Mechanical Design of an Instrumented Cane for Gait Prediction by Physical Therapists

By

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Thesis

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Chapter 1
Introduction

1.1 Research Motivation

Among the American adult population aged 65 and older, falls are the leading cause of both fatal and non-fatal injuries. In fact, one in three adults in this category will experience a fall each year [1] [2]. As a person ages their ability to walk unaided decreases, making it harder to get around. In order to reduce this risk of falling and maintain a level of independent mobility, an individual may use a walking aid such as a cane or walker. As the baby boomer generation begins to age, the use of assisted walking devices has already increased by 50% since 2004 [3]. As of 2013, adults 65 and older made up 14.1% of the entire population in the United States [4]. Given this and the expectation that this population group will nearly double between the years 2012 and 2050, a reasonable conclusion can be reached that the use of walking aids will continue to increase during this period [5].

Due to their low cost and abundance of availability canes are the most commonly used assisted walking device, with ten percent of the 65 and older population using such devices [6]. However, with many of these individuals never receiving proper instructions on how to select and use the devices, their risk of experiencing an injury can actually increase. If an individual relies on a cane to assist with walking, the risk of experiencing a fall or other injury due to the improper use of the cane could be decreased if the cane itself could help detect such cases and notify the user through some type of interaction.

Physical therapists can analyze an individual’s gait and, based on their observations, can make a recommendation on whether or not that person should use a walking aid. They are
also able to educate individuals on the correct selection and use of such a device. Analysis of a person’s gait and their corresponding risk of experiencing a fall is done primarily through the use of one of the following major gait assessments: the Dynamic Gait Index (DGI), Functional Gait Assessment (FGA) and Balance Evaluation Systems Test (BESTest) [7] [8]. The DGI consists of eight walking movements that are scored on a scale of zero to three, with a score less than or equal to nineteen indicating a high probability of a fall occurring. Despite its validation as an assessment tool, the scale of the DGI leaves the possibility of not fully capturing a person’s balance during a given activity. As such, a person may score high on the DGI, yet still have a high likelihood of experiencing a fall [8]. The FGA aimed to improve upon the DGI, and the BESTest combined aspects of several balance assessments in order to get a better picture of an individual’s overall balance by assessing their ability in more specific situations.

Using a physical therapist to help with one’s gait can be a great help and lower the risk of experiencing a fall. While their trained eyes can be the best tools to determine if a person’s overall posture are correct or not minute differences, such as pressure changes while gripping a handle or between the base of the device and the ground, will be harder to detect. If a therapist is given access to data at a more detailed level, the ability to assess an individual’s gait improves. With the older population continuing to grow, and with it the use of canes, improving the ability to measure and correct a person’s gait, as well as instructing them on the proper use of walking devices increases the safety of not only the individual, but others around them.
1.2 Research Scope

The goal of this work is to design an instrumented walking device and determine its usefulness in a clinical setting as a diagnostic tool. An initial iteration of this concept was previously done by students in the Robotics and Autonomous System Laboratory (RASL) at Vanderbilt University. The device was improved by redesigning the previous electromechanical system around a standard offset cane. Given the shape of an offset cane (described in Section 2.1.2) and the number of sensors required of such a device, coming up with a design which creatively utilized the space available was the primary focus of the project. As such, a large portion of the project was spent on the design of the handle of the cane which is used to house the design’s electronics, as well as rapid-prototyping housings for the other sensors. Other additions to the system included a load cell at the base of the cane and an ultrasonic range finder to detect possible objects at the foot and lower-leg levels.

The handle is one of the most important components of a cane since it is how the user interacts with the device. If a handle is too bulky, hard to grip, or uncomfortable, a user will be less likely to use the device [9]. The handle is also an area of high concern to a physical therapist, as using a cane with an incorrect grip can lead to an increased risk of a fall or other injury, such as carpal tunnel syndrome [10]. Several Force Sensing Resistors (FSRs) were placed at strategic points on the handle in order to gauge the amount of pressure a user applies at various points in their grip. By adding sensors at the handle, a physical therapist could see if an individual’s grip changes before, during, or after specific maneuvers, or notice if pressure is being applied unequally and make adjustments accordingly. If a physical therapist is not used and an individual uses a walking device with improper technique, over
time they may develop other injuries, such as discomfort in the wrist joint. This is important as it was shown in a study by Betani and Marki [11] that 30 to 50 percent of users will stop using a walking aid soon after acquiring it. By providing a device that logs load data, both the user and clinicians can become aware of when a user’s technique is faltering and fix it, lowering the odds that the device will be abandoned. However, adding sensors that a user would directly interact with presented its own challenges. A thin layer would need to cover the handle in order to protect the sensors without decreasing their measuring effectiveness. The protection layer could not be too thin though or there would be noticeable bumps along the handle where the sensors were located, making it awkward to grip. Finding a balance would result in a handle that was both an effective measuring tool and comfortable for the user.

Given the size constraints of a standard cane (a typical cane diameter is 19.05mm), designing a system to fit so many sensors and electronic components would prove too difficult just a few years ago. However, given advancements in technology these same devices can now fit in a much tighter space, allowing for a greater number of components and features to be introduced into the system as described in Section 3.4. The components in the handle were connected to a load cell added near the base of the shaft using a USB 3.0 connector. The load cell replaced a single FSR that was used in the initial iteration of the cane, and measures the axial force experienced by the shaft. The addition of the load cell required rapid-prototyping a housing that would hold/protect the sensor as well as connect the shaft and base of the cane together. Due to the combined power draw of the electronics, a single 3.7 V battery would not allow for the system to be used for an extended period of time. The
voltage of the battery could not simply be increased as doing so would exceed the power rating of the electrical components. This, along with the small volume to work with in order to keep the system within the confines of the diameter of a typical offset cane, required creating a custom battery configuration.

The final aspect of the project was the addition of an ultrasonic range-finder sensor above the load cell housing on the shaft. This sensor provided a method of detecting possible obstacles that are near the user’s feet, which may go unnoticed and lead to a serious fall. The sensor is connected to a vibration motor in the handle that was programmed to turn on if an obstacle is detected within one meter in front of it. The added feature not only increases the user’s overall level of safety, but also provides reliable feedback data to both the user by increasing their awareness of their surroundings, and the physical therapist by recognizing if certain situations are more likely to cause the individual to fall.

1.3 Thesis Outline

This thesis begins with a discussion of background work in Chapter 2. An overview of the different types of canes available in today’s market is first covered. The discussion of the variety in canes is important so that the reader may better understand the differences between the types as well as the decision to switch cane types between the initial design and the design covered in this project. Then, a description of the initial cane design is discussed, as it was the precursor to this project. Following this, previously related research conducted by other parties will be discussed as they were influential on development decisions made throughout this project.
Chapter 3 discusses the design process and sensor selection for the various aspects of the cane that were considered for this project. Chapter 4 provides an overview of the electronics system used in the design. This chapter briefly describes the components in the system and the programming interface used. This part of the system was designed over two iterations, first by a group of multidisciplinary students at Vanderbilt University (VU) prior to this project, with the second iteration of the system designed by Joshua Wade in the RASL at VU. Chapter 5 discusses the procedures used and data collected in order to validate each sensor in the design. In Chapter 6, recommendations of improvements for future iterations of the design as well as other future work that could come of this project are discussed. To evaluate the prototype’s effectiveness as a diagnostic tool, participants will be asked to take part in a study representing a modified DGI gait assessment. Each task will be performed using the prototype under the supervision of a certified physical therapist, with the participants comprised of two distinct adult population groups. In addition to future implementations and improvements, an overview of this study and its associated procedures are also discussed in Chapter 6. The report concludes with Chapter 7 by outlining the contributions that have been made by this thesis.
Chapter 2

Background

2.1 Survey of Assisted Walking Canes

Before discussing the design of this project, it is important to review the different features of walking canes and the different types of canes that are available in today’s market. It is important to provide some background on this topic as it will provide the reader with a better understanding of what features are being modified as they are discussed in Chapters 3 and 4. In addition, it will provide the reader with a better understanding as to why it was decided to switch the cane type that was used in the system between the iteration prior to this project, and the current iteration.

A walking cane consists of four major components: a base, shaft, collar, and handle (see Figure 2.1). The base of the cane is the part of the cane that interacts with the surface the user is walking on. Due to this constant interaction, a ferrule is typically placed over each contact point, or tip, the cane has with the ground to increase its durability and provide a more slip-resistant base. Ferrules were initially used in previous centuries due to the poor road conditions, which would cause the cane (which at the time was made with wood) to
wear down quickly [13]. While ferrules then were usually made out of metals or bone and varied in height, ferrules used in walking canes today are typically made of rubber, only a few inches tall, and used for improving the stability of the device and increasing the contact grip experienced between it and the ground. Depending on the desired increase in stability they can vary in diameter, although it is most common to see a diameter that is slightly larger than that of the cane’s shaft. See Figure 2.2 for a few examples of ferrules found on today’s common walking cane.

![Ferrules examples](image)

(a) Three-Tip [14]  (b) Metal [15]  (c) Rubber [16]

Figure 2.2: Examples of Ferrules used with Walking Canes

The shaft of the cane is the component that connects the ferrule with the handle, and is typically made of either wood or aluminum, with an emphasis placed on a lightweight design. Aluminum canes have become the most common in today’s society as they can be made so that the user can adjust the height to fit their body. This is an important feature as using a cane that is an incorrect size for the user can be painful and possibly create balance issues, as shown in Figure 2.3. The shape of the shaft will vary depending on the type of cane that is used. A collar will go between the cane’s handle and shaft, and are for strengthening
the shaft where the handle attaches. A collar can also be styled to make the cane more aesthetically pleasing to the owner.

![Figure 2.3: Visualization of How Cane Height Affects Posture [17]](image)

Handles come in all shapes and sizes, and are designed with different purposes in mind. As such, each type has its pros and cons depending on the use case. The optimal handle type for a use may vary depending if the user is using the device to support their weight, or to help keep their balance [18]. The material of the handle can vary widely, from wood to ABS plastic, based on the user’s aesthetic preferences and support needs. As manufacturing capabilities have advanced, more customizable ergonomic and personalized canes have also become readily available in order to reduce stress and increase comfort as much as possible. An overview of various walking cane types will now be presented.

### 2.1.1 Standard Cane

The standard cane is the device that most individuals think of when they think of a walking cane. It consists of a shaft that is typically made from either a wood or aluminum,
has one surface contact point, and a handle which can come in a variety of styles. Two such examples are the crook and derby style handles. Older style canes have a curved, or crook handle, and are liked for their look, although they provide little support. Another popular handle for a standard cane is the derby handle. The derby handle is thick compared to the shaft and other handle type, and follows a pattern similar to a long-period sine wave that fits the generic contour of your hand. Derby-style handle canes are widely supported by medical professionals as providing adequate support [19]. Examples of crook- and derby-style standard canes are shown below in Figure 2.4. Due to their design and the placement of the hand relative to the shaft, the standard cane is better suited to those who need more of an aid for simple balance rather than weight-bearing support [20].

![Figure 2.4: Examples of Walking Cane Handles](image1)
(a) Derby Style [21]  (b) Crook Style [22]
2.1.2 Offset Cane

Similar to a standard cane, the offset cane has one surface contact point. However, the design of the shaft differs so that the placement of the hand is over the cane’s center of gravity. Rather than being perpendicular to the handle of the cane like with a derby-style standard cane, the shaft of an offset cane curves from vertical at the base to horizontal at the handle, as shown in Figure 2.5. The device used for the handle then encloses the horizontal portion of the handle, increasing its strength and the amount of weight that can be supported. This lends it to be much more beneficial to individuals who need the device to support their weight.

Figure 2.5: Offset Cane [23]
Following the curvature of the shaft from the base to the handle, the handle-end of the shaft should go towards an individual’s back as this allows for the weight bearing load of the hand to go into the bend of the shaft. However, due to its shape, without proper education on its use many use the device backwards with the handle-end of the shaft pointing to the front of the person [10]. If properly educated on the device, an offset cane is a popular choice by medical professionals because of the additional support it provides over the standard cane.

2.1.3 Quad Cane

For individuals who need more support than an offset cane can provide, yet do not quite need a walker, the quad cane is an available option. Rather than one, the quad cane (as its name implies) has four contact points with the surface, as shown below in Figure 2.6. The base is typically made out of metal, with two “branches” making up the four contact points and wider base. Quad canes come in a variety of base sizes, depending on the required level of support needed.

With four points of contact, the weight-bearing support this cane provides is substantially higher than the previously two canes described. As this class of cane is determined more by its base component, quad canes can come in either the straight/standard or curved/offset type of shaft, along with the variety of handles previously mentioned.
2.2 Initial Iteration Overview

Wade et al. [25] developed the initial system that this thesis is based on. Their initial design had many of the same design requirements that were required of the second revision. Their goal was to equip an off the shelf device with various sensors in a way that made it indistinguishable from other canes. The less the device would “stick out” in the eyes of the public, the higher its possible societal acceptance. These sensors would log position, orientation, and force data as the user interacted with the device. All of the components are housed either in the handle or the base of the cane. Their design centered around a standard, 19.05 mm diameter aluminum cane, with a rapid-prototyped derby-style handle. By designing a handle that housed all of the electronics, they substantially increased the modularity of the system. The base of the cane was also fabricated with modularity in
mind. The base attaches to the shaft with a spring detent, and allows for the hot-swapping of several types of ferrules, so long as they are compatible with a 19.05 mm diameter cane. While multiple handles were produced in order to test various battery sizes, this could easily be extrapolated to catering towards user preferences. Allowing users to customize the device towards their liking (e.g. they may prefer a thicker handle and a quad-contact base over a thinner handle and single-contact base) also boosts such a system’s market viability.

