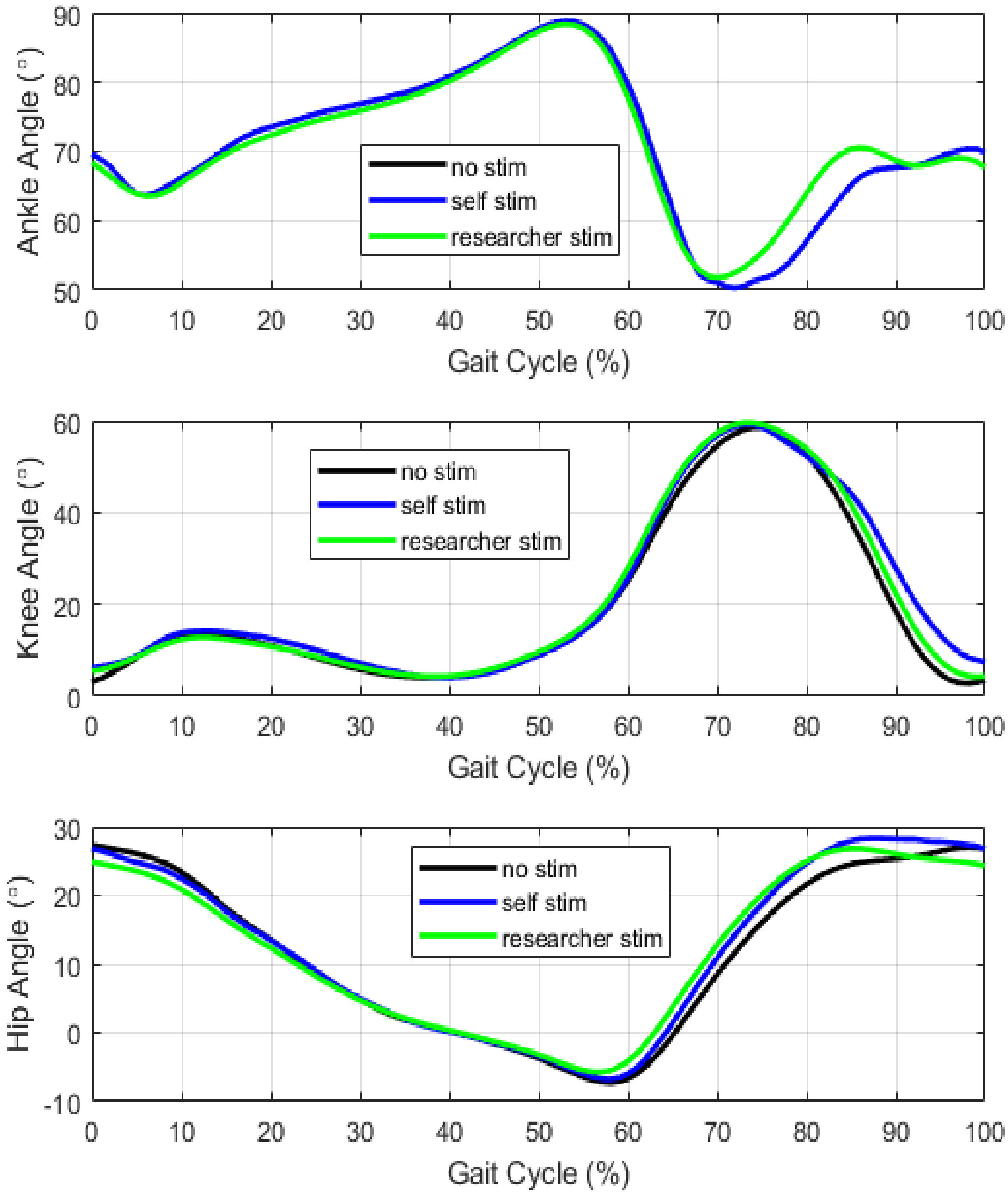

Figure 5.1 Joints angle measured by the motion capture camera system

Figure 5.2 shows the foot, knee and hip angular velocities verses gait cycle percentage. The velocities were calculated by a forward difference method (Equation 3.5). The length of the velocity data set is less than the angle data set by one point. As a result, the final velocity point is not included in the plot.

The deviation of the angular velocities follows the same pattern as the angles. Meaning that the differences between the tests with and without the FES are more in the swing phase than the stance phase. In the trials without the FES and the trials which the researcher was switching the FES, the lines are smoother and do not fluctuate as much as the trials which the participant was switching the FES.

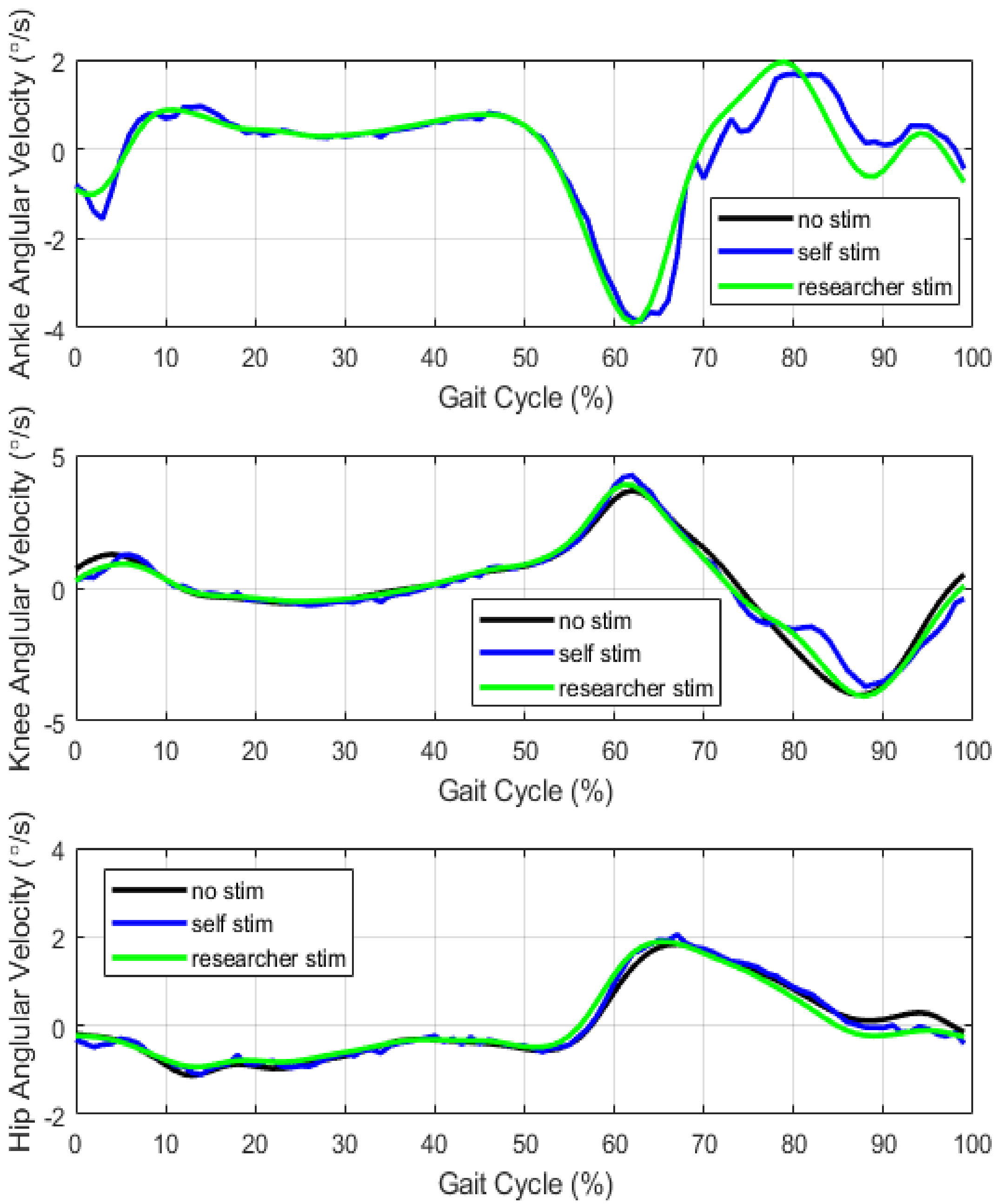

Figure 5.2 Joints velocity measured by the motion capture camera system

Figure 5.3 shows the data points for three different joint angles: the foot, the knee and the hip. All three joint angles are combined in one 3D plot. The hip angle is represented by the z axis, the foot angle is represented by the x axis and the knee angle is represented by the y axis. It can be observed that the right side of the plot shows the stance phase. This can be determined because data points are closer together which means the angular velocity is slow. In the swing phase the angular velocity points are further apart due to the faster velocity. This can be seen on the left side of the plot. In the stance phase it is noticed that all three dotted lines are close together, this is an indication that the stimulation was not affecting the gait. The opposite is true for the swing phase.

