Design of a Range of Motion Sensor for Ankle Rehabilitation Monitor

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Electrical and Biomedical Engineering Project Report

submitted in partial fulfillment of the degree of

Bachelor of Engineering

McMaster University

Hamilton, Ontario, Canada

April 9th, 2010

ABSTRACT

The purpose of this project is to design a convenient and cost-effective way for a physiotherapist or other health-care professional to quantitatively measure progress in the rehabilitation of a patient’s range of motion in their injured ankle in a clinical setting. Ankle injuries are one of the most common types of athletic injuries and can severely affect performance and daily functioning and proper rehabilitation is frequently necessary to recoup diminished proprioception and range of motion in the joint. Currently, no fast, cost-effective, and convenient quantitative method of tracking a patient’s improvement in range of motion in a clinical setting is available. This project attempts to accomplish this result using two bend sensing resistors embedded in an ankle sleeve along the anterior and lateral sides. These sensors have a variable resistance, which changes for each sensor when the ankle moves in plantarflexion/dorsiflexion and in inversion/eversion. The sensors are attached to a voltage divider circuit that outputs two voltages that vary relative to the changes in resistance of each sensor. The output voltages are converted to the respective angles of bend in the two different directions on the computer and are displayed on graphs vs. time on the computer. The data acquired from the sensors during an exercise specified by the health-care professional can be saved on the computer for further review and can be compared with other past results in order to gain a quantitative perspective of the patient’s improvement in range of motion over different sessions. All of these components were designed and assembled for the prototype. The prototype components of this project are all fully functioning except for the bend sensors, which produce neither precise nor accurate results, and do not function properly or consistently. Because of this, the prototype does not achieve the desired goals, but if a more precise or accurate bend-sensing product were available the design concept would function.

Keywords: Range of motion; bend sensor; bi-directional bend sensing resistor; ankle joint; rehabilitation; and physiotherapy.

ACKNOWLEDGEMENTS

The authour would like to acknowledge the immeasurable help and contributions provided by Evan Downie in partnering this project and Dr. Patriciu for his support and guidance while supervising.

Thank you to everyone else who assisted along the way, including, but not limited to, Mike Willand and the course T.A., Saba Mohtashami, for helping with the some of the work done in LabVIEW, Dr. Doyle for teaching the course and to Rakefet Hootnick in the David Braley Athletic Centre for providing advice with the biomechanical and physiotherapy-related aspects of the project.

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NOMENCLATURE

Following is a list of terms used in the report, with their respective definitions, listed in alphabetical order.

ADC – Analog to Digital Converter, converts a continuous analog voltage input into a discretized digital form for the software to interpret.

DAQ – Data Acquisition system, used to acquire a signal in order to process it.

Dorsiflexion – Movement of the foot at the ankle joint such that the toes point upwards [opposite of plantarflexion].

Eversion – Movement of the foot at the ankle joint such that the sole moves laterally (i.e. outwardly from the body) [opposite of inversion].

IE – Short form adopted for inversion/eversion. Describes motion around subtalar joint axis that includes inversion and eversion. The IE Sensor is the sensor on the lateral side of the device, used to measure movement in IE.

Inversion – Movement of the foot at the ankle joint, such that the sole move medially (i.e. inwards towards the centre-line of the body) [opposite of eversion].

LabVIEW – Laboratory Virtual Instrumentation Engineering, a visual programming environment from National Instruments used for data acquisition and displaying graphical outputs.

PCB – Printed Circuit Board: A circuit board with foil on one side connecting rows of holes to allow for soldering circuit components to it.

PD – Short form adopted for plantarflexion/dorsiflexion. Describes motion around talocrural joint axis that includes plantarflexion and dorsiflexion. The PD sensor is the sensor on the anterior portion of the ankle joint in order to measure movement in PD.

Plantarflexion – Movement of the foot at the ankle joint, such that the toes point downwards [opposite of dorsiflexion].

ROM - Range of Motion, the extent to which a joint can comfortably move in the given directions.

CHAPTER 1

INTRODUCTION

1.1 Background

Ankle injuries are very common and proper rehabilitation of the joint is imperative to daily functioning and performance, since proprioception and range of motion can be significantly affected by injury [1]. Motion at the ankle can be described by rotation about three different axes, allowing for movement in plantarflexion-dorsiflexion (PD), inversion-eversion (IE), and internal and external rotation [8].