![Figure 2.7: Initial Iteration Force Sensing Resistor Arrangement and Locations](image)

Eight force sensing resistors (FSR) are used in the system, with seven attached to the outside of the handle, and one in the base of the cane in line with the shaft. These sensors measure the force applied at various points of contact in the user’s grip, as well as the axial reaction force of the ground as the user walked with the device. The FSRs were strategically placed on the handle by observing how several individuals gripped the cane. The handle and base of the cane, as well as the placement of the FSRs can be seen in Figure 2.7 (a) and Figure 2.7 (b), respectively.

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The electronics contain six main components: a battery, a power management unit, a wireless microcontroller, two separate 9 degree-of-freedom (DOF) IMU’s, and an analog to digital converter (ADC) to translate data received from the FSRs. An IMU was placed in the base and the handle of the cane to acquire linear acceleration from both ends of the device. The remaining components are contained within the handle in a custom-made, 14W x 14H x 35L mm rapid-prototyped housing (Figure 2.8). The IMUs and ADC were controlled by the wireless microcontroller, as shown in Figure 2.9. The housing also had a connector at either end, one to connect to the battery, and the other to allow for device reprogramming without needing to disassemble the housing. Keeping all of the electronics contained within the confines of the handle was a product of two design choices: using a single battery, and custom-made printed circuit boards (PCB). Using a single battery, while decreasing the use-time that could be extracted of the device, freed up valuable space within the handle. Custom PCBs allowed the group to creatively design the boards to better fit
within the handle. The circular boards had a diameter of 9.8 mm and a thickness of 1.6 mm, and were connected by soldering wires between their pads.

Figure 2.9: Initial Iteration’s System Interconnect [25]

Their results from a system validation study that was similar to the one described in Chapter 6.1 indicated that their device could be able to recognize several walking activities through gait patterns and sensor responses. With that said, their design had some sticking points that could be addressed to improve the system. While the single battery allowed for it to be placed in the handle, the resulting battery life was too low to be considered for future iterations. Also, while the design’s primary goal was research-oriented in a clinical environment, ideally such a device could be owned by an individual to use in their day-to-day life, rather than only during physical therapy appointments. Although the prototype was
low-cost by design, if the design were to be marketed to the public any sensors would result in a higher cost to the buyer than a non-instrumented cane. The first iteration provided several benefits to a physical therapist, but none that directly an end-user. In order to convince someone that they should spend even a dollar more for a product when there is a cheaper alternative, there would need to be features that make someone without the knowledge of the more advanced sensors want to buy the product. By adding such features using low-cost sensors, the overall hypothetical cost of the product would minimally increase while making the device more marketable to the public. However, their research using this prototype showed that there could be merit in using such a device in a clinical setting and further research was warranted.

2.3 Literature Review of Related Research

As previously mentioned, given the current and expected continued increase in the aging population, there has been a substantial amount of research conducted in recent years regarding assistive devices from a wide variety of approaches. While this paper focuses on a passive, untethered system, much research has been done on analyzing an individual’s gait via robotic devices and/or wearable sensors. Wang et al. [26] designed a three-wheeled walker with a three-dimensional accelerometer and a set of encoders to analyze the differences in gait patterns between young adults and adults 69 and older. While the younger group was able to typically move at a slightly faster pace, it took them a bit longer to move from point A to point B. They found that the younger group leaned on the device more than their counterparts, thus altering their gait. While the majority of walking device users are elderly, many will use a device while rehabilitating an injury. If the walking aid needs to
be used for a significant amount of time and the user isn’t properly trained on how to walk with one, their long-term gait could be affected. Hassan et al. [27] developed a gait analysis prototype using an instrumented cane, and a single-leg exoskeleton. The exoskeleton used an inertial measurements unit (IMU) on each section of the leg, while force sensors were placed in the insoles of their shoes to provide insight on the experienced reaction forces with the ground as they walked. Their cane acted as the main hub for all of the data received by the other sensors, and sent the responding commands back to the exoskeleton via wifi. By using a combination of sensors at various points on a user’s body, they were able to analyze an individual’s gait from a more holistic approach. However, while wearable systems will always be able to provide more accurate data than a separate device, they are much more difficult to apply in non-research environments.

In addition to general gait analysis, research has been conducted in this area to try and either improve the clinical rehabilitation process or improve the lives of those individuals that require a walking aid to maintain their independent mobility. Kim and Cho [28] attached an ultrasonic sensor to a white cane (a type of cane typically used by those with visual impairments). Individuals that took part in their study responded favorably to the device as this provided users with more of an advanced warning of potential obstacles than is normally given using the white cane. Lacey and MacNamara [29] also focused on the blind by developing the Personal Adaptive Mobility AID (PAM-AID), although their device was geared more specifically to the elderly. While its large size would place it closer to the category of a walker the long-running project, which spanned 6 years from its inception to the publishing date of the paper (and continued on after), and its multiple trials provided
a lot of valuable feedback about what users looked for most in a walking aid. They tested a device in passive, active, and shared-use modes, and questioned users on their opinions regarding the safety, reliability, learnability, and overall usability of each. The primary sensor on the device was a laser-range finder which was used to detect obstacles and turns in the environment. The results predictably found that as the severity of an individual’s visual impairment decreased, the less favorable they were towards the active mode as they preferred the higher sense of control of their movement.

Mercado et al. [30] developed a system using a quad cane that provided auditory feedback based on the force applied to the handle by the user. This is useful to the user as it provides them with the ability to understand when they are using the correct technique or not, which can be difficult if they are just starting to use a walking aid. Auditory feedback can also be useful to a physical therapist as they can hone in on more precise points in time where the user’s technique falters. By better knowing when an individual struggles with mobility the therapist can spend more time working with the individual on these specific situations. The decision to include vibration motors as the primary feedback device in this project was influenced by Mercado’s design. Culmer et al. [31] developed a prototype specifically for use as an evaluation tool in a clinical setting. They developed a custom load cell to measure the axial load applied, and recorded inertial information using a wireless module that attached to the cane. As stated in [25], “this system [one with a force sensing resistor in the cane shaft rather than a load cell] demonstrated good orientation tracking and load monitoring, but force and orientation information alone may not have the capacity to detect gait patterns that can predict fall-likelihood or other anomalies associated with the gait”. The use of a
load cell was thus incorporated into this project as it provided a more precise measurement of the axial force being applied by the user at the handle.

Finally, a lot of research has been spent studying fall detection and prevention, a typical area of concern for walking aid users. Di et al. [32] presented a concept coined “intentional direction” (ITD), used to help individuals using a walking aid during normal (non-falling) walking situations. Their design was built to optimize three main uses related to a walking aid: using a cane as a guide, to help with rehabilitation, and for fall prevention. Using an individual’s natural tendency to move in the direction they look, researchers were able to use the change in force as an individual turned to help guide the device in the intended direction of movement. A laser range finder was used to reinforce the intended direction of the user by detecting corners and obstacles within the environment. The group also proposed a method for detecting falls by using a particle filter to dynamically estimate the center of gravity using the individual’s body position and load placed on the cane handle. They also proposed using a low-level impedance controller to prevent a fall. While their method demonstrated a way to detect and prevent falls without a wearable device, the device itself was likely not low-cost due to the laser range finder and other sensors. In order for a walking aid to be used by the public on their own or in a rehabilitative setting, any solution would have to be low cost and as close to one of the typical devices found on the current market. Huang et al. [33] modified the prototype in [32] to try and improve their handling of fall detection by combining the data returned by a laser range finder with a camera facing down on top of the user. They could detect a fall by following the individual’s head and legs using a color tracker, and comparing the current position of the head to its position during normal walking conditions.
This tracking data was then combined with the laser range-finder data discussed in [32] in order to form a more well-educated basis for determining if a fall was about to occur or not.

Almeida et al. [34] designed a walking stick to detect falls by comparing the magnitude of the angular velocity in two directions to a predefined threshold. While the approach is rather basic, their simple design allowed for them to query the angular velocity at regular intervals, with the device in a sleep mode the other times. As the battery in any design would decrease with each measurement (and the ideal end goal for the design would be that it would last a user all day) the conservation of battery power is a high priority. If a more sensor-loaded design were able to switch between high and low power modes, its viability outside of a research environment would increase substantially.

Ojetola et al. [35] studied fall events in order to try devise a decision-tree algorithm that could be used to predict if an individual had fallen down. Participants fell in a variety of manners while wearing two sensors; one on the chest and one on the thigh. Each sensor included an IMU, microprocessor, and bluetooth device. While their classifier was able to predict falls with an 81% success rate for a seven-person training data set, the model was not in real-time. However, its low level of complexity made it a candidate for them to continue to research. One UCLA group, Wu & Au et al. developed a smart cane in order to provide an assistive device to the elderly in [20] [36]. They designed a cane to record acceleration and angular velocity usage data that could be monitored in real-time via a bluetooth sensor. The design included a piezoelectric speaker to provide audio feedback to the user upon any improper cane usage. Improper usage is determined by a classifier that uses several thresholds which are determined through manually training the device on a per user basis.
Their offset cane utilized a pressure sensor in both the base and handle, and packaged the design as a single unit that could be attached to angled portion of the shaft, making the device highly modular. Having a design that was similarly compact and as inconspicuous as possible was of utmost importance for this thesis.

Their research then was continued by Xu et al in [37], who expanded their decision-tree logic to wearable devices in an attempt to detect falls using a variety of biosignal data. An example of the biosignal data they were after is the pressure distribution that occurs in an individual’s feet as they fall. They created a “smart insole” by placing 40 force sensors and a 3D accelerometer, gyroscope, and magnetometer in a cushioned shoe insole. In addition to the insole and their smart cane, they measured individual’s electrical brain activity as they fell using a simple EEG circuit formed into a headset roughly the size of a quarter. A surprising observation was found that while most physical signals that appear before an individual falls only occur within a few tenths of a second prior to falling, some physiological signals such as an EKG or EEG can provide as much as up to three seconds of warning. This provides a strong case for analyzing an individual’s gait using multiple intelligent systems as the sooner a possible fall can be detected the higher the likelihood that the fall can be stopped, whether through cognizant decision-making or the intervention of an assistive device. An instrumented cane could provide vital data that occurs during day-to-day life, while the wearable sensors could be utilized in a research and development atmosphere in order to try and push this type of data into the device so that falls may be better prevented.
Chapter 3

System Design

This chapter first presents the design requirements, follow by a discussion of the various features of the system and the design process behind their creation. This includes the reasoning behind selecting the components that were used, the fixtures that were created to house them, and how the individual parts are connected to the rest of the design. A solid model (Computer Aided Design) representation of the final assembly is shown below in Figure 3.1.

![Figure 3.1: CAD Representation of Final Cane Assembly](image-url)
3.1 Design Requirements

The design requirements for this project were based on reviews of the previous version of the cane, and functional requirements based on tasks that this version was expected to be able to perform. While the initial prototype was functional, its main purpose was to prove that further research into such a tool was warranted. As such, it was desired to improve the mechanical design of the system and transition it from the look of a pure research project. This included trying to keep the entire system within the typical volumetric envelope of a standard offset cane as much as possible. In other words, the cane’s width, thickness, and height should be changed as little as possible as a result of adding components to the device. In this paper, the height is the distance from the base of the cane to the top of the handle. The width refers to the distance from back-end of the handle to the point where the shaft of the offset cane begins to bend, as this is the widest part of the cane. The thickness refers to the distance between the left-most and right-most points on the cane’s shaft. For a non-instrumented cane, this is the same as the shaft’s diameter. If this thickness of the cane became too large, the device would come into contact with the user’s leg as they used the device. Not following this requirement would result in a cane with a bulky look to it, and make it harder to use. For the device to have a chance of eventually being used by the public, it could not be much heavier than the typical walking canes on the market today. If the weight of the additional components became easily noticeable, users would be more likely to opt to use a non-instrumented cane. Also, any components added to the system would have to be placed in a way that they did not prevent adjusting the height of the cane shaft. Given the importance of good walking posture and the wide range of heights in
individuals, having a non-adjustable walking cane is a non-starter. It was also required that
the prototype maintain its modularity with regard to the ability to hot swap bases if a user
desired to use one other than the single-tip ferrule.

Another focus of the first version was determining if force data could be gathered to
accurately recognize a type of gait activity, such as walking normally or up a flight of stairs.
The second version of the project would be subjected to this analysis, so the device should
maintain the ability to measure force at both the base of the cane as well as the handle.
The device should be able to record force whether a petite or larger person were to use the
cane. In order to maintain all of the previous prototype’s functionality, the device would
also have to be able to house a similar electronic system in the newer prototype. In addition,
to improve the device’s usability it needed to be easily rechargeable. Since a walking device
accompanied an individual all day, the system would have to be able to last several hours on
a single charge.

While the initial cane provided functionality that is useful to researchers, it lacked features
that would be directly useful to the end-users of the devices. As fall prevention is a major
concern for individuals who utilize walking aids, it was decided to add the functionality to
detect and alert the user of potential objects that are in their current trajectory. Finally,
although cost did not have a direct effect on the current design, it was considered when
researching sensors and electronic components. This is due to the reality that even if a
device was indistinguishable from canes on the market today and had a bevy of features,
if its cost was markedly higher than a typical cane, the cost would always outweigh the
benefit. A summary of the design requirements for the design (and how they compare to a non-instrumented cane) can be seen below in Table 3.1.