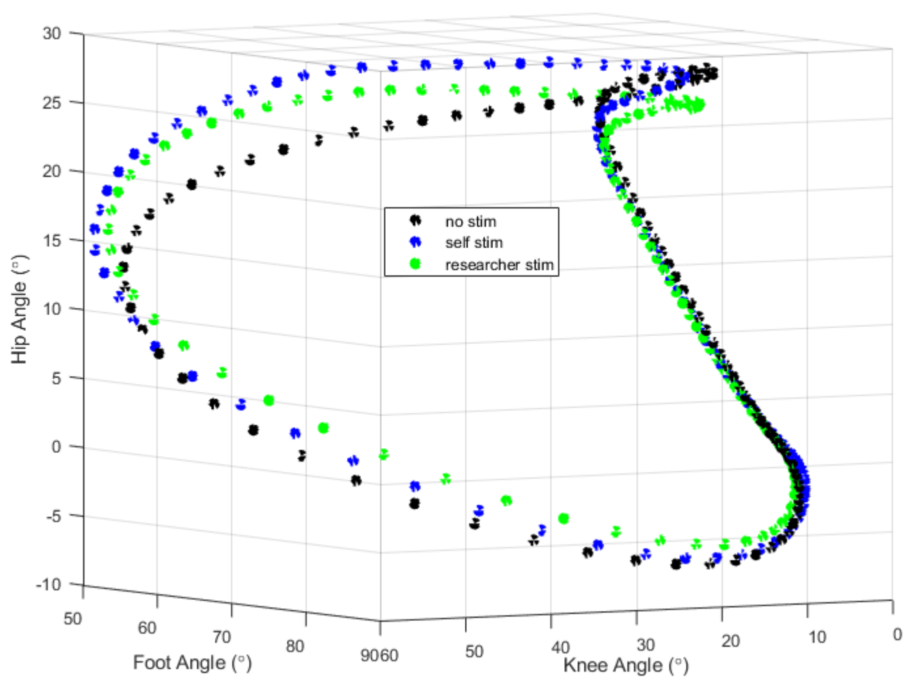

Figure 5.3 Foot, knee and hip angle

Tables 5.1 and 5.2 show the correlation coefficient (Equation 3.6) of the joint's angle and angular velocities. The correlation is between the non-stimulated walking data and the self-stimulation as well as the researcher stimulation. The correlation coefficient is between -1 and 1. As a rule of thumb, the correlation coefficients from 70% to 90% are considered as high correlation and above 90% are considered as very high correlation [29]. All correlation coefficients in Table 5.1 and 5.2 are above 90%; therefore, the joints angle and angular velocity, with and without the FES were highly correlated. The closer the correlation coefficient is to 1, the closer the correlation is between the two series of data. The less correlation between two series of data in these tables, the more affected that the experiment is by the FES. The ankle angle and angular velocity data while the researcher was switching the FES has the correlation coefficient of 1 and this set of data seems to be incorrect. Tables 5.1 and 5.2 show that the FES affected the ankle more than the knee and hip angle because the ankle has less correlation coefficient than the knee and the hip.

Table 5.1 Correlation Coefficient of the joints angle

Correlation Coefficient	Ankle	Knee	Hip
Self-stim	0.9781	0.9884	0.9939
Researcher stim	1	0.9976	0.9836

Table 5.2 Correlation Coefficient of the joints angular velocity

Correlation Coefficient	Ankle	Knee	Hip
Self-stim	0.9524	0.9652	0.9874
Researcher stim	1	0.9903	0.9749

Tables 5.3 and 5.4 show the Root Mean Square Error (RMSE), calculated by Equation 3.7, of the joints angle and angular velocities. The errors were calculated using the non-stimulation condition's data. The greater the RMSE, the more effect the FES had on the experiment. A value of zero means there is no difference which is unlikely to happen in a real-life experiment.

Table 5.3 RMSE of the joints angle

RMSE	Ankle	Knee	Hip
Self-stim	2.3604	3.4365	1.5959
Researcher stim	0	1.9443	2.2611

Table 5.4 RMSE of the joints angular velocity

RMSE	Ankle	Knee	Hip
Self-stim	0.3896	0.4647	0.1389
Researcher stim	0	0.2466	0.1860

5.2 IMU data

The data collected using the IMUs is described in this section. The three directions of the IMU placed on the torso were not aligned with the sagittal plane, therefore the research team was not able to extract the hip angle in the sagittal plane.

Figure 5.4 shows the ankle and knee angle for the three different conditions of the test in the gait cycle. The resolution of the plots in the x axis is 1% of the gait cycle. The figures show the results of averaging the collected data from all four trials. The heel strike event was extracted from the FSR located at the heel. This event was used to initiate the gait cycle.

In this figure, the black lines show the data without the FES. The blue lines show the data when the participant was triggering the FES. The green lines show the data when the researcher was switching the FES. The ankle angle with the FES deviates from the normal walking's ankle angle. This deviation starts around 65% gait cycle and continues in the entire swing phase until 100% gait cycle.