Monitoring of improvement in ankle function, specifically range of motion (ROM), is an important aspect of rehabilitative therapy. Physiotherapists will often try to account for improvement in a patient's ROM using subjective, qualitative measures like watching a patient perform an activity or feeling the patient's passive ROM manually, while comparing findings with the healthy ankle as the 'normal' range [8]. These techniques do not allow for quantitative measurements of improvement in ROM or quantitative comparisons of the joint with the healthy ankle and quantitative techniques like using a goniometer to measure ankle angle at a fixed point do not offer information on the full range of a patient's movement.

While some techniques have been designed to monitor ankle movement, most of these techniques are rely on controlled laboratory settings or involve cumbersome devices, none of which are suitable for regular use in a clinical environment and many of them do not directly measure ankle ROM.

By using thin bend sensors embedded within a wearable ankle sleeve to determine the angles of a patient's ankle bend while they perform an activity, this project can be used in a clinical setting, allowing physiotherapists or other health care professionals a more quantitative analysis of patient rehabilitation, while not adding a lot of inconvenience or wasted time to implement it. When used in concert with the Plantar Forces Sensor, this design concept also merges both the position of the leg relative to the foot as well as the downward forces experienced by the foot, which should offer to illuminate even more information about the ankle mechanics than is currently available.

1.2 Objectives

The objective of this project is to produce a device that can monitor the range of motion of a patient's ankle joint in dorsiflexion/plantarflexion (PD) and in inversion/eversion (IE) in order for a health-care professional to quantitatively examine the patient's recovery from injury. The device is an ankle sleeve to be worn by patients of physiotherapists or other health-care professionals or trainers, meaning that it is designed for easy, convenient use in clinical settings and is designed to operate within normal athletic footwear while not hindering the patient's ability to move. The device is designed to offer quantitative information about the patient's angle of bend in PD and IE throughout an exercise, operating through the full range of motion of both PD and IE, described in Table 2.1, and should be precise on the order of degrees. Data from the device should be acquired at a rate of 1000 samples per second and the information is aimed for convenient display on a computer, allowing a health-care professional to scan through the patient's movements on a graph with the ability to save data, review data and place comparative data, such as data coming from a second sensor on the opposite ankle or previously saved data, on the same axis.

1.3 Methodology

The project's design process was split into four distinct parts. The first portion of the project involved establishing a way to determine the angle of a patient's joint in both PD and IE. A number of types of sensors were explored before bend sensing resistors were settled on as the modality of choice. Once these sensors were acquired, calibration was necessary. Due to the nature of these sensors and their implementation, calibration could not be completed until the rest of the components were designed and implemented.

The design of an ankle sleeve used to house the sensors was another distinct component of the project. The purpose of the ankle sleeve is to hold the sensors in place so they can properly measure angles in PD and in IE while not hindering the movement of the patient. First, research was performed in order to determine the best placement for these sensors. Store-bought materials were used to sew together the ankle sleeve with slots for the sensors located along the lines determined for repeatability and that would experience the most amount of bend for the direction they were measuring. During and after assembly of this ankle sleeve, testing was performed to ensure ease of use, mobility of the ankle joint while the sleeve was on, functionality of the device with the sensors embedded, and whether the device would comfortably fit within normal athletic footwear such as running shoes, skates, etc.

Another component of the design process was the design of the circuitry to implement the sensors. A voltage divider circuit was determined to work with the sensors obtained and so the resistor value necessary for the circuit's operation was determined and the circuit was assembled. The circuit was later soldered onto a store-bought PCB along with batteries hooked into voltage regulators acting as the sources for the circuit and along with the circuitry components of the Plantar Forces Sensor that accompanies the Ankle ROM Sensor for the Ankle Rehabilitation Monitor. The circuit was designed to output a voltage that is dependent on the behaviour of the bend sensors to the NI ADC available in the course laboratory.

The final component of the project was the software implementation, which was programmed in LabVIEW. The first function of the software is to retrieve the voltage data from the circuitry at the appropriate sampling rates and to translate these voltage values into the corresponding angular information. This first involved the calibration of the sensors to determine the transfer function necessary. The angle values obtained were then programmed to be output onto a graph along with a few comparative data options that are available to be controlled by the user. The software was programmed for use in a clinical setting by an individual who doesn't necessarily have a programming or computers background. This involved using a number of push buttons and simulated LEDs to light up when a particular option has been selected. The software was programmed to allow the user to adjust the scales on the graph and to be able to pause and review data, reset graph, as well as the ability to record data and save it to a file and to review saved data. Comparative value options were including, allowing the user to display a 'comparative value' on the same axis as the current incoming values on the graph. The two options for the comparative inputs made available to the user include the ability to display a second set of live incoming data (coming in from a second sensor placed on the opposite ankle) and the ability to display pre-recorded data from a saved data file on the same axis as the live incoming input.