<table>
<thead>
<tr>
<th>Property</th>
<th>Standard Cane</th>
<th>Prototype Requirements</th>
<th>Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height (m)</td>
<td>0.75 - 0.98</td>
<td>0.75 - 0.98</td>
<td>Range</td>
</tr>
<tr>
<td>Width (m)</td>
<td>0.18</td>
<td>0.18 - 0.21</td>
<td>Range</td>
</tr>
<tr>
<td>Thickness (m)</td>
<td>0.0254</td>
<td>0.0254 - 0.0635</td>
<td>Range</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>0.635</td>
<td>0.68</td>
<td>Max</td>
</tr>
<tr>
<td>Maximum Force Experienced (N)</td>
<td>1335</td>
<td>1335</td>
<td>Max</td>
</tr>
<tr>
<td>Minimum Distance to Detect Objects (m)</td>
<td>—</td>
<td>0.1524</td>
<td>Min</td>
</tr>
<tr>
<td>Maximum Distance to Detect Objects (m)</td>
<td>—</td>
<td>1</td>
<td>Max</td>
</tr>
<tr>
<td>Battery Charge (hr)</td>
<td>—</td>
<td>8</td>
<td>Min</td>
</tr>
</tbody>
</table>

Table 3.1: Cane Design Requirements

Before completing the overview of the design requirements it is important to understand that the goal of this project was not to improve upon the basic structure of the standard offset cane, but rather to improve upon a system design of which a standard offset cane is a part of.

### 3.2 Obstacle Detection & Notification Subsystem

In order to detect potential obstacles in front of an individual and notify the individual of said obstacle, a subsystem was designed using an ultrasonic sensor and vibration motor. The ultrasonic sensor was chosen as the sensor to detect if objects were within a certain minimum distance to the cane, while the vibration motor acted as the device that interacts with the user in order to warn them. The next two sections discuss why it was decided to use the components that we did, how the specific components were selected, as well as how the components fit into the cane design. The third section provides an overview of this subsystem including how the components are connected and how potential obstacles are detected.
3.2.1 Ultrasonic Sensor

Per the system’s design requirements, a feature needed to be added that could detect a potential obstacle in the user’s path. In order to detect objects without any computer vision techniques (which can be computationally expensive), it was determined that an emission sensor should be used because of their relative low cost. These sensors work by emitting a signal and determining how far away an object is by recording the amount of time the signal takes to return to the sensor after it bounces off of the object, as shown below in Figure 3.2. The closer an object is, the less time it will take for the signal to return. Two such devices are infrared (IR) and ultrasonic sensors, with IR sensors using light with IR wavelengths while ultrasonic sensors use sound. While IR sensors are typically cheaper, they come with some drawbacks. First, since the sensor emits light, the sensor’s accuracy is very low outdoors or even indoors if in direct or indirect sunlight. Also, light reflects differently off of different surfaces, so two objects that are the same distance away but vary in texture or color will be recorded with one object further away than the other. As such, an ultrasonic sensor was determined to be the best option for our application.
Since ultrasonic sensors use sound rather than light, they are able to detect objects with a high level of accuracy regardless of color or texture (with the exception of sound absorbing materials), in both outdoor and indoor environments. As sound travels about 343 meters per second, objects can be detected very quickly (e.g., the time for a pulse to hit an object 1 meter away and return to the sensor is just 5.83ms). However, they are more expensive than IR sensors with a typical cost of over $25. In selecting a sonar sensor, the device needed to have an operating voltage in the 2.2-3.7 V range due to the batteries used. Given the average person’s gait, it was determined that the sensor needed to pick up potential objects as far away as 1 meter, so the max range would need to be at least this high. Doing so would allow enough time for an obstacle to be detected and any implemented notification system to relay a warning to the user. The sensor that fit all of these requirements was the Maxbotix LV-MaxSonar-EZ1 (Figure 3.3).
The LV-MaxSonar-EZ1 uses a serial interface and provides readings every 50ms at an operating voltage of 2.5-5.5V. The device detects objects from around 6 inches all the way to 6.45 meters by emitting a pulse of sound at a frequency of 41kHz. Given its lower power requirements, with a current draw of only 2mA, it is ideal for a battery-based system. It also pairs well with the Arduino microcontroller used as it only requires three connections: +5V, ground, and analog out. Another pin can be used to configure the sensor so that it only takes readings at certain times, or be turned on or off completely, allowing for battery conservation at a time when obstacle avoidance is not an issue. Although the cost of the sensor is expensive relative to the cane ($26), the cost can be accepted given the cane’s current research-based focus. If the device were to try to be marketed in the future, the cost of the sensor would be driven down by purchasing in mass-quantities or by finding a cheaper ultrasonic alternative.
The compactness of the sensor was also important, as the sensor would need to be placed on the outside of the cane’s shaft. Being on the outside also required designing a fixture to house the device. The LV-MaxSonar-EZ1 was beneficial in these regards as its size is only 19.9mm X 22.1mm X 15.5mm (L x W x H - height including the PCB and sensor), and had 2 mounting holes already on the board. A simple fixture (Figure 3.4a) was 3D printed using PLA plastic (Replicator 2, Makerbot, USA) that utilized these mounting holes. The housing is split into two pieces which go around the shaft and lock into place once the sensor is attached. The sensor is secured to the housing by placing the board’s mounting holes.

Figure 3.4: Rapid Prototyped Fixture for Ultrasonic Sensor
through the housing’s two 3.1mm diameter pins. The mounting pins allow the board to be used as the housing’s connecting mechanism from the front, while two 5mm screws connect the two halves from the back. An additional 5mm screw hole was added to the top of the housing’s front to further tighten the part around the shaft. Creating a housing that could easily be repositioned was important as the ideal location for the sensor on the cane’s shaft was unknown and could vary depending on the height of the user. If the sensor is too high or too low, its location can be adjusted by loosening the connecting screws, moving to the appropriate height, and re-tightening the screws (Figure 3.4b). As shown in Figure 3.5, the housing plus sensor combination took the shaft’s width and thickness from 19.05mm and 19.05mm to 35mm to 56mm, keeping it well within the design requirements and out of the way of the user as the cane is used.

Figure 3.5: Dimensions of Ultrasonic Sensor in its 3D Printed Housing
3.2.2 Vibration Motors

While the ultrasonic sensor provided the ability to sense potential obstacles in front of the individual using the cane as they walked, any detection still needed to be relayed back to the user. As such, a decision had to be made whether to provide this information in an auditory, visual, or tactile manner. In many cases where an obstacle would need to be avoided, time is of the essence in order to provide adequate time for the user to react to any notification. With this in mind, it was determined that visual notifications would not suffice as users would be drawn more to the warning displayed on a screen than to look for the potential obstacle. Also, it would have required installing a miniature screen into the cane’s handle. While this could prove useful in relaying other, non-time-sensitive data, such as estimated distance walked or calories burned, it was out of the scope for this project and set aside for any potential future revisions. Auditory tones have the benefit of not drawing your line of sight from in front of you, however, a large portion of individuals that enlist the use of walking devices are over the age of 65. Considering nearly 1 in 3 adults between 65 and 74 suffer from hearing loss [39], installing a sensor that relied on the user hearing any warning sound could easily introduce more issues than it would solve. Due to this it was decided to go with notifying users through tactile sensations, more specifically vibrations.

Vibration notifications are also very common in today’s society, from cell phones to vehicles using such methods to get a user’s attention. In addition to not requiring modifying your line of sight to notice the notification (as in visual warnings), vibrations have the benefit of emitting a warning that can be felt by a part of your body that is in direct contact with the cane, making it much easier to detect. Vibration motors are also readily available at low-cost
in a variety of sizes and specifications. The vibration would need to be strong enough to get the user’s attention, but not too overpowering to become a distraction, similar to a cell phone on vibrate mode.

![Figure 3.6: Vibration Motor’s Location Inside the Handle of the Cane](image)

This would allow for users to feel the vibration through the plastic of the handle. However, after testing several samples it was determined that just one motor would be sufficient. As most of the user’s weight is placed at the base of the hand as they push down to support themselves, the motor was better suited to be placed in the top half of the handle where a larger portion of their grip would be. The placement of the vibration motor in the cane’s handle relative to the other electronic components is shown above in Figure 3.6 at the tip of the red arrow.
In selecting a specific motor, several constraints had to be met, all of which were satisfied by Parallax, Inc.’s Vibration DC Motor Capsule (Figure 3.7). First, the motor’s 3.0 voltage rating was within the 2.2-3.7 V operating voltage needed to be within the rating of the batteries used with the rest of the electronics system. Second, the motor needed to be easily fixed to the inside of the handle. To make this easier, a search was further reduced to motors that were fully enclosed, so it could be easily attached to the plastic of the handle. The Parallax motor is ideal for this as the motor is covered by a plastic shell. This also removed any concerns about having a moving component near other components. If the capsule were to become detached from its mount in the handle, there wouldn’t be any worry about the spinning motor damaging the internal boards. Also, the motor is fairly compact, with a length of 23mm and diameter of just 8.8mm. Initially it was thought to place two vibration motors in the handle, one in the bottom half of the handle where the fingers grip, and one in the top half where the base of the palm rests. However, with a speed rating of 12000 +/-
2500 rpm, it was determined that only one motor was needed to produce the desired amount of vibration. This was a great benefit considering the tight space requirements inside the handle.

3.2.3 Subsystem Overview

Using an ultrasonic sensor and vibration motor, our cane can detect potential obstacles in front of the user and notify them of the hazard. The microcontroller in the handle acts as a middleman between the sensor and the motor by executing the logic used to identify obstacles. As the ultrasonic sensor takes its readings, if it is decided that an obstacle is in the path of the user, a signal is sent to the vibration motor to turn on for 2 seconds to alert the user. This combination provides a low-power solution to quickly and accurately warn the user of any obstacles. Figure 3.8 depicts the connectivity of the subsystem. The power and ground connections of the sensor and motor are connected to the power and ground ports on the microcontroller. The sensor has two additional connections to the Arduino: an analog port for the microcontroller to read in the distance values of the sensor, and an analog port for the sensor to read in the sensor configuration specifying how often to take a reading. While the vibration motor is directly connected to the microcontroller, the ultrasonic sensor is connected via a USB cable, which also connects the microcontroller to other system components found in the base of the cane which are discussed in sections 3.3 and 3.5.
One issue to consider when determining when to alert the user of an obstacle is to first decide at what point it is considered that an obstacle is actually present. As the user walks around a dynamic environment, many objects will momentarily pass in front of the user, such as a person walking in front of them. In this case, you would not want to alert the user every time the sensor picks up an object in its 6.45m range, otherwise the handle would constantly vibrate. To get around this, the extensive range of the sensor was used to our advantage. As the sensor can pick up objects several steps ahead, any potential objects that are picked up but subsequently dropped within a minimum threshold can be disregarded as a passing object and not an obstacle. If an object is picked up in the user’s path and remains there for a certain number of readings, it is more likely to be an actual obstacle. The ultrasonic sensor is configured to take a reading 20 times per second. An obstacle is considered to be detected if the sensor picks up an object within 1 meter for 20 out of the past 25 readings. Simplified pseudocode for this logic is given below in Algorithm 3.1.
Algorithm 3.1 Obstacle Detection Algorithm

1: function UPDATEDISTANCES(distances)
2:    remove oldest element in distances
3:    distances.push( sensor.getReading() )
4:    if objectThreshold exceeded for x out of y readings then
5:        return true
6:    end if
7:    return false
8: end function
9: ⊳ Initialize array of past y measurements to be effectively infinity (no obstacle)
10: var distances = $+\infty \times \text{ones}(1, y)$
11: while sensorOn is TRUE do
12:    bool pulseMotor = updateDistances(distances)
13:    if pulseMotor then
14:        motor.pulse()
15:    end if
16: end while

While ensuring that the user isn’t alerted to every object that passes through the sensor’s field of view, it must also be restrained so that it doesn’t continuously alert the user to objects that are within the 1 meter range. An example of this is the user talking to an individual in front of them 0.5 meters away while not moving. In its current state, the sensor would continuously log an object at 0.5m away, but it is in fact not an obstacle. To get around this, once a potential object has been detected (per the method above), if subsequent readings remain within the 1 meter range for more than 2 seconds, it is assumed that the reading is not actually an obstacle, and no additional signals will be sent to the vibration motor. Algorithm 3.2 modifies the code in Algorithm 3.1 to handle this case.

Once the sensor records that an object is no longer within 1 meter, the user will again be alerted to any new potential obstacles. Given the placement of the ultrasonic sensor on the shaft of the cane, this also handles the case where the user is walking up a flight of stairs.
Algorithm 3.2 Modified Obstacle Detection Algorithm

```c
1: bool nonObstacle = false
2: 
3:   ▷ Initialize array of past y measurements equaling z seconds to read no obstacle
4: var distances = +∞ × ones(1, y)
5: while sensorOn is TRUE do
6:   bool pulseMotor = updateDistances(distances)
7:   if all distances < obstacleThreshold then ▷ non-obstacle in picture; ignore
8:       nonObstacle = true
9:   else if all distances > obstacleThreshold then ▷ non-obstacle out of picture
10:      nonObstacle = false
11:   end if
12: 
13:   if pulseMotor ∧ ¬nonObstacle then
14:       motor.pulse()
15:   end if
16: end while
```

As the steps will consistently be in the sensor’s line of sight, once an initial alert is given as the user approaches the staircase, the readings while they are moving up the stairs will be disregarded.