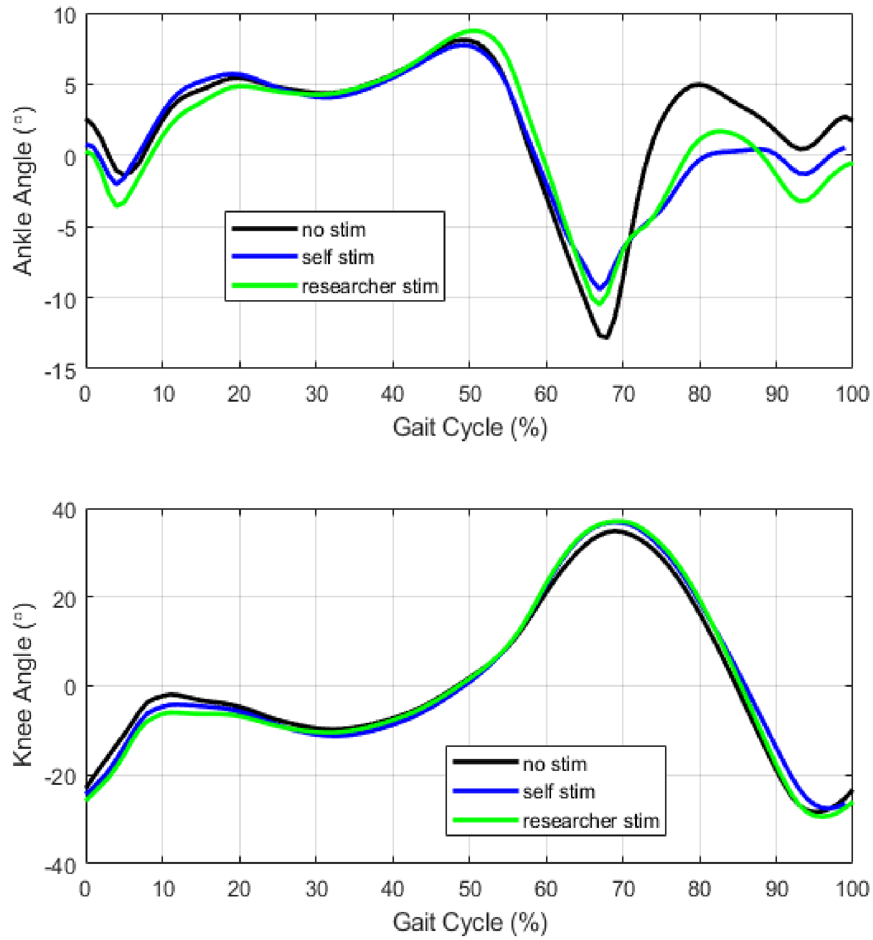

Figure 5.4 Joints angle measured by the IMUs

Figure 5.4 shows the ankle and knee angular velocities for the three different conditions of the test verses gait cycle. The angular velocities were calculated using the forward difference method. This figure shows that the ankle angular velocity is similar in the three different conditions before 68 percent of the gait cycle where the foot switches from the stance phase to the swing phase and the FES switched on. After around 68 gait cycle the angular velocity in the tests with FES deviates from the normal walking angular velocity. As in Figure 5.4, where the knee angle is similar in the three different conditions, the derivative of these curves are also close to each other.

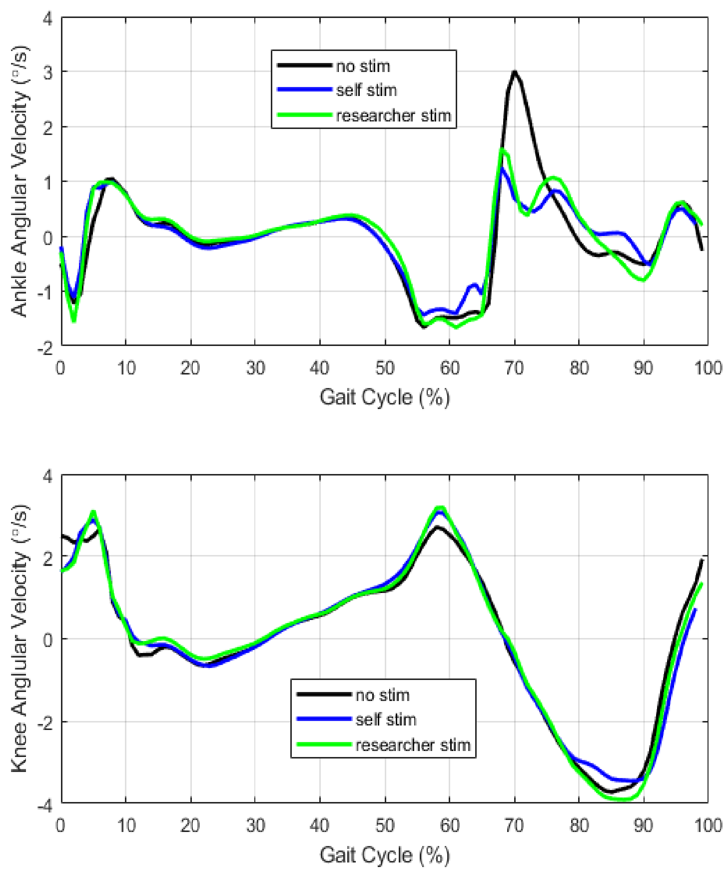

Figure 5.5 Joints velocity measured by the IMUs

Figure 5.5 shows the ankle and knee angle standard deviations for the three different conditions of the test in gait cycle. The standard deviation was calculated to show the variability of the data in gait cycle. The standard deviation of the ankle angle in normal walking has one spike. This spike occurs around 70% gait cycle where the toe off happens. The standard deviation of the tests with stimulation are higher than the normal walking standard deviation. The knee angle standard deviations of the normal walking are less than the trials with the FES.

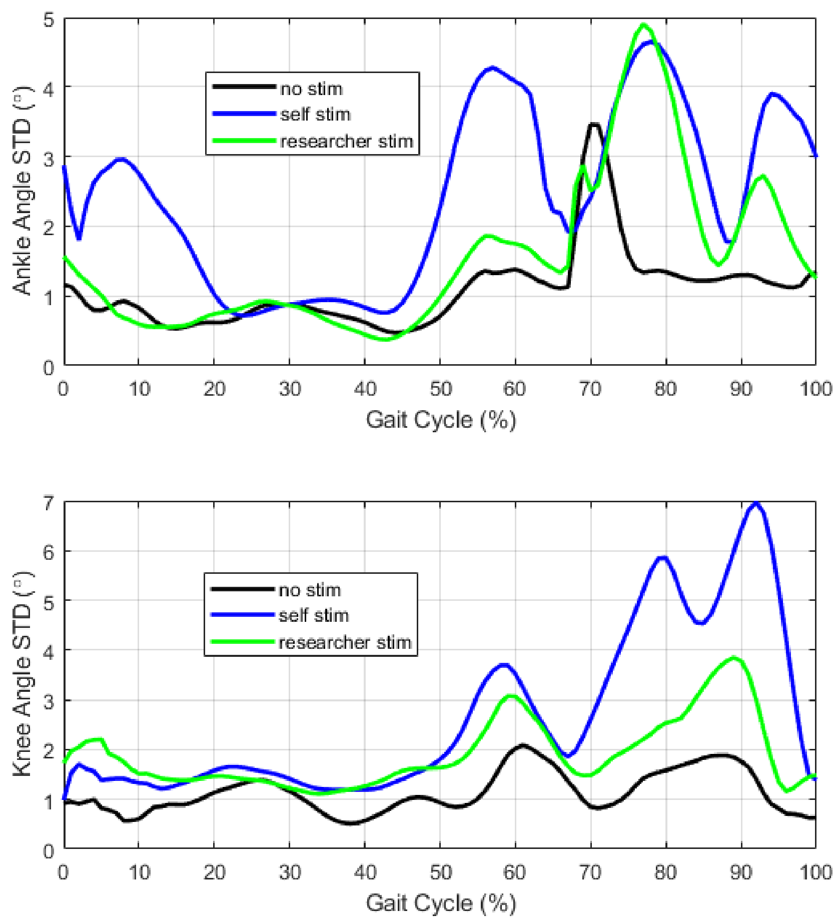

Figure 5.6 Joints angle standard deviation

Table 5.5 and 5.6 show the correlation coefficient of the joints angle and angular velocity in the trials with stimulation compare to the normal walking. Table 5.5 shows that the knee angle has more correlation than the ankle angle. The same scenario for the angular velocity as mentioned in Table 5.6, stating that the stimulation affected the ankle angle more than the knee angle.

Table 5.5 Correlation Coefficient of the joints angle

Correlation Coefficient	Ankle	Knee
Self-stim	0.8966	0.9936
Researcher stim	0.9006	0.9958

Table 5.6 Correlation Coefficient of the joints angular velocity

Correlation Coefficient	Ankle	Knee
Self-stim	0.8506	0.9853
Researcher stim	0.8518	0.9915

Table 5.7 and 5.8 show the RMSE value of the joints angle and angular velocity in the trials with stimulation relative to the normal walking. In Table 5.7 knee angle has more RMSE value than the ankle angle, but it is the opposite in Table 5.8 for the velocity of the knee and ankle angle. Therefore, the FES changed the velocity more than it changed the angle.