The design and implementation of each of these components is discussed in further detail in Chapter 4.

1.4 Scope

The scope of this project includes the design and prototype assembly of each of the components discussed in the design section. This includes an ankle sleeve to hold sensors, basic calibration of the sensors, a circuit to use with the sensors and design of software to display the information from the sensors. The theoretical design of this project is aimed towards a clinical setting; so all aspects of the project have been designed with this idea in mind. However, due to various constraints, some aspects of the project have been designed as a proof of concept and their prototype implementation differs somewhat from how a market-ready version of this product would appear.

The bend sensors purchased for this project are very inexpensive, store-bought sensors that are used to prove the design concept instead of for assembly in a fully functioning device. Due to the limitations in the bend sensor technology, the sensors were not calibrated (this is discussed later in the report), however, the intended calibration technique originally planned for the project is described in the report. The sensors purchased are an experimental technology that is not market-ready at this stage. However, since the sensors offer a more attractive design implementation than the more established bend sensors already available, these sensors were chosen for use in the design of this project, despite their limitations.

The ankle sleeve portion of the project includes the design of a prototype sleeve to hold the bend sensors. This sleeve was hand-sewn and the placement of the sensors was determined to try and optimize the ability of the sensors to determine bend on the ankle. This sleeve was aimed at being comfortably wearable within normal athletic footwear, while offering as little hindrance as possible to the test-subject’s ankle. The prototype built was designed to work specifically with the test-subject in order to prove the design concept. This is instead of being designed to have a more universal fit, which is what the future aim of the product would be.

The circuitry designed and soldered for this project is fully functioning and has been soldered to a store-bought PCB rather than a custom designed one. Wires from the lab are used, attaching the sensors to the circuit, again to prove the design concept, since future applications would likely be wireless or at the very least offer more flexible wires that are more amenable to moving around and being bent.

A fully functioning user-interface was designed with an aim towards easy use and straightforward presentation of information. The data presented to the user is not manipulated further providing information such as torque (which is another future design consideration that will be discussed), but the UI allows the user to record data, compare two different types of data on the same axis and review saved data. The software also allows the user the ability to change the scales on the graphs and pause and review data coming in, as well as the ability to reset the graphs. The software has been designed to handle most types of errors that might arise and attempts to be very user friendly. The UI was designed using LabVIEW, meaning that a computer with LabVIEW is currently required to use the software for this project.

CHAPTER 2

LITERATURE REVIEW

Ankle injuries are one of the most common types of athletic injuries, and can severely affect performance and daily functioning. After an ankle sprain, proprioception and range of motion tend to decrease and if the injury isn't treated properly, chronic ankle instability is a common result [1]. This is why proper rehabilitation of the joint is imperative. Ankle injuries are also a very common injury for elderly people and can severely affect mobility and quality of life.

Motion at the ankle can be described by rotation about the 3 axes perpendicular to the anatomical planes [2]. Normal ranges of motion about these axes are listed in Table 2.1. Plantarflexion and dorsiflexion (referred to collectively as PD throughout this report) can be described by rotation around the talocrural joint axis, shown in Figure 2.1. Inversion and eversion (referred to collectively as IE throughout this report) can be described by rotation about the subtalar joint axis, also shown in Figure 2.1 [8]. Using these parameters of movement, locomotion is generated by transmitting torques and forces to the ground [2]. Using the information from Table 2.1, it is apparent that the range of motion from maximal dorsiflexion to maximal plantarflexion is between about 58 and 76 degrees total, while range of motion from maximal inversion to maximal eversion is between about 24 and 39 degrees.