3.3 Force Measurement Subsystem

In the initial iteration of this cane design in [25], a set of Force Sensing Resistors (FSRs) were used to analyze the applied load at different areas of the cane. Several FSRs were placed on the handle to measure the force applied by the user’s grip, while a single FSR was placed in line with the shaft near the base of the cane to measure the force the user exerted downwards on the cane. For this iteration of the cane, this subsystem was improved by replacing the FSR in the shaft with a load cell. The following sections discuss the selection process of these components, as well as an overview of the subsystem’s connectivity with the rest of the cane.
3.3.1 Force Sensing Resistors

In order to measure the force exerted by the user as they gripped the cane’s handle, it was decided to use several FSRs. FSRs are robust polymer thick film sensors that work by decreasing the amount of resistance between two wires as the force applied to the surface of the sensor increases. The sensor consists of non-conductive and conductive layers separated by a thin spacing layer. When a force is exerted on the non-conductive film at the application area, this FSR material touches the conductive layer, decreasing the resistance between the wires. If no force is applied, the spacing layer ensures that the two layers do not interact at all, making it easy to tell if no force is applied.

![Interlink Electronics FSR 402](image)

Figure 3.9: Interlink Electronics FSR 402 Force Sensing Resistor

The FSR that was used for this design was Interlink Electronics’ FSR 402, and is shown in Figure 3.9. The FSR is a round sensor that is 18.28mm in diameter (active application area of 12.7mm), and a thickness of 0.55mm. The FSR 402 has a force sensitivity range
between 0.1 and 10.0 Newtons. This range is adequate as the bulk of the load experienced in the cane is located in the shaft as the individual bears their weight as they walk, and not specifically in the grip. With repeated readings stated at being within 2% of each other, by attaching several of these sensors to the cane’s handle we were able to continuously measure the overall force applied by an individual’s grip with a high degree of accuracy.

For this application, FSRs were a great choice due to their low-cost nature and very small thickness, allowing for them to be placed on the handle without altering its general contour. This was important because if the handle required modifying an individual’s grip, the likelihood of them returning to their previously used walking aids would increase. The major design decision related to force measurement in the handle revolved around how many to use. The force exerted by an individual’s grip varies at different points, so in order to return the best data, the placement of these devices was very important. By analyzing the grip of several individuals, it was determined that using 8 FSRs would provide sufficient coverage of the handle’s surface area.

Figure 3.10: Placement of Force Sensing Resistors on Cane Handle
By measuring the force in multiple locations on the handle, how an individual’s grip changes as they perform different actions can be analyzed. Based on the results of the study conducted with the initial cane iteration, it was determined that the FSR placements were indeed sufficient. As such, the locations of the original 7 were kept the same in the newer cane design, with an additional FSR placed on the bottom part of the handle. These placements are shown above in Figure 3.10. Figure 3.10a shows the FSRs on the bottom part of the handle (top of the figure) and the FSRs on the top part of the handle (bottom of the figure). Figure 3.10b shows their positions as an individual would see them looking at the assembled device.

3.3.2 Load Cell

To provide more accurate readings of the axial force experienced in the cane’s shaft, it was decided to switch from a single FSR to a compression load cell. It was decided to use a load cell in the shaft of the cane as they have long life cycles and provide a high degree of accuracy relative to FSRs. Load cells work by deforming strain gauges in a wheat stone bridge configuration as a force is applied. As the magnitude of the force increases, the strain gauge measures the deformation as a change in electrical resistance, and returns a voltage output. Since the amount of stress that the load cell experiences is far below the material’s limits, these electrical signals need to be amplified in order to use the data.

The load cell that was selected was Measurement Specialities’ FC23 Compression Load Cell [40], and is shown in Figure 3.11. This force sensor fit our application needs well due to its small size and low cost.
The FC23 encases silicon piezoresistive strain gauges in a stainless steel substrate, and comes in at just 47.23 grams. This helps keep the device’s overall weight down per the design requirements. The sensor was also selected as it provides a way to output either unamplified or amplified force data, with a sensor response time of only 1ms. This removes the need for an additional component to turn the millivolt-magnitude data into a useable form. The sensor operates with a 2.0-10.0V and 3.3V-5.25V supply voltage range in its unamplified and amplified states, respectively, so both are within the power requirements of the batteries chosen. The FC23 purchased is able to sense with 1% accuracy from 0 to 2000 lbf. While the FC23 comes in a variety of force-sensing ranges, the largest range was chosen as it was guaranteed to cover the design requirement of supporting/measuring a maximum load of 1335N (2000lbf = 907.185kg = 8896.45N).
3.3.3 Subsystem Overview

By integrating several FSRs and a load cell into the system design, we are able to accurately measure the force experienced at the two extreme ends of the device. Whereas the obstacle avoidance subsystem provides a direct benefit to the end-user, this subsystem provides a more direct benefit to a physical therapist by providing an additional tool to analyze an individual’s gait and diagnose any issues. This subsystem is also beneficial to researchers as it provides information that can be used to programatically determine what type of walking activity is occurring as the experienced load changes (such as normal walking, walking with head turned, walking up stairs, et cetera). The data from the FSRs is received and interpreted by the Analog-to-Digital Converter (ADC), which is then fed to the microcontroller. The load cell outputs a voltage, which is then converted to a load for human reading. The data is fed from the load cell to the Arduino using a USB 3.0 connector, which is located on the end of the electronics housing in the handle. In order to accommodate the adjusting of the cane’s height, a 1 meter long cable was placed in the shaft. The connectivity of the subsystem is shown below in Figure 3.12.

As the data is acquired, the microcontroller sends the data from both the FSRs and the load cell to the host machine running the executables over a wireless connection using an Addicore nRF24L01 RF transceiver. Providing a way to offload the processing of data from on the cane to a separate device drastically lowers the power requirements of the onboard system. A separate program running on the host machine then post-processes the raw data and displays force statistics to the physical therapist.
Figure 3.12: Force Measurement Subsystem Connectivity

The force exerted on the handle is calculated by converting the resistance returned by each FSR into a force, and averaging the outputs from the 7 sensors to determine the average force. The axial load experienced by the shaft of the cane is known by the voltage output of the load cell. Given the load cell’s amplified voltage range (3.3 to 5.25V), force range (0 to 2000lbf), and the fact that the output signal is linear, converting the voltage return value to a force is trivial. By utilizing the wireless transfer of data, the cane is able to be used in a wider variety of tasks in a physical therapy setting as the workspace area is confined to the large area coverage therapy setting as the workspace area is confined to the large area coverage of a wireless network compared to the small area coverage of a tethered device.

3.4 Handle

As the majority of the system’s electronics needed to be contained within the cane, it was decided to design a custom-module that could hold all of the hardware inside the handle while still maintaining the appearance of a typical offset cane handle. The handle required an especially high level of attention to detail as it is the part of the device that an individual
The handle is broken up into two components: the outer handle that the individual grips as they walk, and a unique housing that slides in and out of the cane’s shaft which is used to hold all of the electronic components. An image of the assembled handle module is shown below in Figure 3.13. In these images the handle is completely assembled, after the FSRs have been wrapped in a parafilm cover, but prior to being covered in a second, more durable protective coating. Figure 3.13a demonstrates the handle’s look with the electronics tray fully inserted, while Figure 3.13a demonstrates the tray’s ability to easily slide out. In this figure the tray has been pulled out in order to reprogram the system. Rather than having to disassemble the entire module, the tray only needs to slide out far enough to expose the pins on the microcontroller in order to connect to the Micro-USB to Serial UART breakout board (which is connected to a desktop machine) shown in the figure.

![Figure 3.13: Cane Handle](image)

(a) Cane Handle with Housing Closed  
(b) Cane Handle with Housing Open

Figure 3.13: Cane Handle
3.4.1 Electronics Housing

The electronics housing is used to hold several of the system’s major components in place. These included the microcontroller, RF transceiver, 9-DOF IMU, and female USB connector, which allows for the components in this part of the device to connect to the components in the base. The design of this component was a big focus considering the limited space that is available to the designer. Space is limited in an offset cane due to the fact that, unlike a standard walking cane, a large part of the cane’s shaft actually goes through the handle itself. This required designing a component that allowed us to utilize the space within the shaft to be able to store all of the needed components. Two concerns from this approach were ensuring that whatever insert we created could keep the components secure and not fall down the shaft, as well as preventing the insert from sliding in and out during the device’s use. Preventing component and housing motion was an especially big concern for the IMU, as any extraneous motion would affect data recorded as the device was used. The CAD file of the designed part is shown below in Figure 3.14.

This design is optimal as it divides the insert into separate sections for each hardware component, and provides a way to keep the insert secure within the shaft of the cane. The half-circle going around the back end of the insert is the part of the component that resides within the cane’s shaft. This circle-boundary is the exact diameter of the shaft’s 20.32mm inner diameter. This creates a very secure fit between the two parts, ensuring the insert will not slide in and out of the shaft as the device is used. From the back of the insert to the front, the order of the component’s placement is the USB connector, RF transceiver, IMU, and the microcontroller. The USB connector resides in the back as it is where the
main connection occurs that links the components in the base and handle together. The elevated platforms within the circle-boundary portion of the design match the width of the 15.24mm-wide transceiver and IMU. This allows the boundary that prevents the part as a whole from moving to also create a tight fit for each component, preventing them from moving side to side. Each of the components were secured to the housing using hot glue.

The microcontroller is placed in the front portion of the insert due to its width being too large to fit within the shaft of the cane. As such, this part of the insert was created wider...
than the rest of the part. However, this causes this part of the design to serve an additional purpose by preventing the insert from being pushed too far into the shaft of the cane, acting as a guide to know when the housing is fully inserted. Figure 3.15 above shows the placement of each of the above mentioned components after they were wired together. The majority of wires are signals coming from the USB to the microcontroller. The remaining connections are for powering and connecting the transceiver and IMU to the microcontroller.

3.4.2 Outer Handle

The outer handle of the cane was split into a top and bottom half so that they would be easier to work with during assembly. As with the other parts, the handle was created using a 3D printer (Replicator 2, Makerbot, USA) and PLA plastic filament. The contours of each part are consistent with that of a standard offset cane so that users would not feel any different when using the device compared to other canes. The bottom half (shown in Figure 3.16a) has a circular extruded cut along the handle where the insert of the cane rests. This allows for a seamless transition between where the shaft of the cane rests in the handle and the part of the insert that rests outside of the shaft. The back end of this half of the handle has a small open pocket for storing of the block of resistors that connect the FSRs and the microcontroller.

The vibration motor of the device is stored in a slot in the top half of the handle (Figure 3.16b) where it does not interefere with the shaft or the microcontroller underneath it. The top half of the handle was an ideal placement for the vibration motor as it is where most of an individual’s grip is placed when walking with the cane, and as such the location where they are most likely to feel the vibrating notification that occurs when near a potential obstacle.
As shown by Figure 3.13a, the insert and handle are designed such that when the housing is fully inserted, its front end is flush with the front end of the handle.

![Figure 3.16: Solid Model Showing Inside Design of Cane’s Handle](image)

The two halves are secured to the shaft of the cane by applying hot glue to the areas contacting the shaft. The two halves are attached to each other using 6 1.75mm diameter pins and pin holes at the front, center, and back of the parts (Figure 3.16). Parafilm is then wrapped around the handle, securing the eight FSRs that are located on the outside of the handle. Parafilm was used as it is keeps the FSRs in place, but is still thin enough to prevent the additional layer from lowering the accuracy of the sensors. While the parafilm keeps the FSRs in place, the solder joints where wires connect to the sensors create bulky, and sometimes sharp bumps along the handle. In addition, if the sensor is at a location where it isn’t perfectly flat (e.g., along a curve), a sharp edge can be created where the sensor is wrapper to conform to the curve. As this would cause the device to be uncomfortable to use, the handle was then sprayed with a liquid rubber. This serves two purposes by creating
a cushion-like layer so that users are not able to feel the FSRs and it further secures the handle to the shaft of the cane. An additional benefit is making the outside of the handle water resistant, preventing damage to the FSRs. The liquid rubber is also thin enough to not decrease the accuracy of the sensors.

The implemented design works well with the current application, as it allows the electronics in the handle to be accessed without having to unassemble (and effectively destroy) the rest of the handle. As shown in Figure 3.13b, while the outer part of the handle is permanently fixed to the device, the insert is able to slide in and out, allowing work to be done on the electrical system. If the designer would like to adjust the microcontroller’s programming, or test an electrical connection on a component in the handle, all that is needed is to slide the insert out where the USB cable can then be unplugged, completely separating the system’s electronics from the rest of the device. Once work is completed, the USB cable can be plugged in and housing slid back into the shaft of the cane, reconnecting the entire system with only one connection.

3.5 Base

The base of the cane is a custom module that goes between the ferrule and shaft that was created to contain the electronic components not found in the handle. While its primary purpose is to house the FC23 load cell, it also contains the batteries powering the system as well as connectors allowing communication between the sensors in the lower half of the cane and the microcontroller in the handle. The base module (as assembled with the rest of the cane) is seen in Figure 3.17 below.
In order to utilize the FC23 load cell (whose diameter is wider than the shaft of the cane) a housing needed to be constructed that could both protect it and keep it from moving as the cane was used. In addition to being low-weight, the FC23 also comes in a compact form, with the sensor’s base just 31.75mm in diameter and 10.20mm thick, with a load application diameter of 8.00mm in its center. This allows for a housing to be constructed that keeps the device secure while also keeping it from protruding too far past the cane’s 19.05mm diameter shaft. In order to return the most accurate results possible, it was important to keep the load cell as close to the ground to avoid the force dissipating as it traveled from the ground.
up the shaft. To achieve this, a housing was used that would serve as a connector to both the shaft of the cane as well as the ferrule. The housing was rapid prototyped with PLA plastic (Replicator 2, Makerbot, USA).