Table 5.7 RMSE of the joints angle

RMSE	Ankle	Knee
Self-stim	2.2271	2.2376
Researcher stim	2.2851	2.3279

Table 5.8 RMSE of the joints angular velocity

RMSE	Ankle	Knee
Self-stim	0.4868	0.3078
Researcher stim	0.4646	0.2483

5.3 IMUs and camera motion capture system's data comparison.

In this section, the IMUs and the camera motion capture system's data were represented to observe the differences between these two methods of measurement. The camera motion capture system and IMUs' data need to be compared at the same gait phase to have a proper comparison. Therefore, they need to have the same zero percent gait cycle. The method of detecting the heel strike was different in the IMUs' data with the camera motion capture system's data. Therefore, the heel strike was not necessarily similar in the IMUs and the camera motion capture system's data. The heel strike in the IMUs' data was configurated using the FSR sensor located at the heel, which is a more accurate method of detection than the visual observation. As a result, the camera motion capture data was shifted to have the maximum correlation with the IMU data. This was done by shifting the camera motion capture data in 1% increments starting from 1% to 100% of the gait cycle. The correlation between the shifted camera data and the IMU data was calculated after each increment. Then, the shift value was selected that maximized the correlation function between the two sets of data.

The IMU data has an angular offset from the camera motion capture system's data. This offset is associated with the thickness of the muscle to which the IMU was attached. The camera motion capture system uses multiple markers to measure the joints which is a more accurate method. To find this offset in ankle joint angle between the motion capture systems, the IMU data was shifted in 0.1 degrees increments starting from -5 to 5 degrees. The RMSE between the shifted IMU data and the camera data was calculated at each increment. The offset value that minimized the RMSE was selected.

Figure 5.7 shows the ankle and the knee angle measured by the IMUs and the camera motion capture while the participant walked normally. The IMU and camera data followed the same pattern in the ankle and knee angle. The difference between these two measurement systems is seen in the ankle angle plot during the 20% to 80% gait cycle.

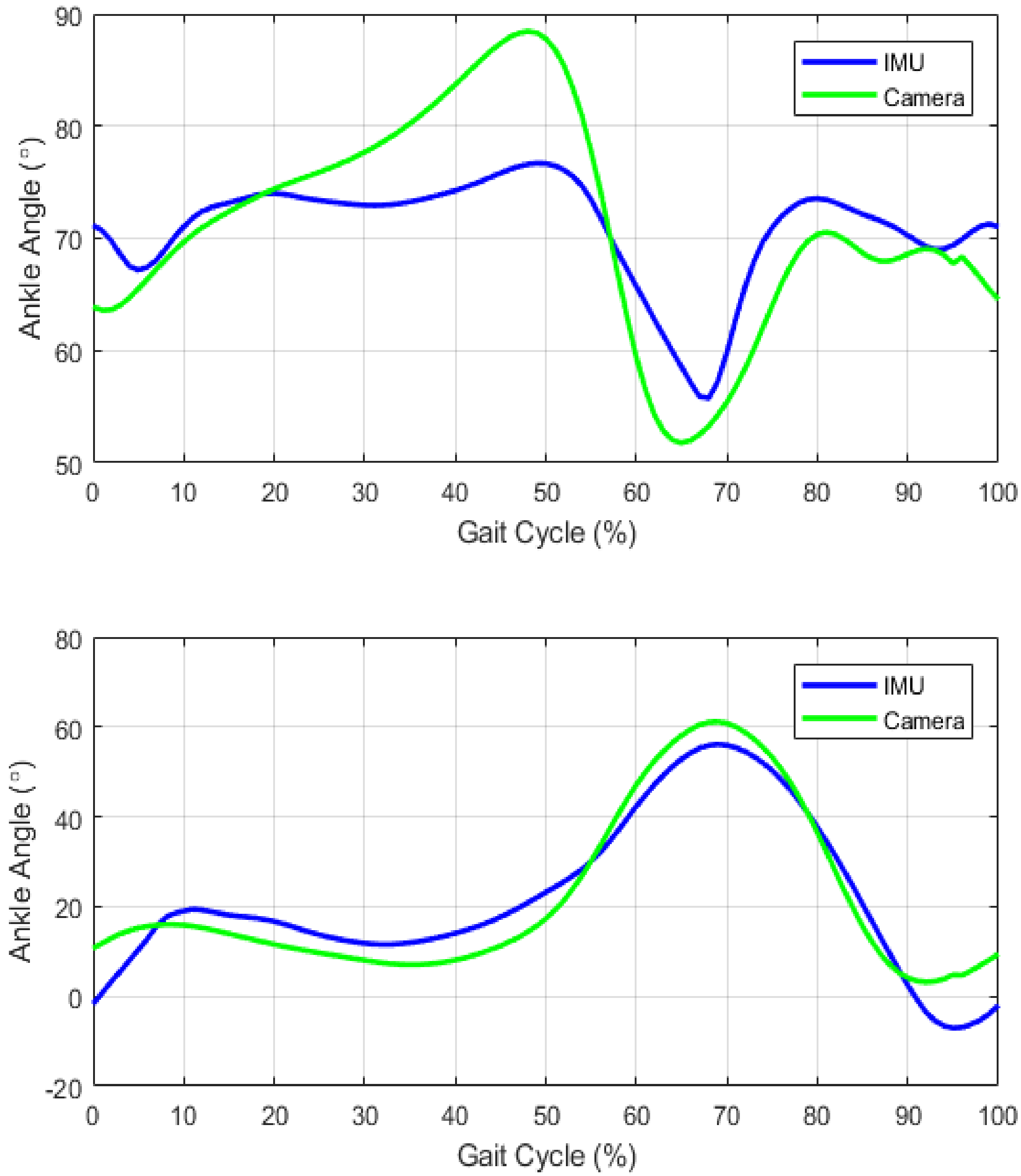

Figure 5.7 Normal walking

Figure 5.8 shows the ankle and the knee angle measured by the IMUs and the camera motion capture while the participant was triggering the FES to stimulate the GN muscle.