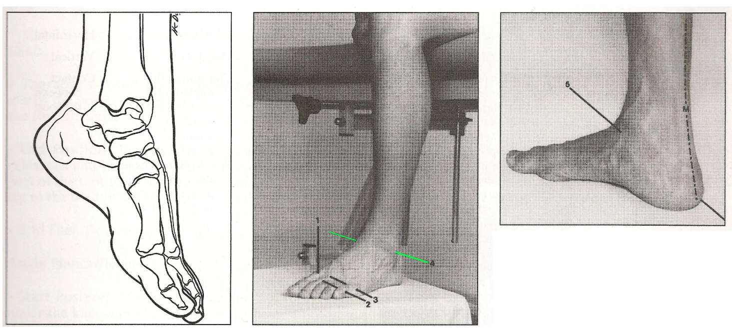

Figure 2.1; Axes of movement at the ankle joint. Middle (green): talocrural joint axis for PD. Right: subtalar joint axis for IE. From [8]

In clinical settings, physiotherapists will measure passive ROM and active ROM using more subjective measures involving their feel and sight. Active range of motion can be analyzed by a physiotherapist by having them watch a patient perform some task, such as walk back and forth, while monitoring their ankle's range with their sight. Passive range of motion is established by a physiotherapist by manipulating a patient's ankle with their hands and arms and feeling for range of movement experienced, shown in Figure 2.2 [8]. Quantitative measurement of the patient's ROM can be performed using goniometers while performing these ROM assessments. However, these measurements are only taken for maximum ranges in the ankle joint and don't show any functionality. Furthermore, these quantitative measurements can only take place while the patient is stationary and on their back, allowing the physiotherapist to manipulate their ankle in order to perform the measurements [8]. This means that quantitative information on a patients active ROM cannot be gathered while the patient is active and performing an exercise. During rehabilitation, physiotherapists will compare the injured ankle to the healthy ankle in order to try and figure out what the 'normal' function of that patient's ankle should be. These comparisons could be greatly enhanced with more quantitative analyses of the ankle joint during rehabilitation [8].

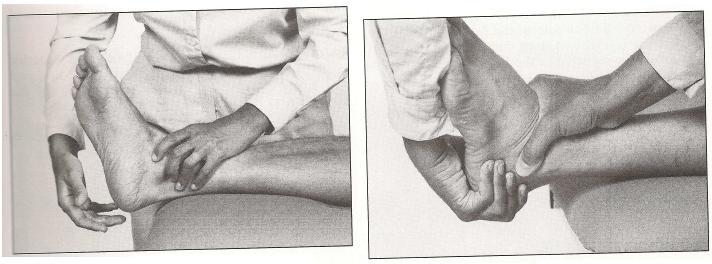

Figure 2.2; Physiotherapist feeling for passive ROM

Many techniques are currently being explored to model the function of the ankle during activity. For example, measurements of temporal and spatial parameters in gait are frequently used for rehabilitation purposes by identifying gait deviations, screening elderly for risk of falling, determining the effectiveness of therapy interventions and monitoring patient progress [3]. Products such as GAITrite and other gait monitoring devices can be used in this way by recording a patient's gait using a network of force sensors [3] either attached to the patient's feet or embedded in the walking surface, air pressure sensors [4], video monitoring [5], and various other techniques. Other implementations for modelling ankle function and tracking rehabilitation include use of rehabilitation robots [6] and other even more cumbersome techniques. Implementations for measuring range of motion in other joints also tend to be cumbersome [7], since only specific, controlled motions are needed for most joints and less controlled, weight-bearing activity with motion on multiple axes is not involved.

These methods of monitoring tend to rely on being in a controlled lab setting and involve cumbersome devices that cannot be easily implemented outside of the lab. Furthermore, they tend to focus solely on controlled motions, such as a steady walk on an even surface for gait analysis, as opposed to various other types of activities in more of a real-world setting. To add to that, the devices that monitor gait do not measure ankle function based on both ankle position (ROM) and downward force.

Motion	Maximum Allowable Motion (in degrees)
Dorsiflexion	20.3 – 29.8
Plantarflexion	37.6 – 45.8
Inversion	14.5 – 22.0
Eversion	10.0 – 17.0
Internal Rotation	15.4 – 25.9
External Rotation	22.0 – 36.0

Table 2.1; Normal ranges of motion for the ankle joint [1]