![Rapid Prototyped Part](image1)

(a) Rapid Prototyped Part

![Part with Ferrule and Load Cell](image2)

(b) Part with Ferrule and Load Cell

Figure 3.18: Bottom Portion of Base Module Housing

The housing is split into two halves that are secured with 3 5mm screws. The load cell rests in the bottom half while the top half acts as a lid. The bottom half has a cylindrical extrusion that can slide into a ferrule, and is shown above in Figure 3.18. For this project a single-tip 0.8 inch diameter rubber ferule was used, so the housing was fitted to these specifications. However, using a standard ferrule means that a different type of ferrule could be used if the need occurred so long as it was of a similar size. A dowel pin is inserted in
the bottom cylindrical extrusion, and is what is used to apply a force to the load cell. As an individual pushes down on the cane as they walk, the normal force of the ground pushes on the pin (inside the ferrule), which in turn applies a force directly on the load cell’s application area. Figure 3.19 below demonstrates this process. By using the housing as a connector to both the shaft and the base, the overall height of the cane is only increased by 31mm, and the sensor is able to remain just 56mm off the ground, ferrule included.

Figure 3.19: Section View of Cane Base Showing Interaction Between Load Cell and Dowel Pin as Cane is Used
As shown in Figure 3.20a above, the top half has a cylindrical tube extruding from the base that the shaft of the cane can slide into. A spring detent allows for the housing to be
secured to the shaft of the cane while also allowing for quick and easy removal should the need arise. In addition to the load cell, this part of the module also houses the system’s batteries, an IMU, and female USB 3.0 connector. As shown by the section view in Figure 3.20c, the lid has three main areas. The open compartment directly above the base of the lid is where the IMU is located, which allows for acceleration data experienced at the base of the cane to be captured in addition to the acceleration data at the handle. The IMU’s signals are transmitted to the microcontroller in the cane’s handle using a USB 3.0 cable with a female connector in both the handle and base of the cane. The second main area is the rectangular slot where the shaft of the cane comes into contact with the lid. This slot holds the base’s USB connector and is secured in place using hot glue.

Finally, in order to provide adequate battery life, slots for three 3.0V 600mAh batteries (Tenergy Li-ion RCR123A, 34mm length, 16mm diameter) were created around the shaft, although only two were needed to sufficiently power the system. As seen by the top view of the housing in Figure 3.20b, the connector on the far left provides an easy way to recharge the batteries by plugging the connector into the charger, preventing the user from needing to remove and recharge each battery separately. This view also shows the USB connector. Wires attached to the positive and negative terminals of each battery are connected to the bottom of the USB connector by being pulled through small openings in the base of each slot and into the main lid compartment before being combined into one positive and one negative signal. Wires from both the load cell and ultrasonic sensor also go through this main compartment in order to connect to the USB connector. As a layer of extra protection, a simple lid was printed to cover the exposed wires and batteries, and can be
seen above in Figure 3.17. Even with the batteries positioned around the shaft of the cane, the device’s effective diameter (the diameter of the shaft plus the added width as a result of any attachments) is only increased by 37mm, from 19.05mm to 56mm.
Chapter 4

Electronic Hardware and Software Overview

In addition to the obstacle detection and force measurement subsystems discussed earlier, it is important to review the interconnectivity of the cane device and how it can be utilized by someone such as a physical therapist. This chapter aims to achieve this by breaking the topic into two components: the electronic hardware used to log/send/receive data signals, as well as the software used to control the system and make predictions about the cane’s current state. This software was developed by Joshua Wade in the RASL at Vanderbilt University. Although the focus of this thesis is in the mechanical design of the cane, it is important to provide the reader with a high-level discussion of these components as they are an integral part to making the cane a successful device.

4.1 Hardware

The hardware needed to relay sensor data to the prediction software consists of three components not previously covered in detail. These include a microcontroller, which allows us to sample data from all of the sensors; a RF transceiver, which is used to wirelessly send/receive data to/from the host computer the software is running on; and a USB-to-serial breakout board which provides a connection to the host machine for receiving the transmitted data. Previous iterations of the cane used a Texas Instruments CC2530 microcontroller, which comes with a built-in RF module. However, due to the complexity and level of expertise needed to work with the device, only select individuals that had been working on the project for a long amount of time knew how to make non-trivial changes. As such, it was decided to switch to a different microcontroller. The Arduino Pro Mini was an
ideal replacement because of its small size and the relative ease-of-use that comes with using an Arduino board. This makes updating the device’s programming or adding future functionality much easier for individuals on or new to the project.

Since the Arduino does not have an RF module, an additional component was needed to extend the Arduino’s functionality to wirelessly forward the sensor data. An Addicore nRF24L01 transceiver was chosen for its compatibility with Arduino microcontrollers at 3.3V and on-board support components including a 2.4 GHz antenna. This eliminated the need for any additional components. The transceiver is also beneficial because of its low power consumption (11.3mA transmitting, 13.5mA receiving [2 Mbps], and 26uA standby) - vital for extending the cane’s battery life.

![Figure 4.1: Relationship Between Cane Electronics and External Transceiver](image)

Figure 4.1: Relationship Between Cane Electronics and External Transceiver

While the cane has the ability to send sensor data, an external transceiver (henceforth a dongle) is also needed on the host machine in order to receive the data being transmitted. The dongle provides two-way communication between the cane and PC so that data can
both be sent to the PC and commands can be sent to the cane. This is achieved by using a 
second Arduino Pro Mini and Addicore nRF24L01 pair with a FTDI USB-to-Serial Breakout 
Board. The two RF transceivers create the wireless connection that data from the two 
microcontrollers travels over, while the USB-to-serial breakout board creates a wired link 
between the off-board microcontroller and host machine. This relationship between the 
cane and dongle is depicted above in Figure 4.1. While it was not required to use an 
identical microcontroller in the dongle, doing so simplified the problem by allowing similar 
sending/receiving code to be used on both the on-board and off-board components.

4.2 Software

The software program implemented for use with this device, coined the Mobility Aid 
Analytics Tool (MAAT), combines several executables that work together to gather and 
display inertial data recorded by the cane. The purpose of the software is to improve a 
physical therapist’s assessment of a patient with objective data from the device rather than 
solely relying on observations. To do this, the program aims to classify user activity based 
on usage patterns using machine learning methods. The architecture of the system is broken 
into four components, and is shown below in Figure 4.2.
The first module of the system is the embedded software (stage M1), which lives on the cane and contains the hardware discussed in Section 4.1. This is the component that is directly affected by user interaction. The remaining components are programs run on an off-board Windows machine that communicates with the device over the wireless connection. The sensor data in M1 is then sent to this machine to be processed (stage M2). Depending on the mode of operation the program is running in, the processed data is then used to either train a classification model or predict the walking mode that is occurring based on the previously-trained model (stage M3). This prediction, along with general inertial data, is then presented to the physical therapist in a clean, easy to read GUI (stage M4). Finally, a main controller connects the individual software components together and handles system-level information. The following sections break down each of these components into more detail.
4.2.1 M1: Embedded Software

The embedded software is the logic that resides on the cane controlling the logging and sending of sensor data, as well as receiving/implementing system-level commands issued by the software on the host-machine. The data recorded by the FSRs and load cell are acquired and stored in a 40-byte payload buffer by the microcontroller. The payload includes readings from each of the sensors along with a progressive package indicator, a time stamp, the battery level, and two synchronization start/stop bytes. Factoring in wireless communication, the average sampling rate of the payload by the M2 acquisition module is around 60 Hz. The sensors are queried whenever a request from the dongle is received. Once all readings are stored, the microcontroller then transmits the data package to the dongle over a 2.4 GHz frequency using the Addicore transceiver. This module is connected to the M2 module shown in Figure 4.2, which acquires the data discussed here and forwards it to the computer.

4.2.2 M2: PC-Side Acquisition

The role of this stage of the MAAT is to obtain and parse data sent from the cane to the dongle and forward said data to the processing module. Data is acquired from the dongle via serial communication (UART). A benefit of this architecture is that as the data is sent via serial communication, the behavior of this module does not depend on the type of microcontroller used. This allowed for a seamless transition between the previous iteration’s Texas Instruments device and the Arduino Pro Mini. The data received is extracted from the buffer based on the size of the data type (e.g., single precision is 4 bytes). This subsystem interacts with the embedded system on the cane by both receiving sensor data and issuing system commands such as initializing and shutting down monitoring or data-transfer
processes. The process of receiving and parsing data from the dongle is a relatively trivial task, making the primary objective of this part of the program not receiving/parsing the data, but doing so as quickly as possible. As such, this process was separated from the other host-machine tasks so that it could be implemented in C++ to maximize speed. After parsing, the data is forwarded to the M3 module for processing and analyzation via UDP socket.

4.2.3 **M3: Processing, Training, and Prediction**

![Figure 4.3: M3 Module Architecture](image)

Once the data is received from the M2 acquisition module, machine-learning methods are used to predict user activity based on the sensor signals. Due to the high-level of complexity, this module was implemented in C#. Depending on the user’s objective, the module has modes of operation for both model training and activity prediction, of which both methods are depicted in Figure 4.3. After pre-processing, the data is stored in a moving window of roughly 240 samples (around 4 seconds) with 50% overlap. While the window is configurable, it was set to 240 samples to try and maximize both the number of training samples and size.
of the training window [25]. The data is pre-processed to eliminate high noise frequency and DC baseline wander component by applying a low-pass filter with a 4Hz cutoff frequency and a 0.3Hz cutoff frequency high-pass filter, respectively. The window allows for the signal to be tapered prior to inputting it into a Fast-Fourier Transform (FFT).

<table>
<thead>
<tr>
<th>Time Domain</th>
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<tbody>
<tr>
<td>Mean</td>
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<tr>
<td>Standard Deviation</td>
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<tr>
<td>Skewness</td>
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<td>Kurtosis</td>
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<td>Quantization Bins (10 bins) [41]</td>
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<tr>
<td>Correlation Coefficient</td>
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<td>Mean Crossing Rate</td>
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<th>Frequency Domain</th>
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<td>Spectral Energy</td>
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<tr>
<td>Spectral Flux</td>
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Table 4.1: Categories of Features Extracted from Data

In order to improve the accuracy of each classifier, features are extracted in both time and frequency domains (listed in Table 4.1). A FFT was used to extract spectral energy and spectral flux features. The total number of extracted features totaled over 800, so the best 12 features were selected to generate a model that would be used for gait recognition. Four different classifiers were trained using WEKA (www.cs.waikato.ac.nz/ml/weka/). These classifiers were C4.5 decision tree, Artificial Neural Network (ANN), Support Vector Machine (SVM) with a radial basis function kernel and Naive Bayes. A WEKA .dll file provided the WEKA API calls needed for classification. As features are extracted, a prediction is made for each of the classifiers when the program is run in prediction mode. Predictions along with other metadata are logged, with the predictions then being forwarded to the M4 GUI module via TCP socket.
4.2.3.1 Module Design

The M3 module also acts as the bridge tying the other modules together. It has four major software components: a main controller, acquisition module (different from the M2 module), publication module, and a Moving Window Manager. In addition, there are several smaller, minor files that interact with the larger components. These include classes for feature vectors and classifiers, as well as general utility functions, among others. Each of the major components are modeled as a Finite State Machine (FSM) that run in parallel with the other components.

The main controller acts as a supervisory object that handles system configuration (e.g., is the program training or predicting) and monitors/reports system-level errors. The controller has 3 distinct modes: wait for configuration file specification, wait for the acquisition module to connect to the UDP server, and normal execution until user termination. Figure 4.4 displays a diagram depicting the FSM for the main controller.

![Finite State Machine for M3 Main Controller](image)

Figure 4.4: Finite State Machine for M3 Main Controller
The acquisition module manages the client-side UDP socket, and handles all network communications with the M2 module. As it receives information from M2 it forwards data to the Moving Window Manager for feature extraction and prediction. This component is also modeled as a FSM with 3 discrete states. Either the module is waiting for a configuration file to be specified, connecting to the UDP server, or experiencing normal communication with the UDP server. Figure 4.5 depicts the acquisition component’s FSM.

![Finite State Machine for M3 Acquisition Component](image)

Figure 4.5: Finite State Machine for M3 Acquisition Component

The publication module forwards activity predictions to the relevant GUI (module M4), and is also modeled as a 3-state FSM. The module is either in a sleep-state waiting for the user to initiate the application, connecting to the TCP client, or experiencing normal communication with the TCP client. Figure 4.6 depicts the FSM that the publication component is based on.
The Moving Window Manager is the portion of the program that implements the processing and prediction steps discussed earlier. As mentioned earlier its role consists of maintaining the moving data window, processing input signals through high/low-pass filters, extracting features, and predicting the cane’s current activity. This component consists of a 2-stage discrete FSM with the system either waiting for the user to initiate the program or accepting feature vectors and adding them to the moving window. Figure 4.7 shows the FSM that the Moving Window Manager is built on.