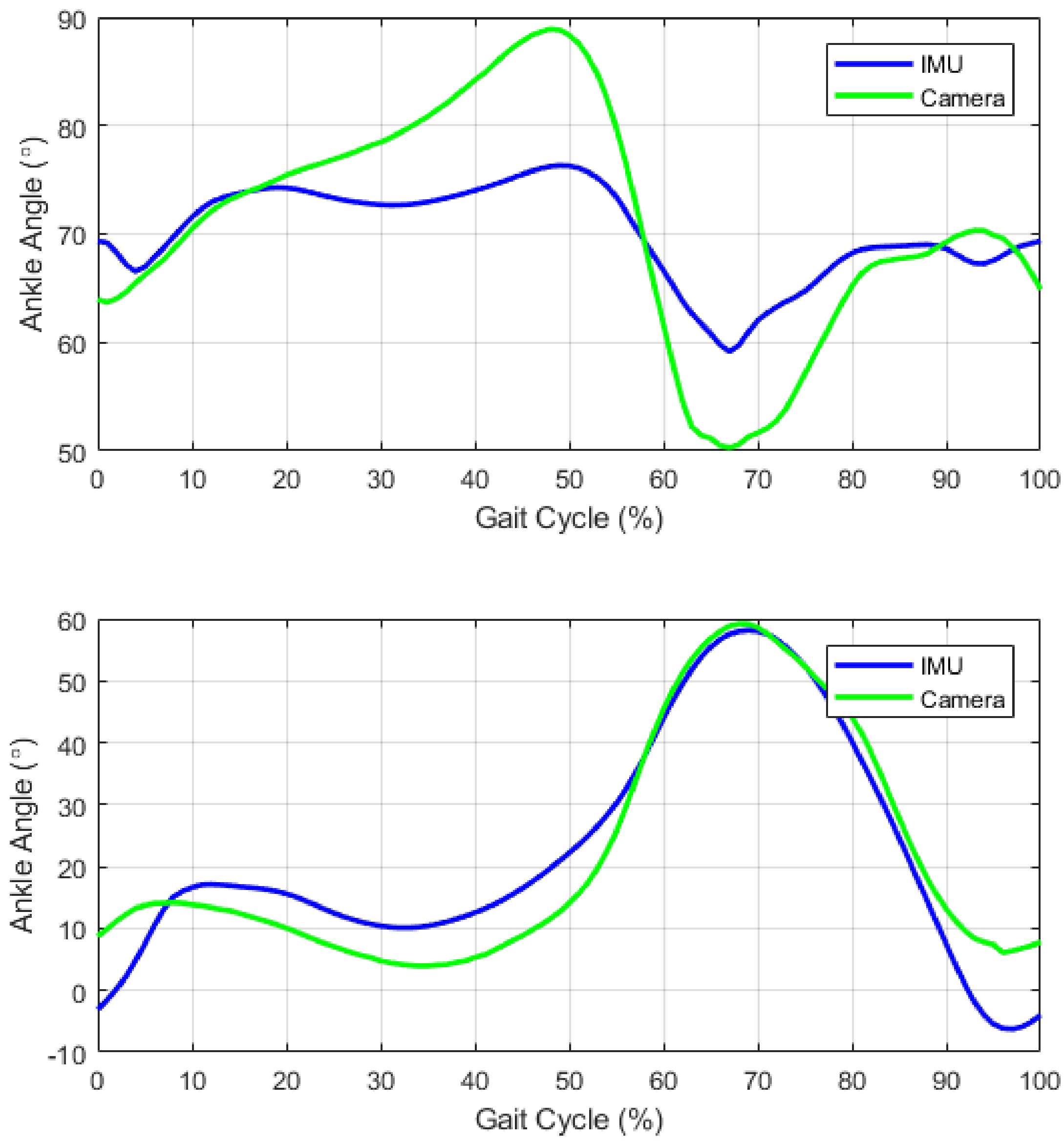

Figure 5.8 Participant was switching the FES

Figure 5.9 shows the ankle and the knee angle measured by the IMUs and the camera motion capture while the researcher was switching the FES. In Figure 5.8 and 5.9, knee angles have similar values. The ankle angle measured by the two measurement systems are closer to each other in Figure 5.8 than Figure 5.9.

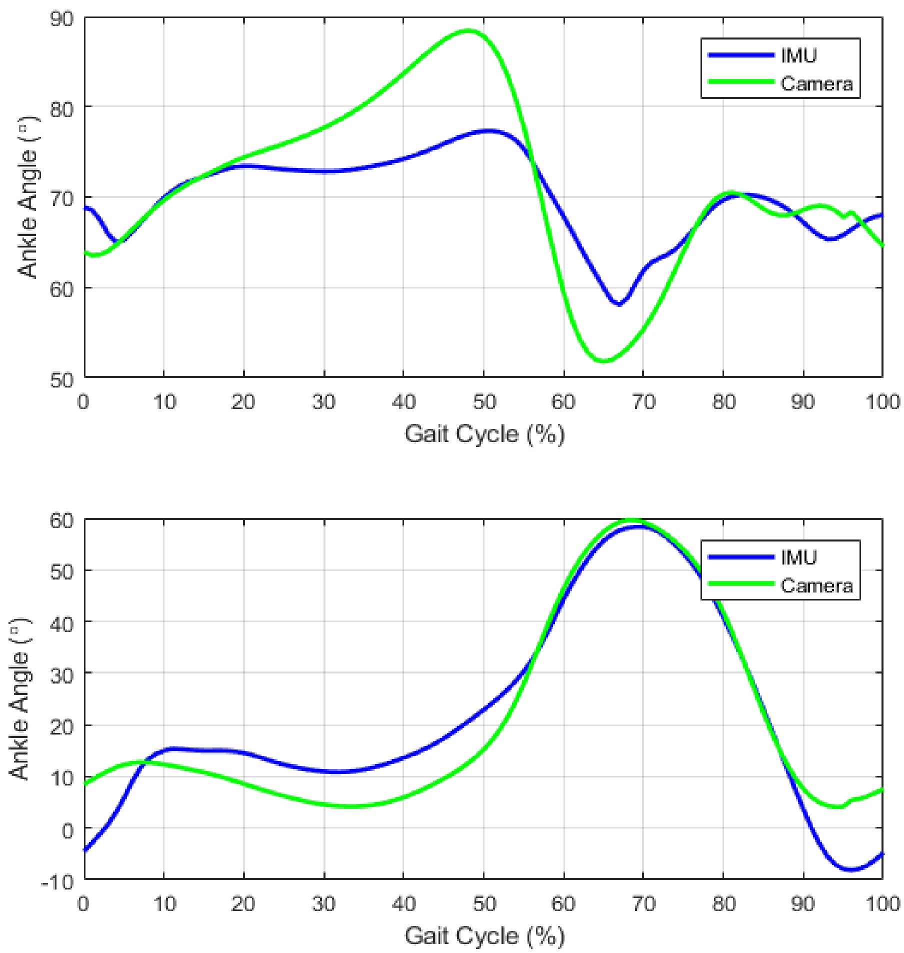

Figure 5.9 The researcher was switching the FES

Figure 5.10 shows the joints angle error in the IMU data measurement relative to the camera motion capture system. The error data is represented in the three different condition of the experiment. In both ankle and knee angle, the errors in three different conditions starts to deviate from each other around 60 percent gait cycle. The ankle angle errors show that the maximum error between the IMU and camera system is about 12 degrees and occurs two time. One around 50 percent gait cycle and another time around 60 percent gait cycle. The maximum error between two methods of measurements for the knee angle is around 12 degrees. The maximum error happens at the beginning and end of the gait cycle. The errors seem to be lowest in the swing phase but not at the end. This error may be due to shifting the camera data to synchronize with the IMUs data.

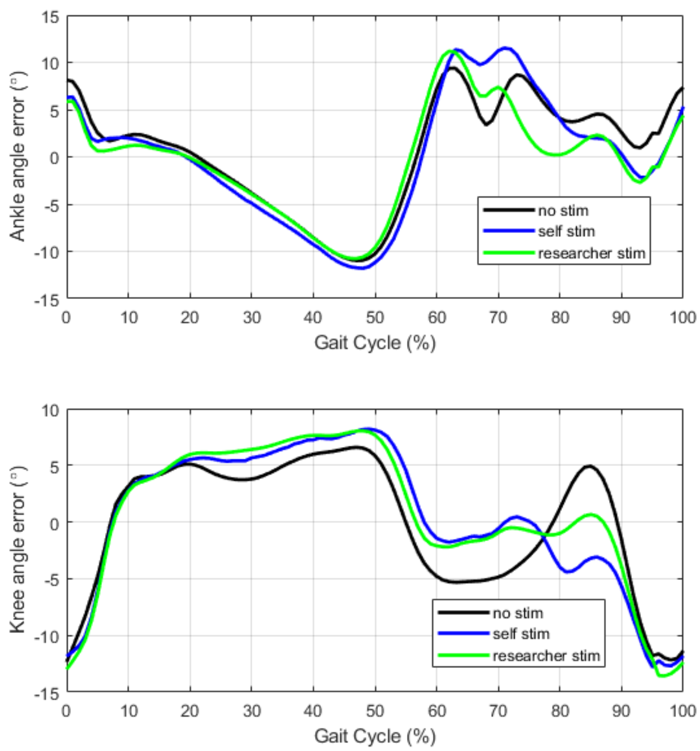

Figure 5.10 IMUs error

Table 5.9 shows the correlation coefficient between the IMU and the camera motion capture data. The correlation coefficient was computed for the ankle and the knee angle.

Table 5.9 Correlation Coefficient of the joints angle

Correlation Coefficient	Ankle	Knee
No-stim	0.8804	0.9503
Self-stim	0.9529	0.9425
Researcher stim	0.9355	0.9460

Table 5.10 shows the RMSE of the IMU data relative to the camera motion capture data. The RMSE values were calculated for the ankle and the knee angle. The average RMSE for this table is about 6 degrees.