Figure 4.6: Finite State Machine for M3 Publication Component

Figure 4.7: Finite State Machine for M3 Moving Window Manager
Finally, a simple hierarchical state machine (HSM) is implemented to handle the updating of each component’s FSM. If the main controller reports that a termination event has occurred, the application closes. The source code definition for this HSM is shown below in Figure 4.8.

```
static void Main(string[] args)
{
    //initialize the data members
    terminate = false;
    mainController = new MainController();
    accModule = new AcquisitionModule();
    pubModule = new PublicationModule();

    //begin application
    while (!terminate)
    {
        //update the state machines
        terminate = mainController.UpdateFSM();
        accModule.UpdateFSM();
        pubModule.UpdateFSM();
        MovingWindowManager.UpdateFSM();
    }
}
```

Figure 4.8: Source Code Definition for Hierarchical State Machine Used to Update M3 Finite State Machines

4.2.4 M4: Graphical User Interface

The role of the program’s GUI is to receive and display predictions from the M3 processing module in a user interface for use by physical therapists. This module is only utilized when the program is run in the prediction mode, with the application run solely through a command-line tool when training the model. The program consists of two GUI’s: one for displaying the predicted activity, and one for general sensor data monitoring. The simple GUI was developed using the Unity 3D game development platform (Figure 4.9). It displays two predictions of the current state of the cane (walking, standing, or resting): one from a random decision forest classifier, and another from a C4.5 classifier.
The second GUI (shown below in Figure 4.10) is a Windows Form Application written in C# for displaying system/sensor monitoring and patient metadata. For system monitoring, the dongle’s connection status is stored along with the wireless connection’s metadata. This
includes the data chunks received from the Arduino since system initialization. Input fields are provided for patient and experiment metadata such as patient identification and the current, known DGI task the user is doing. A total of 27 signals are logged as well to provide up-to-date sensor information. This includes information from the 9-DOF IMUs in the base and the handle, as well as the 8 FSR outputs and current reading from the load cell. Currently, this is a separate executable than the M4 module’s Unity3D GUI, with the intention of merging the two applications into one concise GUI in the near future.
5.1 Force Measurement Subsystem Validation

Validating the components making up the force measurement subsystem was vital to ensure that the device was registering the correct applied force as it was used. Knowing that the device is capturing the correct weight is important if a physical therapist is to use the measurements as a basis for treatment or device recommendation. In order to validate this subsystem two independent tests were performed to validate the load cell and FSRs for accuracy.

5.1.1 Load Cell

To validate the load cell we needed to determine that the raw values output by the sensor corresponded to the actual weight applied as the device was used. Such a task is more involved than simply testing the load cell as a separate component due to how the dowel pin in the cane’s ferrule interacts with the sensor. The load cell data is affected by force applied directly to the application area of the sensor. While the dowel pin contacts this application area, the reaction force at the other end of the dowel pin is not between the ground and the dowel pin, but rather the ground and the rubber ferrule, distributing the force across the base. As such, the reported force by the load cell will not match the actual weight applied. By applying known weights to the cane and logging the corresponding outputs, a best-fit approximation can be calculated. The approximation can then be put into the data-logging program to output the corrected value.
Before generating the load approximation curve the data returned by the load cell, which is in millivolts, needed to be converted to Newtons. The load cell tends to have an initial bias that differs from sensor to sensor and may drift according to the (amplified) zero force output. Also, an initial offset needed to be taken into account due to the initial load applied with no extra added weight. The formula for computing the output in Newtons is

$$F_{out} = m \times F_{in} + b$$ (5.1)

where $F_{out}$ is the output force, $m$ is the sensitivity scalar, $F_{in}$ is the supplied force, and $b$ is the specific sensor’s bias offset. The units for $F_{out}$, $m$, $F_{in}$, and $b$ are V, V/N, N, and V, respectively. Since we are interested in the mass applied ($F_{in}$), the equation is rewritten as

$$F_{in} = (F_{out} - b)/m$$ (5.2)

To compute the values to feed to the formula, several parameters that depend on the power supply needed to be computed. The datasheet gave these parameters with respect to a 5V supply, so the values needed to be converted to the 3V supply that was used for the cane. The values could be easily computed as the parameters were listed as ratiometric to the supply. The voltage Span became $[0V, 2.4V]$ because the ratio of supply-to-Span is 0.8 (i.e., $3V \times 0.8 = 2.4V$). $m$ is computed as the Span divided by the sensor’s range, therefore $m = 2.4V/2500N$ (or 0.00096V/N). $b$ is computed based on the initial bias, which for our case was about $113.5 \times 2.4V/1024$ (or 0.266015625V). $F_{out}$ is computed as the raw value returned by the sensor in millivolts, $F_{raw}$, (e.g., 114, 115, 134, etc.) multiplied by the scalar $2.4V/1024$. 

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Acting under the assumption that the initial bias does not change much as the device is used, the equations to calculate the applied force in Newtons becomes

\[
F_{\text{out}} = F_{\text{raw}} \times \left(\frac{2.4}{1024}\right)
\]  

(5.3)

\[
F_{\text{in}} = \frac{(F_{\text{out}} - 0.266015625)}{0.00096}
\]  

(5.4)

A set of 9 known weights were then applied to the cane for 10 seconds each. To reduce the effect of noise, each returned reading is added to a moving window of size 10. The median of the window is then taken and used for prediction. Following the test, the median of the entire test data set is then taken and input to the equations above to convert the millivolt output to Newtons. As the test weights used were in pounds, the Newton output was divided by 4.448 to convert the test data to pounds. The test weights used to generate the data set were in the range of \([0.0, 55.0]\) pounds. A test range of 0 to 55.0 pounds was deemed sufficient as the cane was designed to support a maximum of 1335 Newtons, or roughly 300 pounds. When a cane is used, only 15 percent of an individual’s weight should be beared by the device [42]. Taking this into consideration, the 55.0 pound maximum covers the expected use cases \((15\% \times 300 = 45)\). The results of these tests can be seen below in Table 5.1, and reflect the expected result of the load cell interpreting loads as less than the ground truth due to the weight being distributed across the base rather than completely on the load cell.

Figure 5.1 below plots this data, showing a somewhat linear relationship between the applied and output load, which is the expected behavior of the load cell. However, there are certain areas within the range that do not keep with the linear trend. To return the most accurate predictions for weights outside of the testing data set, a third degree polynomial fit
was used. A first or second order fit produced a noticeable margin of error at certain areas within the range, while the third order fit allowed the curve to more closely follow the test data.

<table>
<thead>
<tr>
<th>Applied Weight (Lbs)</th>
<th>Output (N)</th>
<th>Output (Lbs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.5</td>
<td>8.5449</td>
<td>1.9210</td>
</tr>
<tr>
<td>5.0</td>
<td>13.4277</td>
<td>3.0187</td>
</tr>
<tr>
<td>7.5</td>
<td>20.7519</td>
<td>4.6652</td>
</tr>
<tr>
<td>10.0</td>
<td>25.6347</td>
<td>5.7629</td>
</tr>
<tr>
<td>20.0</td>
<td>54.9316</td>
<td>12.3491</td>
</tr>
<tr>
<td>30.0</td>
<td>84.2285</td>
<td>18.9353</td>
</tr>
<tr>
<td>40.0</td>
<td>125.7324</td>
<td>28.2658</td>
</tr>
<tr>
<td>50.0</td>
<td>159.9121</td>
<td>35.9497</td>
</tr>
<tr>
<td>55.0</td>
<td>178.9638</td>
<td>40.0661</td>
</tr>
</tbody>
</table>

Table 5.1: Load Cell Known Weight Application Test Data

Figure 5.1: Third-Order Approximation Relating Actual Applied Load to Load Cell Interpreted Load
While the approximation above demonstrates that a relationship exists between the applied load and what the load cell records, the function needs to be applied to data points outside of the data set used above to verify it. To accomplish this two tests were conducted with the same procedure above using two weights not previously used within the testing range. However, rather than output what the load cell is interpreting as the force, the output is fed to the third-order approximation formula to return the predicted actual result. If these predicted values were close enough to the actual weight applied, this would verify that the approximation’s accuracy, validating its use for our application. Table 5.2 below shows the results of these tests.

<table>
<thead>
<tr>
<th>Applied Weight (Lbs)</th>
<th>Predicted Output (Lbs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>15.0</td>
<td>15.39061</td>
</tr>
<tr>
<td>35.0</td>
<td>35.11664</td>
</tr>
</tbody>
</table>

Table 5.2: Load Cell Third-Order Approximation Accuracy Test Results

The test of 15 pounds predicted 15.39061 pounds, or an error of 2.604%. The 35 pound test predicted 35.11664 pounds, or an error of 0.33%. Some error is expected due to the resolution of roughly 0.54 pounds for the load cell itself, however both test cases produce results within the sensor’s resolution. This validates the output returned by the load cell, allowing the third-order approximation to be used in computing the force that is applied by an individual as they use the device. To further improve the accuracy of the approximation, the raw $F_{out}$ values from these tests were fed back into the curve fitting calculation to refine the polynomial coefficients.
5.1.2 Force Sensing Resistor Validation

In order to ensure the metadata describing how an individual was gripping the handle could be used in a clinical setting, the FSRs used in the design needed to be validated for accuracy. Once validated, the force data logged as an individual walked with the device and changed movements could be trusted by physical therapists. To simplify the problem it was assumed that each of the sensors were similarly manufactured, and as such the performance of one of the FSRs could validate the others. The FSR that was chosen to be tested was on the top half of the handle, closest to the pinky finger-side of the palm as it had the flattest surface and would be easiest to test given its placement on the handle.

Similarly to the load cell validation, a set of known weights were placed on top of handle. However, to validate the FSR it needed to be checked that no part of the weight came in contact with the handle so that the entirety of the weight was picked up by the small application area of the FSR. To accomplish this a 6.35mm, 0.5098g nut was placed on the FSR’s application area, after which the weight was placed on top of the nut. To reduce the amount of noise in the sensor, the logging program was run for 30 seconds after placing the weight into position, with the average of the sample then taken, factoring in the weight of the nut. This test was repeated 9 times for a series of gradually increasing weights. The goal of this sensor validation experiment was to generate the force-to-voltage curve created by measuring the voltage output by the FSR at each known weight data point. This curve could then be compared to the curve in the FSR’s datasheet (shown below in Figure 5.2) to confirm its accuracy. As discussed earlier, this voltage output is dependent upon the resistor
that is placed between the sensor and the microcontroller. As the resistance rating of the connecting resistor decreases, the outputted voltage will also decrease.

![Interlink Electronics FSR402 Force-to-Voltage Curve](image)

Figure 5.2: Interlink Electronics FSR402 Force-to-Voltage Curve[43]

While the figure above does not have a data series for a 1kΩ resistor, the general curve pattern for the 3kΩ curve can be used as a baseline to compare the validation data against. Although the curve is not linear, we would expect the output voltage for the 1kΩ curve to be somewhere around 1/3 of the voltage values for the 3kΩ curve. As shown above, as the resistor rating decreases, the curve becomes more linear from the start, with less of an initial increase in Voltage. Figure 5.3 below displays the generated force-to-voltage curve for the FSRs on the handle of the cane given the 1kΩ connecting resistor.
Figure 5.3: FSR Force-to-Voltage Experimental Data Curve for 1kΩ Resistor

Although the data does not follow a perfectly smooth curve, the general form of the 3kΩ curve in Figure 5.2 is followed. By matching similar x-value data points along the two curves, the data can be compared to see if the 1kΩ resistor produced results roughly 1/3 of the 3kΩ case. These similar data points are 100g, 200g, 500g, and 700g. For the 3kΩ case, the output voltages for these weights are (roughly) 0.6V, 1.0V, 1.5V, and 1.75V, respectively. For the 1kΩ case, the corresponding output voltages are 0.1030V, 0.2657V, 0.4697V, and 0.7411V, respectively. To better compare the two curves, the 3kΩ curve was added to the 1kΩ curve, which was then scaled to match the dimensions of the plot in the datasheet. Once scaled, it becomes clear that the FSR is returning the expected output voltage. This plot comparison can be seen below in Figure 5.4.
5.2 Obstacle Detection Subsystem Validation

The algorithms developed for detecting and alerting individuals of potential obstacles needed to be verified to ensure obstacles were indeed alerted to the user and non-obstacles were not. As discussed in Section 3.2, the developed algorithm was set up to detect obstacles within 1 meter in front of the cane. After consecutive readings detected an object for 1 second, a signal was sent to the microcontroller to activate the vibration motor for 1 second. If the object remained in view after the initial vibration, it is assumed that the detected object is not actually an obstacle and future vibrations are disabled until the object goes out of range and another obstacle is detected.

To validate the ultrasonic sensor and vibration motor were working together correctly, two tests were run while the status of these two components were logged. There were three possible outcomes of interest during both tests:

- **0** - Ultrasonic sensor does not detect an object and the vibration motor is off.
- **2** - Ultrasonic sensor detects an object, but the vibration motor is off.
- **3** - Ultrasonic sensor detects an object and the vibration motor is on.
As the activation of the vibration motor depends on the ultrasonic sensor detecting an object, the case where the motor is on but no detection occurs is not possible. Since the tests do not depend on how far away a specific obstacle is, testing the subsystem simply requires three motions: walking in front of the cane outside of the 1 meter range, walking in front of the cane within the 1 meter range, and standing in front of the cane for more than 2 seconds. Since the time each motion occurs is known, the output can be directly compared against the motion conducted to check the subsystem’s accuracy. The first experiment tests the first two motions, while the second tests the third. Each experiment logged system readings for thirty seconds.

Figure 5.5: Obstacle Detection Subsystem Validation: Experiment 1
Figure 5.5 included 4 completed motions. The first motion occurred at a time of 3.1s, however the output here is zero as the motion occurred in front of the device, but outside of the 1m range. In the next two motions, an individual walked in front of the device at a time of 8.4s and 23.5s, respectively. In the first case the obstacle is only present for 0.6s. Although overall the second motion is picked up for 1.2s, for several readings in the middle no obstacle is detected; thus no pulse signal is sent. The third motion saw an individual walk in front of the cane within the 1m range and slow enough to be detect for more than 1 second. After 1 second it is shown that the vibration motor was activated for another second, after which no obstacle was detected resulting in an output of 0.

Figure 5.6: Obstacle Detection Subsystem Validation: Experiment 2
Figure 5.6 plots the resulting system output for a similarly set up experiment to test that the user was not alerted to a potential obstacle if the object was within the detection range for more than 2 seconds. This is important so that the device does not constantly vibrate while an individual is walking up a flight of stairs or talking to someone who happens to be standing in front of them. The first motion saw an individual do just this - they stood in front of the device from 5.5s to 15.5s. After the 1 second threshold the vibration motor is activated. Subsequent readings detect the obstacle but do not trigger a vibration. The second motion saw an object pass in front of the cane outside of the 1m range at 22.25s, while the third motion saw an object pass within the range at 28.2s. After 1 second, the vibration motor is activated, demonstrating how the detection algorithm resets after the constantly detected object (which temporarily disabled the vibration motor) is no longer detected. These two experiments demonstrate that the algorithm controlling the two components is working correctly and validates the subsystem.

5.3 Inertial Measurement Unit

The IMU in the handle of the cane is an important component to validate as it influences the classification model when predicting a user’s current gait activity. The IMU captures the acceleration the device is being subjected to in the X, Y, and Z directions. In a steady state, if one axis is pointed directly up, two of the axes should measure 0g while the third will measure 1g, to reflect the acceleration due to gravity. To validate the sensor it needed to be shown that when the positive portion of the sensor’s coordinate system X-axis is vertically aligned and pointed up, the output for the X signal equals 1. When the positive portion of
the X-axis flips 180 degrees and is pointed straight down, the X signal’s output should be negative 1. Similar tests are also conducted for the sensor’s Y and Z axes.

Performing such a test is relatively simple given the fact the IMU’s location within the cane is fixed and the orientation of its coordinate system is known. The sensor is fixed to the tray insert in the cane handle with its Z-axis pointing directly up. The X-axis is pointed straight ahead when the device is used properly; directly in front of the user’s walking path. Following the right-hand rule, the sensor’s Y-axis is pointed to the left of the device. Figure 5.7 depicts this coordinate system.

![Coordinate System of Handle’s Accelerometer](image)

Figure 5.7: Coordinate System of Handle’s Accelerometer

To confirm the accuracy of the IMU six tests were run, testing both the positive and negative portions of the X, Y, and Z axes. As the software program already output the data for each of the three axes, the cane simply needed to be positioned so that each end of each axis was positioned vertically up. Once in each of the six positions, the data logging program was started and run for twenty seconds without moving the cane to keep the device...
in a steady state and reduce the effect of noise. Positioning the device for each test such that the axis of interest was perfectly vertical was not realistic due to the device’s non-flat and non-uniform shape. For this reason, the tests were run to demonstrate when one axis is primarily vertical, one component of the signal should be closer to 1 while the other two components were closer to 0. To analyze the data returned by the IMU, the unsigned integer outputs are multiplied by a scalar of $4/32768$ to convert the data to a multiple of $g$ (acceleration due to gravity). The 4 is a scaling factor to limit the measurement range to $\pm 4g$, while 32768 is the resolution of the sensor.

<table>
<thead>
<tr>
<th>Test</th>
<th>Average X (g)</th>
<th>Average Y (g)</th>
<th>Average Z (g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Positive X-Axis Up</td>
<td>0.9144</td>
<td>0.0291</td>
<td>0.0171</td>
</tr>
<tr>
<td>Negative X-Axis Up</td>
<td>-1.1030</td>
<td>0.0327</td>
<td>-0.0910</td>
</tr>
<tr>
<td>Positive Y-Axis Up</td>
<td>0.0931</td>
<td>1.0175</td>
<td>0.0229</td>
</tr>
<tr>
<td>Negative Y-Axis Up</td>
<td>-0.2280</td>
<td>-0.9751</td>
<td>-0.0809</td>
</tr>
<tr>
<td>Positive Z-Axis Up</td>
<td>-0.1517</td>
<td>-0.0150</td>
<td>0.9473</td>
</tr>
<tr>
<td>Negative Z-Axis Up</td>
<td>-0.0503</td>
<td>0.0541</td>
<td>-1.0318</td>
</tr>
</tbody>
</table>

Table 5.3: IMU Accelerometer Validation Results

Table 5.3 shows each test’s average output values for the accelerometer’s X, Y, and Z components. The data in the X-direction has the largest error, however this is expected as the cane’s natural resting position on its side can not be perfectly flat due to its shape. For the other tests, the component’s that were expected to be closer to 1g are indeed larger than the other components, which are closer to 0g. This validates the accuracy of the IMU, and concludes the sensor validation experiments.
5.4 System Case Study

With each of the major sensor components individually validated, a simple case study was put together to analyze the data output by the entire system as it would be used in an actual clinical setting. The case study involved three subjects (1 female, 2 male) walking normally with the device in a straight line on laminate and carpet flooring, as well as walking up and down one flight of stairs. While all tests aimed to see the relationships between the different sensor data, the laminate/carpet flooring tests also aimed to see how data varied across different surfaces, with the stair tests used to examine the detection algorithm’s performance on a flight of stairs.

For the laminate and carpet flooring tests, each subject was asked to walk in a straight line for several steps, with a turn in the middle for the carpet tests due to space restrictions. Figure 5.8 displays the corrected load cell output in pounds for each subject’s two trials. The step each subject takes can be clearly seen with each peak, with anticipated zero-load outputs in between, corresponding to the individual bearing weight on the other leg and the cane part of either the back or forward swing. As expected, the output was much larger for the two male subjects (Subjects 2 and 3) compared to the female subject (Subject 1). There was not however much difference between the output when comparing each patient’s laminate trial to their carpet trial. Most likely this has to do with the fact that the carpet which the trials were executed on was fairly thin and on top of an already hard surface. Since the carpet does slightly decrease the applied force, it would be expected that this trend would follow if tested on progressively softer surfaces such as thicker carpet or grass.
Further confirming the expected behavior of the system is shown when comparing the frequency of the load cell output with the magnitude of the base IMU’s inertial components in Figure 5.9. It is expected that the IMU in the base of the cane would register very little movement as the load cell registers an applied force, as this is when the individual is pushing down to support themselves. It then follows that the magnitude of the IMU should vary
Figure 5.9: Time v. Base IMU Magnitude and Load Cell Output On Laminate and Carpet Flooring for Subjects 2 and 3
in between steps when the load cell does not register a force (i.e., during the back-swing or follow-through).

Figure 5.9 charts these two signals for the laminate and carpet trials for Subjects 2 and 3. For each trial it can be seen that as the load cell registers a force, the IMU’s magnitude very close to constant, while in between steps the IMU’s signal varies. The sensor responses for Subject 3 followed a similar trend, showing that the sensor data from the different base components does indeed match up with expectations.

The device’s gyroscope data can also be analyzed to see how rotation about the X, Y, and Z axes for the handle’s IMU varies across subject and walking surface. The results are plotted in Figure 5.10. As expected, rotation about the X-axis (swinging side-to-side or perpendicular to forward motion) is relatively constant throughout subject and surface. Consistent spikes in the X-axis data would likely correspond to improper cane usage and could be used to help physical therapists detect and prevent these occurrences through educating the user. Rotation about the Y-axis (e.g., parallel to forward motion) is also logically the direction with the largest swings in rotation, corresponding to either the forward or back-swing of the individual’s gait. Somewhat surprisingly, however, is the variance in the Z-axis (e.g., rotation about the shaft of the cane’s center-axis). Disregarding the Z-axis spike for each subject’s carpet trial (which corresponds to the individual turning 180 degrees), one would expect this rotation to be fairly minimal considering the task of walking forward. However, it’s variance is closer to the variance of the Y-axis than the X-axis.

As the gyroscope data is consistent for each patient on both surfaces and the Z-axis relationship is seen in each trial, it is likely this is a result of two side effects. First, the
Figure 5.10: Time v. Gyroscope Output On Laminate and Carpet Flooring for All Subjects
individual’s used were healthy individuals that did not need to rely on the device in order to walk, thus likely somewhat skewing the data. However, more importantly, is that none of the subjects were properly educated on how to use a walking cane prior to the case study. It can be seen from the charts that as the subjects push off the cane onto their non-cane-using leg and the cane swings back (i.e., GyroY’s signal goes negative), GyroZ’s signal goes positive. Based on this inversely proportional relationship and knowing the orientation of the IMU, it is known that the positive GyroZ signal corresponds to subject turning/twisting their wrist inwards as they push off. None of the subjects reported remembering twisting their wrists to the left (i.e., towards their body, as each subject used the device in their right hand) as they pushed off, displaying an excellent use case for this device. The motions were quite small, yet as this motion corresponds to improper device usage, could have negative effects down the road were the behavior to continue long term. By looking at this data, a physical therapist would be able to quickly pick up on this trend and educate the subjects on how to properly walk with the cane, keeping the wrist as stationary as possible.

By asking each subject to walk up and down a flight of stairs, sensor data can be analyzed to see how the system is used in two common day-to-day movements. Figure 5.11 visualizes the response of the load cell for each subject as they completed both stair-related trials without holding on to the stairwell railing, with the aim of seeing if weight-bearing needs of the subject changes depending on the activity. With the exception of Subject 1, the two activities did not result in a noticeable difference for applied force, with the first most likely a result of improper usage due to the extreme difference. Similar behavior between the two activities is the logical expectation as the device is required of the individual to support
In addition to checking the output of the load cell, walking up a flight of stairs allows for the obstacle detection subsystem to be further tested. While the algorithm was previously validated against objects passing in front of the cane, the system had not yet been subjected to the stair test. Each subject started with the device pointed away from the stairs, walked up the flight of stairs, followed by turning towards the second flight of stairs to complete themselves, regardless of the direction they are moving if they do not utilize the stairway railing.

Figure 5.11: Time v. Load Cell Output Going Up and Down a Flight of Stairs for All Subjects
Figure 5.12: Time v. Obstacle Detection Algorithm Output for Each Subject Walking Up Stairs

The motion. This process allowed for the system to start without detecting an object and resetting after completing the movement. As such, the expected output is that the vibration motor would be activated twice for each subject: once for the initial step, and once after they have reached the last step and turned to face the first stair of the second set. The output of the algorithm for each subject is plotted in Figure 5.12, with a vibration represented by an output value of 3.

While Subject 1 and 3 displayed the expected behavior, Subject 2 did not. After completing the motion, the Subject did not turn towards the last stairs, so only 1 vibration at the beginning was expected. However, the test actually produced three vibrations. To determine the cause of this behavior, the other sensors on the device are looked at for Subject 2 and 3 during this motion (Subject 1 was not subjected to further analysis as it portrayed the expected behavior). In order to get some context as to when the vibration
motor was activated for each subject, the output of the algorithm was overlayed with the distance measurement output of the ultrasonic sensor. This measurement is the distance (in inches) detected between the ultrasonic sensor and the closest object to it. Figures 5.13a and 5.13b plot this data, along with the 36 inch threshold at which a detection is considered an obstacle.

Figure 5.13b displays the expected behavior, with a notification sent after sensor readings have been under the 36 inch threshold for 1 second, and again at the end of the motion. However, after picking up the initial vibration, Figure 5.13a shows that the motor is activated twice more while under the imposed detection threshold, despite the algorithm not outputting a zero prior, further demonstrating an unexpected behavior during the second subject’s trial.

The cause of this issue is discovered when the raw ultrasonic sensor distance is examined against the detection output for the subject’s trial. Figure 5.13 had been initially used with a low-pass filter to produce more readable results. However, this filter masked an issue with the wiring of the system that was created during a device re-assembly between the component validation experiment and case study. The rewiring had created a short circuit, resulting in intermittent readings registering infinity. This infinite value then resets the algorithm to trigger the vibration motor activation for future obstacles. Figure 5.14 overlays the detection output with the unfiltered ultrasonic distance measurement, with the viewport zoomed in to better highlight the two vibration areas of interest. In both cases where the motor is activated, an infinity reading can be seen within 1 second of the vibration. With the faulty wiring causing random shorts to occur, it happened by chance that the other trials were not similarly affected, and thus explains the false-positives during Subject 2’s trial.
Figure 5.13: Time v. Detection Algorithm Output with Ultrasonic Distance Measurement for Subjects 2 and 3

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Although the false-positives were discovered to be the cause of faulty wiring, the investigation into this issue exposed other possible sources of error that should be addressed in future iterations. The main issue with the algorithm’s current implementation is that it assumes the device will maintain a forward-facing orientation while moving up the flight of stairs, however this is not the case. While the previous tests showed that the device is not rotated about the Y-axis (along the direction of motion) enough to have the ultrasonic’s field of view to go above the stairs in front of it, the algorithm does not take into consideration that the handle may twist as they user moves up or down the stairs. Although this was not the cause of error in this trial, it could be an issue in later trials.