Table 5.10 RMSE of the joints angle

RMSE	Ankle	Knee
No-stim	5.8449	5.6686
Self-stim	6.5730	6.2250
Researcher stim	5.4226	6.2601

CHAPTER 6: DISCUSSION

6.1 Comparison between the IMUs and the camera motion capture system

The objectives of the project were to design, build and test a neural prosthesis device to stimulate the gastrocnemius (GN) muscle while walking and obtain movement characteristics from the gait. In this section, the performance of the neural prosthesis is discussed and the strengths and the weakness of the system are explained.

The neural prosthesis device successfully caused contraction on the GN muscle while walking. The level of the stimulation by the FES was enough to cause movement in the ankle and change the gait. The 90% correlation coefficients in Table 5.6 shows that walking with and without the GN muscle stimulation were different. These changes in the gait were observable both visually and by using the motion capture systems. The effect of the GN stimulation on the gait is discussed in more detail in 6.2.

The use of the IMUs were beneficial in studying the effect of the GN muscle stimulation on the gait. The IMUs data were able to indicate the differences in walking with and without the FES (Table 5.5 and 5.6). The accuracy of the IMUs were close to that found in the literature. For example, the RMSE between the IMUs and the camera motion capture system in measuring the knee angle was around 5.67 degrees. This value was more accurate than some similar studies in comparing IMUs and camera motion capture systems; other studies had RMSE around 7 degrees [32] and 6.42 degrees [33]. The average correlation coefficient between the two measurement systems in measuring the knee angle was about 95%. This value was more accurate than the 93% correlation coefficient in a similar study [33]. Although the test results for knee angles measured by the IMUs and the camera motion capture system were close to some literature [32] [33], this system was less accurate than found by previous researchers using the similar IMUs (MPU-9250) [15]. In another study for comparing the knee angle using the similar IMUs and a Vicon camera motion capture system [15], the average RMSE was 1.85 degrees and the correlation coefficient was 99% [15].

The conducted experiment showed that the use of the FSRs to detect the gait events was effective. The FSRs were able to detect contact between the foot and the ground during the experiment. Initially, there were concerns with the effect of increasing the temperature on the FSRs performance. However, the FSRs were able to detect contact with the ground during an hour of conducting the experiment. The test results show that the ratio of the stance phase to the whole gait cycle estimated by the FSRs was 66% (Figure 5.4), which is close to the 60% normal value of this parameter [8].

The sources of the errors were not only from the neural prosthesis device. There were errors on both IMUs and the camera motion capture systems. The possible sources of the errors and the weaknesses of the systems are discussed in the following:

First, the IMUs and the motion capture camera systems were not synchronized. The data from these two measurement systems need to be compared at the same instant in time to have a proper comparison. The research team was unable to synchronize the data during the experiment. Therefore, the synchronization was done in post-processing by converting the measurements in the gait cycle, which is discussed in section 5.3. This method is an estimation and does not guarantee the synchronization. Therefore, it can be a cause of error in the IMUs and the camera motion capture system's test results.

Second, the conversion of the camera motion capture system's data from the time to the gait cycle was not accurate. This is normally done by detecting the heel strike using a force plate, however, one was not available. Therefore, the heel strike moment was detected from watching the video of the experiment, which was not as accurate. The camera motion capture system's data showed that the toe off event happened around 69% of the gait cycle for the participant's normal gait (Figure 5.1). The toe off event happens at 60% of the gait cycle for the average healthy individual (section 2.1.1). This difference shows that the actual heel strike probably happened after the estimated heel strike.

Third, there were errors in the camera motion capture system's data. The ankle angle measured by the camera motion capture system for walking with and without the FES were too close to each other (Figure 5.1) and had 100% correlation (Table 5.1). This shows that there could have been errors in the data collected using the camera motion capture system. Because the camera data was assumed to be accurate, this inaccuracy could have affected the results in comparing the IMUs with the camera motion capture systems.

Fourth, a lower dimension model was used to estimate the foot's high degree of freedom movement. In this project, the foot was assumed to be a rigid segment for simplicity, but the foot has more degree of freedom. For example, in the push off phase, the toes were almost horizontal, having zero angle with respect to the horizon but the top of the foot and the heel were in the air, having a nonzero angle. Yet, there was only one IMU placed on the top of the foot to measure the foot angle (Figure 4.3). On the hand, the camera motion capture system had multiple markers placed on different locations of the foot, giving a better estimation of the foot's multiple degrees of freedom movement. This error is seen in Figure 5.10, where the deviation in the IMU's data compared to the camera motion capture system reached its maximum during the heel off to the toe off. Therefore, the 10 degrees of deviation can be associated with the one-degree of freedom assumption in calculating the ankle angle with the IMU.

Fifth, the IMUs' placement was a cause of error. The IMU for measuring the torso angle was attached to the hip bone as shown in Figure 4.6. After the torso angle was post-processed, it turned out that the IMU was not aligned with the sagittal plane. Hence, this IMU's data was not used to estimate the hip angle. To prevent this error, the IMU should be placed on an upper side of the torso.

At this point, the possibility of the gait analysis using this system needs further study and experiments. As a result, one cannot rely on these test results to validate the IMUs measurement for gait analysis. Thus, these five concerns need be addressed in the future works. Most of these concerns can be addressed in a future experiment using the same device. Also, from the collected data it is not observable that the FES was able to improve the gait.

6.2 Effect of the electrical stimulation on the gait

Observable results were obtained when the FES was used while walking. One observation during the experiments with the FES was that the foot was hitting the ground after the stimulation in some trials. This can be due to two factors: First, the stimulation of the calf in the push off phase, lifted the foot prior to the normal toe-off phase and expedited the initiation of the swing phase. The FES is stimulating the GN but the other muscles such as the hip flexors and the tibial anterior are working normally. Thus, when the toe off was expedited, the hip flexors were still acting similar to the stance phase and did not swing the leg. Also, the tibial anterior muscle cannot compensate for the extra plantarflexion caused by the FES. As a result, the foot dropped during the swing phase and the toe hit the ground in this phase.

Second, when the calf muscle was stimulated manually, because of the human error, the FES could not be turned off exactly in the toe off. Therefore, the FES was turned on in the beginning of the swing phase and caused extra foot plantarflexion, which was another reason for the foot dropping and hitting the toe to the ground.

In this project, the GN muscle was chosen as the target muscle to modify the ankle movement and the gait. It was assumed that GN stimulation would modify the ankle angle and have limited effect on the knee and the hip. Figure 5.1 shows that this was a reasonable assumption; because the knee and hip angle in the stimulation trials had less deviation from the normal walking compared to the ankle angle. Also, the correlation coefficient of the knee and the hip for the trials with stimulation was about 0.99 and around 0.90 for the ankle (Table 5.1) which shows that the effect of the GN stimulation was more on the ankle angle. This is because the GN muscle actuated the ankle's angle directly. When stimulating the GN muscle, the ankle's angle changes but the knee and the hip did not change significantly. Therefore, the changes in the knee and the hip angle by the GN stimulation was not a result of direct actuation on these joints but from the coupling of the dynamic of walking between the calf muscle and knee and hip.

The changes in the ankle angle due to the FES can be explained by looking at the angle, angular velocity and standard deviations plots:

Ankle angle: Figure 5.4 shows the ankle angle was larger in the experiment with the FES than the normal walking. The GN stimulation made the foot more plantarflexion, which caused the heel to move upward and the toe to push downward. This extension in the ankle made the ankle angle larger than normal. The ankle is a dynamical system and has limited stiffness and damping. When switching on the FES, it takes time for the muscle to reach its final state. Also, when turning off the FES, it takes time for the foot to go back to its initial position. Therefore, when turning off the FES after the push off phase, the foot will not go back to the neutral position instantly. As a matter of fact, the foot remains plantarflexion in the entire swing phase

(Figure 5.1 and 5.4). The foot will not reset to its original position until the heel strikes and the foot pushes to the ground (Figure 5.1 and 5.4).