Figure 5.15 plots the gyroscope data of Subject 2’s trials as they walked up and down the stairs. The gyroscope data tells the story of the angular velocity in degrees per second.
Subject 2: Time v. Gyroscope Component Outputs Walking Up Stairs

(a) Walking Up Stairs

(b) Walking Down Stairs

Figure 5.15: Time v. Gyroscope Component Outputs Walking Up and Down a Flight of Stairs for Subject 2
felt in each of the X, Y, and Z directions by the handle of cane. Taking the different scales into consideration, it is seen that the data is fairly similar in the X and Y directions, which is expected. However, the data in the Z-axis is generally more positive while walking up the stairs and more negative while walking down. Given the orientation of the IMU in the handle, this tells you that the subject typically turns the cane slightly towards them while pushing off the device as they walk up the stairs, and turns the cane slightly away from them as they walk down the stairs, respectively. While this most likely corresponds to incorrect device usage (as discussed during the laminate and carpet flooring trials) this also exposes the area of trouble for the detection algorithm. This problem scenario is visualized in Figure 5.16 by overlaying the Z-component of the gyroscope data with the detection algorithm output for Subject 2’s trial up the stairs.

![Figure 5.16: Time v. Gyroscope Z-Component with Detection Algorithm Output Walking Up Stairs for Subject 2](image)

Figure 5.16: Time v. Gyroscope Z-Component with Detection Algorithm Output Walking Up Stairs for Subject 2
As the individual twists, the ultrasonic sensor turns away from the stairs in front of it, towards the sides of the stairwell. While most stairwells have railings or walls along the side, the algorithm is set up such that one reading outside of the detection threshold of 1 meter will reset the device. Were the user to happen to twist the handle such that the sensor took a reading in between the stairway’s railings, it would return a value outside of the threshold. This would cause a subsequent vibration activation signal to be sent as the device was turned back to face the stairs. Although not the cause of the false-positive in this case, it can be seen that just prior to the first false-positive the subject twisted the handle at a much higher rate than between their other steps. Had the device been turned slightly more, the device has the opportunity to register a distance reading beyond the detection threshold. This type of use, while improper and incorrect, should be handled by the system to prevent the user from being unnecessarily notified. Fixing this is as simple as requiring a set of multiple readings in succession to return that no obstacle is detected prior to resetting the algorithm, rather than the current implementation of resetting after just one such reading. This would not only prevent this scenario, but also allow the situation where a user is talking to an individual in front of them while often moving in and out of the ultrasonic sensor’s field of view. While this case study helped analyze and validate much of the system’s behavior across multiple surfaces and gait activities, it also highlighted corner cases that had not yet been thought of, substantially improving the design’s future behavior.
Chapter 6

Future Work and Implementations

Continued research into this project is broken into two core areas: an overall system validation study, and future implementations to improve the design. The study will try to determine the system’s usefulness as a diagnostic tool for physical therapists by having individuals execute a series of tasks to test the system’s classification algorithm. While this study uses the cane designed in this thesis, as it focuses more on the software component of the design it was not completed as part of this thesis specifically. In addition to the study, several areas of improvement and ideas for any next-generation implementations are discussed.

6.1 System Validation Study

In order to assess the design’s overall utility as a diagnostic tool for physical therapists, a research study will be completed involving multiple groups of individuals using our cane. As summarized throughout this paper, the goal of the overarching project is to see if sensors could beneficially augment a physical therapist’s analysis of a person’s gait performance. This involves having individuals perform a variety of walking tasks and predicting the activity using the trained classifiers discussed in Chapter 4. While the study has not been completed, the methods and procedures for it have been finalized as part of the study application. This section details this study by discussing the walking tasks and procedural methods.

To help determine the utility of using our device to aid physical therapists, we need to have individuals execute a series of tasks that the therapist would typically have an individual do in order to have a common baseline. The therapist would be familiar with
the tasks, and the similar nature of the experiment would help show just how much the quantitative measures could improve their ability to best assess the gait of their patients. As such, the study involves a series of seven basic walking tasks akin to a modified Dynamic Gait Index (DGI) assessment. The tasks will be completed at the Pi Beta Phi Rehabilitation Center under the guidance of Patricia Flemming, PT.

The study is comprised of two groups of ten individuals. One group consists of healthy, young adults recruited from within Vanderbilt University while the other group are individuals with demonstrated gait abnormalities. For this reason the second group consists primarily within the elderly population. While the young adults will be selected with a much less stringent criteria, we will rely on the expertise of the physical therapist to identify potential test subjects for the other targeted population. As they do not have any gait abnormalities, the healthy individual group provides a way to collect model data to help train the classification models on reliable data. It also ensures that the system is robust for testing with the second population group. The subjects of both studies engage in the same tasks during the experimental procedure. All of the tasks are daily activities carried out by most mobile individuals and are, generally speaking safe, simple for the participants to perform, and simple for the researchers to manage and monitor.

Our modified DGI walking assessment consists of three major tasks: walking normally for a given distance, walking normally for a given distance with eyes closed, and completing the standard DGI. The first task asks the individual to walk at their normal pace along a straight twenty foot path, and will be performed for three repetitions with brief pauses between each round. This allows a baseline to be developed for the individual. Task two
then asks the individuals to repeat task one but with their eyes closed, continuing to walk until the administrator told the subject to stop.

Task three involves executing each of the DGI’s eight walking exercises once. Exercise one is a repeat of Task one, walking along a straight path normally. Exercise two is a test of an individual’s ability to change gait speed. Individuals walk normally for five feet before walking as fast as they can for five feet when the administrator says “go”. They then walk slowly for another five feet after being told “stop” by the administrator. For exercise three the subject walks normally until they are told to “look right”, after which they continue to walk straight while looking over their right shoulder. Finally, the administrator says “look left” and the subject continues to walk while looking straight ahead. Exercise four tests the individual’s gait performance subjected to vertical head turns. While walking straight ahead normally throughout the exercise, the subject’s line of sight goes from straight ahead, to looking up, to looking down, to looking straight ahead again after the administrator says “look up”, “look down”, and “look straight”, respectively. Exercise five sees how the subject’s gait is affected by suddenly pivoting. While walking straight ahead the individual turns and stops when the administrator instructs them to. Exercise six tests an individual’s ability to step over an obstacle. After beginning to walk at their normal pace, at a certain point the subject encounters an obstacle (e.g., a shoebox) which he/she should step over rather than around and continue walking to the end of the path. As this study will be conducted in a controlled environment and the object is part of the experiment, the feature detection subsystem (including the vibration motor) will be disabled during this time. Exercise seven is similar to exercise six, however the individual is requested to step around the obstacle rather
than over it. Finally, exercise eight assesses an individual’s gait performance on a flight of stairs. The subject begins at the base of the stairs, climbing to the top when prompted by the administrator. After reaching the top of the stairs, the subject turns around and descends the stairs. These tasks provide a physical therapist with a solid basis to help understand the current state of a patient’s current gait.

6.2 Improvements and Implementations

The first improvement that can be made is to store data on-board the cane, allowing the device’s usefulness to be extended beyond the physical therapist’s office. Currently, the cane only streams data to the host machine continuously during a individual’s session with the physical therapist. However, once outside of these sessions, the therapist is unable to detect any improper use of the device (such as unequal applied pressure at the base), or irregular gait patterns that appear outside of the controlled lab environment. This, coupled with the fact that most sessions only last an hour or two once a week, results in the majority of data related to an individual’s interaction with the walking device being lost. This means even the most experienced of therapists are limited to basing their recommendations on the relatively small set of data they have available. Thus, it would be useful to create a device which can gather data during a person’s day-to-day use of a walking device and then be downloaded by the therapist for analysis in between therapy sessions. This download could be executed during a time of non-use, such as when the device is plugged in at night to recharge and is connected to a wireless network. This would provide the therapist with an increased level of understanding as it relates to the individual’s gait. While their trained eyes can be the best tools to determine if a person’s overall posture are correct or not minute differences, such as
pressure changes while gripping a handle or between the base of the device and the ground, will be harder to detect. If a therapist is given access to data covering a larger sample period, their ability to assess an individual’s gait improves.

Another possible use case for this device which would warrant further work is to utilize the obstacle detection subsystem to detect how many and how often obstacles are being encountered by an individual in a given day. Dr. de Riesthal, who works with mobile rehabilitative patients, mentioned that commonly clinicians will visit the homes of patients to get an idea of what furniture may need to be moved around for safety concerns. They will then move the furniture with hopes to reduce the risks of experiencing a fall or near-fall. Were the device able to store data while the device were used in a patient’s home, in addition to detecting/notifying users of potential obstacles, a counter could be instituted to keep track of how many obstacles were encountered during a typical day. Doctors could then analyze this data to see if this furniture-moving activity provides a substantive benefit or not to the patient.

Two other improvements would be to move the batteries from the base module into the shaft of the cane, and redesign the fixture for the ultrasonic sensor. Moving the batteries from their current placement into the cane’s shaft would keep more of the added system components within the cane’s volumetric envelope, making it less distinguishable from a standard, non-instrumented cane. A part could be 3D printed to hold the batteries, and could be placed above the top height-adjustment dowel-pin opening. The major hurdle to this improvement would be ensuring that the batteries do not interfere with the USB cable running through the shaft connecting the handle and base modules. In its current
implementation, while the ultrasonic sensor’s housing allows it to be securely fixed to the
shaft, the sensor is exposed and therefore vulnerable to damage, whether it be physical
contact or a substance like water. While this is not an issue in the cane’s current operating
environment of a lab, for the device to be used outside of this environment more precautions
would need to be handled. Redesigning the housing to fully enclose the sensor would decrease
the likelihood of it being damaged, and allow for outdoor use.

A new feature that could be added to the system is the ability to sync metadata from
the cane to an individual’s smart phone. An individual could load a custom application
on their phone to let them statistics such as how much of a battery charge they currently
have, how far they have walked, how many steps they have taken, et cetera. In addition,
if the previously-mentioned improvement of storing data on-board the cane to allow it to
be periodically downloaded by a physical therapist was implemented, the therapist could
analyze data between appointments. The therapist could then send the patient notifications
to say their appointment could be pushed back because of the patient’s gait improvement,
or to say they should come in earlier due to a noticed decline in performance. This would
result in a better use of both the patient’s and therapist’s time by eliminating non-useful
meetings.

Finally, an idea for a future implementation could be to move many of the electronics in
the handle and base (aside from the ultrasonic sensor, load cell, and FSRs) into an attachable
external module, similar to [20] and [30]. While the current implementation tries to utilize
the current space available within a standard offset cane, doing so presented a number of
challenges given that much of the space in the handle is taken up by the cane’s shaft. Moving
these components to an external module would remove the need for these difficult design
decisions. Another possible benefit of this implementation could be the ability to not only
hot-swap the base and ferrule, but swapping the type of cane used. If all of the system’s
electronics were contained in one module, it could be possible to unattach the module from
an offset cane and attach it onto a standard cane with a derby handle. This would open the
device to be used by individual’s who choose to use non-offset type canes.
Chapter 7

Conclusion

Many individuals will utilize a walking aid at some point in their life, whether it be due to old age or recovering from an injury. Physical therapists provide an important service in this regard by helping ensure an individual maintains a normal gait, as well as diagnose any atypical patterns that they may see. While physical therapists are well-equipped and skilled to distinguish between gaits, their decisions are primarily based on experience and qualitative assessments rather than quantitative, empirical data.

This thesis presented the new design of an instrumented walking cane with the objective of creating a diagnostic tool that could be used by physical therapists. A previous iteration based around a standard cane with a derby handle was redesigned to use an offset style cane, and required major changes be successful. This included redesigning the location of (and housing for) all electronic components that were previously in the handle of the cane due to the fact that the shaft of an offset cane goes through the handle, something that is not the case for a standard cane. To accomplish this, the hollowness of the cane’s shaft was used to our advantage by rapid prototyping an insert that could slide in and out of the shaft, allowing components to be placed inside the cane while still allowing the designer to quickly access the components if needed. A base module was also redesigned to house the system’s batteries and a load cell which is used to detect the force enacted on the ground by an individual as they walk. This load cell, along with eight force sensing resistors on the handle and two IMUs, provide data that can be analyzed by an off-board software program using trained classification models to predict a user’s current walking activity. This data was
acquired by the sensors from a microcontroller and wirelessly streamed to the host-machine using an RF transceiver.

In addition to designing a system for use with physical therapists, a feature was examined that would also be useful to the public in a less-controlled environment. An ultrasonic sensor plus vibration motor subsystem was created to help detect and alert individuals of potential obstacles in their current walking path to help prevent a fall. The detection algorithm was also designed to prevent notifying users in the event of the false-positive identification of obstacles, such as walking up a flight of stairs. Given the country’s aging population, the number of individuals that utilize walking canes, and the likelihood of experiencing a fall while using a walking aid, having a device that can help prevent such a scenario from occurring can be of great importance.

The design was developed with the idea of creating an instrumented device that stayed within the tight volumetric envelope of its non-instrumented counterpart. This thesis discussed the various aspects of the system and provided insight into the hows and whys of various design decisions that were made, as well as their implementations. The system as a whole was also designed with modularity in mind, allowing for the swapping of ferrules and disconnection of the entire base module without breaking up the rest of the system. The evaluation of the design’s subsystems showed that the force measurement and obstacle detection features worked as anticipated. A subsequent study was described which has the ability to demonstrate the reliable prediction of a user’s activity based only on force and inertial data. This data can then be utilized by physical therapists by painting a clearer
picture of their patient’s gait, and base any treatment decision or recommendation off of these objective numbers.

Our device has the potential to be used outside of a controlled setting by individuals to capture a larger set of data during their day-to-day activities and serve a role as a fall prevention tool. In addition, the system addresses a need from physical therapists for a tool that can produce a more quantitative measurement of a person’s ability to sufficiently walk while using a cane. The future of this project will involve investigating other designs that may further promote the device’s use in both clinical and public settings, as well as continuing to increase the system’s modularity.
Bibliography