Ankle angular velocity: The angular velocities of the ankle angle were the same in the gait cycle except after the FES was switched off (Figure 5.2 and 5.5). In the normal walking, the angular velocity reached its maximum value at around 70% of the gait phase when the foot moved from plantarflexion to dorsiflexion (Figure 5.2 and 5.5). This shows that turning on the FES had less effect on the gait than turning off the FES. This means that when the FES is switched on, the muscle reached its final state faster because the FES caused more contraction. When the FES is switched off it takes more time for the muscle to return to the resting state.

Ankle standard deviation: The standard deviation of the ankle angle in normal walking was around one for the entire gait cycle, except the toe off phase that the foot switched from the stance to the swing phase (Figure 5.6). In this phase, the standard deviation increased up to 3.5 degrees. In the same phase, the standard deviations of the trials with the FES went up to 4.5 degree (Figure 5.6). The standard deviation of the trials showed that the normal walking had the smallest trial to trial variation. The researcher stimulation trials and the stimulation by the participant himself had the largest variations (Figure 5.6). This showed that FES caused an increase in the standard deviation. This may be because the FES switching time and the muscle response were not consistent in each trial.

CHAPTER 7: CONCLUSION

This research showed that a neural prosthesis utilizing low cost IMUs was able to estimate the joint angle while walking. In addition, the device had an observable effect on the gait when tested on a healthy individual. This experiment showed that only simulating the GN muscle will not be enough. To improves the gait, the level of the stimulation needs to be controlled or more electrodes attached to the other muscles such as the tibial anterior to compensate for the foot drop.

For future work, it is recommended to use a transformation matrix be used to find the orientation of the IMUs. This will help to reduce the dependency of the data on IMU placement. Furthermore, to test the neural prosthesis device, it would better to conduct the experiment on a treadmill which can help to keep the walking speed stationary, which can affect the results. Also, it would be easier to collect data and manually switch the FES by a researcher beside the treadmill. In this project, the FES was switched on manually. For future work, it is recommended to switch the FES on and off automatically. One method for automatically switch the FES is to estimate the gait phase and switches the FES on and off at a specific gait phase. The gait phase can be estimated using two methods: First, a neural network can be trained to estimate the gait phase from the IMUs and the FSRs' data [3]. The neural network should be trained individually for each participant. This method will require to use several sensor's data. Second, an IMU attached on the thigh can be used to estimate the gait phase. The gait phase can be estimated using a single IMU attached on the thigh [34]. This approach will help to reduce the number of sensors.

REFERENCES

[1] National Spinal Cord Injury Statistical Center, Facts and Figures at a Glance. Birmingham, AL: University of Alabama at Birmingham, 2018

[2] Purves, Dale, et al. "Neuroscience. ed." Glutamate. Sunderland (MA): Sinauer Associates (2001).

[3] Meng, Lin, et al. "A functional electrical stimulation system for human walking inspired by reflexive control principles." Proceedings of the Institution of Mechanical Engineers, Part H: Journal of engineering in medicine 231.4 (2017): 315-325.

[4] Schearer, Eric M., et al. "Multi-muscle FES force control of the human arm for arbitrary goals." IEEE Transactions on Neural Systems and Rehabilitation Engineering 22.3 (2014): 654-663.

[5] Hodkin, E. F., Lei, Y., Humby, J., Glover, I. S., Choudhury, S., Kumar, H., ... & Jackson, A. (2018). Automated FES for Upper Limb Rehabilitation Following Stroke and Spinal Cord Injury. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 26(5), 1067-1074.

[6] Farris, Dominic James, and Gregory S. Sawicki. "The mechanics and energetics of human walking and running: a joint level perspective." Journal of The Royal Society Interface (2011): rsif20110182.

[7] The Walkaide System, http://www.walkaide.com/Pages/default.aspx, retrieved 2018-07-31

[8] Deluca, P. A., & Renshaw, T. S. (1995). Gait analysis: principles and applications. Emphasis on its use in cerebral palsy. JBJS, 77(10), 1607-1623.

[9] Tao, W., Liu, T., Zheng, R., & Feng, H. (2012). Gait analysis using wearable sensors. Sensors, 12(2), 2255-2283.

[10] Moore, K. L., Dalley, A. F., & Agur, A. M. (2013). Clinically oriented anatomy. Lippincott Williams & Wilkins.

[11] Donnellan, Clare P., and Kate Caldwell. "TENS and FES for sensory impairment and gait dysfunction following removal of spinal cord ependymoma—a case report." Physiotherapy Research International 14.4 (2009): 234-241.

[12] Guy, J. E., FUNCTIONAL ELECTRICAL STIMULATION RECUMBENT BICYCLE FOR STROKE REHABILITATION, MST Thesis, Western Carolina University, 2013.

[13] Lazar, Eric, and Juan Nicolás Cuenca. "Functional electrical simulation (FES) in stroke." (2008).

[14] Shetty, Yatiraj K. Robust Human Motion Tracking Using Low-cost Inertial Sensors. Diss. Arizona State University, 2016.

[15] Abhayasinghe, Kahala Nimsiri. Human gait modelling with step estimation and phase classification utilising a single thigh mounted IMU for vision impaired indoor navigation. Diss. Curtin University, 2016.

[16] Michał, B., Jakub, D., Marek, I., Mateusz, P., & Jacek, W. (2017). GNIT PHASE RECOGNITION FOR EXOSKELETON CONTROL USING ADAPTIVE NEURO FUZZY INFERENCE SYSTEM. Annals of DAAAM & Proceedings, 28.

[17] Guerra-Filho, G. (2005). Optical Motion Capture: Theory and Implementation. RITA, 12(2), 61-90.

[18] Qualisys Miqus M3, https://www.qualisys.com/cameras/miqus/, retrieved 2018-06-30

[19] FSR 402, https://www.interlinkelectronics.com/fsr-402, retrieved 2018-07-26

[20] Fritzing, http://fritzing.org/home/, retrieved 2018-07-31

[21] MPU-6050, https://www.invensense.com/products/motion-tracking/6-axis/mpu-6050/, retrieved 2018-06-30

[22] Arduino, https://www.arduino.cc/en/reference/wire, retrieved 2018-06-30

[23] MPU-6050 Accelerometer + Gyro, https://playground.arduino.cc/Main/MPU-6050/, retrieved 2018-07-26

[24] EMS-5000, https://www.otcwholesale.com/emsi-ems5000.html, retrieved 2018-06-30

[25] Arduino Uno, https://store.arduino.cc/arduino-uno-rev3, retrieved 2018-06-30

[26] MATLAB, https://www.mathworks.com/products/matlab.html, retrieved 2018-06-30

[27] Teensy 3.6, https://www.pjrc.com/store/teensy36.html, retrieved 2018-06-30

[28] Wilmott, P., Howison, S., & Dewynne, J. (1995). The mathematics of financial derivatives: a student introduction. Cambridge University Press.

[29] Mukaka, Mavuto M. "A guide to appropriate use of correlation coefficient in medical research." Malawi Medical Journal 24.3 (2012): 69-71.

[30] Zhang, Minmin. Multi-sensor inertial measurement system for analysis of sports motion. Diss. University of Pittsburgh, 2015.

[31] Qualisys Track Manager, https://www.qualisys.com/software/qualisys-track-manager/, retrieved 2018-06-30

[32] Takeda, Ryo, et al. "Gait posture estimation using wearable acceleration and gyro sensors." Journal of biomechanics 42.15 (2009): 2486-2494.

[33] Tong, Kaiyu, and Malcolm H. Granat. "A practical gait analysis system using gyroscopes." Medical engineering & physics 21.2 (1999): 87-94.

[34] Quintero, David, et al. "Real-time continuous gait phase and speed estimation from a single sensor." Control Technology and Applications (CCTA), 2017 IEEE Conference on. IEEE, 2017.

APPENDIX A: CODE REVISION ONE

```
#include<Wire.h>
const int MPU_addr_1=0x68;
const int MPU_addr_2=0x69;
int  AcX,AcY,AcZ,Tmp,GyX,GyY,GyZ;
unsigned long time;

void setup() {
  Wire.begin();
  Wire.setClock(400000UL);
  Wire.beginTransmission(MPU_addr_1);
  Wire.write(0x6B);
  Wire.write(0);
  Wire.endTransmission(true);
  Wire.beginTransmission(MPU_addr_2);
  Wire.write(0x6B);
  Wire.write(0);
  Wire.endTransmission(true);
  Serial.begin(250000);
}

void loop() {
  time = millis(); Serial.print(time);Serial.print('\t');

  Wire.beginTransmission(MPU_addr_1);
  Wire.write(0x3B);
  Wire.endTransmission(false);
  Wire.requestFrom(MPU_addr_1,14,true);
  AcX=Wire.read()<<8|Wire.read();
  AcY=Wire.read()<<8|Wire.read();
  AcZ=Wire.read()<<8|Wire.read();
  Tmp=Wire.read()<<8|Wire.read();
  GyX=Wire.read()<<8|Wire.read();
  GyY=Wire.read()<<8|Wire.read();
  GyZ=Wire.read()<<8|Wire.read();
  Wire.endTransmission(true);
  Serial.print(AcX);Serial.print('\t');
  Serial.print(AcY);Serial.print('\t');
  Serial.print(AcZ);Serial.print('\t');
  Serial.print(GyZ);Serial.print('\t');
  Serial.print(GyY);Serial.print('\t');
  Serial.print(GyZ);Serial.print('\t');
```

```
  Wire.beginTransmission(MPU_addr_2);
  Wire.write(0x3B);
  Wire.endTransmission(false);
  Wire.requestFrom(MPU_addr_2,14,true);
  AcX=Wire.read()<<8|Wire.read();
  AcY=Wire.read()<<8|Wire.read();
  AcZ=Wire.read()<<8|Wire.read();
  Tmp=Wire.read()<<8|Wire.read();
  GyX=Wire.read()<<8|Wire.read();
  GyY=Wire.read()<<8|Wire.read();
  GyZ=Wire.read()<<8|Wire.read();
  Wire.endTransmission(true);
  Serial.print(AcX);Serial.print('\t');
  Serial.print(AcY);Serial.print('\t');
  Serial.print(AcZ);Serial.print('\t');
  Serial.print(GyZ);Serial.print('\t');
  Serial.print(GyY);Serial.print('\t');
  Serial.println(GyZ);
}
```

APPENDIX B: CODE REVISION TWO

```
#include<i2c_t3.h>
const int MPU_addr_1=0x68;
const int MPU_addr_2=0x69;
int16_t  AcX,AcY,AcZ,Tmp,GyX,GyY,GyZ;
unsigned long time;

void setup() {
  Wire.begin();
  Wire.setClock(400000UL);

  Wire.beginTransmission(MPU_addr_1);
  Wire.write(0x6B);
  Wire.write(0);
  Wire.endTransmission(true);

  Wire.beginTransmission(MPU_addr_2);
  Wire.write(0x6B);
  Wire.write(0);
  Wire.endTransmission(true);

  Wire1.begin();
  Wire1.setClock(400000UL);

  Wire1.beginTransmission(MPU_addr_1);
  Wire1.write(0x6B);
  Wire1.write(0);
  Wire1.endTransmission(true);

  Wire1.beginTransmission(MPU_addr_2);
  Wire1.write(0x6B);
  Wire1.write(0);
  Wire1.endTransmission(true);

  Serial.begin(250000);
}

void loop() {
  time = millis(); Serial.print(time);Serial.print('\t');
  Serial.print(analogRead(0));Serial.print('\t');
  Serial.print(analogRead(6));Serial.print('\t');
  Serial.print(analogRead(7));Serial.print('\t');
  Serial.print(analogRead(8));Serial.print('\t');
```

```
Serial.print(analogRead(9));Serial.print('\t');

Wire.beginTransmission(MPU_addr_1);
Wire.write(0x3B);
Wire.endTransmission(false);
Wire.requestFrom(MPU_addr_1,14,true);
AcX=Wire.read()<<8|Wire.read();
AcY=Wire.read()<<8|Wire.read();
AcZ=Wire.read()<<8|Wire.read();
Tmp=Wire.read()<<8|Wire.read();
GyX=Wire.read()<<8|Wire.read();
GyY=Wire.read()<<8|Wire.read();
GyZ=Wire.read()<<8|Wire.read();
Wire.endTransmission(true);
Serial.print(AcX);Serial.print('\t');
Serial.print(AcY);Serial.print('\t');
Serial.print(AcZ);Serial.print('\t');
Serial.print(GyX);Serial.print('\t');
Serial.print(GyY);Serial.print('\t');
Serial.print(GyZ);Serial.print('\t');

Wire.beginTransmission(MPU_addr_2);
Wire.write(0x3B);
Wire.endTransmission(false);
Wire.requestFrom(MPU_addr_2,14,true);
AcX=Wire.read()<<8|Wire.read
AcY=Wire.read()<<8|Wire.read();
AcZ=Wire.read()<<8|Wire.read();
Tmp=Wire.read()<<8|Wire.read();
GyX=Wire.read()<<8|Wire.read();
GyY=Wire.read()<<8|Wire.read();
GyZ=Wire.read()<<8|Wire.read();
Wire.endTransmission(true);
Serial.print(AcX);Serial.print('\t');
Serial.print(AcY);Serial.print('\t');
Serial.print(AcZ);Serial.print('\t');
Serial.print(GyX);Serial.print('\t');
Serial.print(GyY);Serial.print('\t');
Serial.print(GyZ);Serial.print('\t');

Wire1.beginTransmission(MPU_addr_1);
Wire1.write(0x3B);
Wire1.endTransmission(false);
Wire1.requestFrom(MPU_addr_1,14,true);
AcX=Wire1.read()<<8|Wire1.read();
AcY=Wire1.read()<<8|Wire1.read();
```

```
  AcZ=Wire1.read()<<8|Wire1.read();
  Tmp=Wire1.read()<<8|Wire1.read
  GyX=Wire1.read()<<8|Wire1.read();
  GyY=Wire1.read()<<8|Wire1.read();
  GyZ=Wire1.read()<<8|Wire1.read();
  Wire1.endTransmission(true);
  Serial.print(AcX);Serial.print('\t');
  Serial.print(AcY);Serial.print('\t');
  Serial.print(AcZ);Serial.print('\t');
  Serial.print(GyX);Serial.print('\t');
  Serial.print(GyY);Serial.print('\t');
  Serial.print(GyZ);Serial.print('\t');

  Wire1.beginTransmission(MPU_addr_2);
  Wire1.write(0x3B);
  Wire1.endTransmission(false);
  Wire1.requestFrom(MPU_addr_2,14,true);
  AcX=Wire1.read()<<8|Wire1.read();
  AcY=Wire1.read()<<8|Wire1.read();
  AcZ=Wire1.read()<<8|Wire1.read();
  Tmp=Wire1.read()<<8|Wire1.read();
  GyX=Wire1.read()<<8|Wire1.read();
  GyY=Wire1.read()<<8|Wire1.read();
  GyZ=Wire1.read()<<8|Wire1.read();
  Wire1.endTransmission(true);
  Serial.print(AcX);Serial.print('\t');
  Serial.print(AcY);Serial.print('\t');
  Serial.print(AcZ);Serial.print('\t');
  Serial.print(GyX);Serial.print('\t');
  Serial.print(GyY);Serial.print('\t');
  Serial.println(GyZ);
}
```