CHAPTER 3

STATEMENT OF PROBLEM

Ankle injuries can severely affect the range of motion of patient's ankle and recovery of range of motion and proprioception are necessary parts of patient rehabilitation. In clinical settings, range of motion is most often judged qualitatively and subjectively by health-care professionals. These methods most often involve manually feeling for passive range of motion while manipulating a patient's ankle by hand or viewing a patient's ankle function and range of motion while the patient performs some activity such as walking back and forth or balancing on a wobble board. These methods of monitoring range of motion lack quantitative information, especially when used to compare the injured joint with the healthy joint, which is a useful tool for physiotherapists and other health care professionals to use since ideal joint function varies from one individual to the next making the healthy joint the best frame of reference for ideal function. More quantitative methods of ROM measurement in settings such as physiotherapy clinics, doctor's offices, or training facilities can be used, including using a goniometer to measure the angle of a joint through its maximal range, but these types of methods are generally more time-consuming and inefficient and don't offer as much information for joint analysis, due to the fact that these methods do not follow the patient's range throughout an exercise but rather only measure a maximum angle of movement around a joint axis. Outside of clinical settings, more quantitative methods of monitoring joint range of motion are available, but these types of methods, such as analyzing gait using force sensors or video monitoring of a subject's ankle, tend to be more cumbersome, time consuming, and inefficient, making them unsuitable for use in clinical settings. A convenient, efficient and inexpensive method of quantitatively monitoring a patient's ankle range of motion during activity could greatly enhance rehabilitative processes in clinical settings by offering quantitative information on a patient's entire range. Analyses of patient joint function could be further enhanced by combining this range of motion information with the plantar force information provided by Evan Downie's Plantar Forces Sensor. Collectively, the information provided by these devices could be used for complex analysis of joint function, including measurement of torques and strains experienced by the ankle, and could have implications in rehabilitative practices, athletic training and the design and implementation of custom ankle braces or other athletic safety equipment.

The goal of this project is to design and build a prototype for a device that can quantitatively measure a patient's range of motion throughout an exercise in two desired axes of motion, plantarflexion-dorsiflexion (PD) at the talocrural joint and inversion-eversion (IE) at the subtalar joint. This device is designed to fit comfortably on a patient's ankle such that they can perform exercises and actions that might take place during a physiotherapy session for quantitative analysis of their joint movement. Information gathered from the device is displayed in real time on a computer screen where the health care professional has the ability to compare the data with other data sets (either pre-recorded or live incoming data) in a convenient, easy-to-use user-interface.

The design process is discussed in detail in the following chapter.

CHAPTER 4

EXPERIMENT & DESIGN PROCEDURES

This project involves the design and implementation of four distinct parts: The use and calibration of bend sensors to determine the angles of bend of a patient's ankle; the design and assembly of an ankle sleeve to hold the sensors in place to obtain ROM data in PD and IE; the design and assembly of circuitry in order to utilize the bend sensors and transmit information to the computer; and the design of LabVIEW software in order to display, record, review and compare ROM information.

4.1 Implementation & Calibration of Bend Sensors

The first step in the project was to determine a way to measure bend in the patient's ankle. Bend (or flex) sensing resistors were chosen for use due to their relative simplicity in implementation. It was determined that they could easily lie along the different planes experiencing bend in order to deduce the angle of bend in that direction. Other types of sensors, such as Hall effect sensors and tilt sensors were considered, but each of the types of sensors considered, aside from bend sensors, did not directly measure the angle of the ankle's ROM.

After researching bend sensors online, it was found from Robotshop.ca that bi-directional sensors, sensors that could detect bend in both directions were available. These differ from two-way bend sensors in that two-way bend sensors have a change in resistance when bent in either direction, but the resistance changes (either increasing or decreasing) the same way in both directions. Bi-directional bend sensors are different in that they increase in resistance when bent in one direction and decrease in resistance when bent in the other direction. This product, however, is still in the experimental stages, and so some issues were to be expected. In order to use two-way bend sensors for the project, a method of determining which direction the bend occurs in would have been needed, likely meaning having to purchase and arrange for four sensors to go into the device. The bi-directional bend sensors allowed the device to only house two sensors, allowing it to be more maneuverable and less expensive for a prototype to be designed, while also allowing the calibration stages to be easier (theoretically).

Figure 4.1; Bi-directional flexible bend sensors used for the project

Two bi-directional bend sensors were purchased from RobotShop.ca, these are the Images Scientific Bi-Directional Flexible Bend Sensors FLX-02, shown in Figure 4.1. These sensors have a nominal resistance when kept straight and the resistance increases when bent in one direction and decreases when bent in the other. The change in resistance can be determined using a voltage divider circuit with a resistor matching the nominal resistance of the sensor (the circuitry is described in Section 4.3). These sensors have a thin profile making them easy to fit inside of an ankle sleeve and within a shoe. The dimensions of these sensors are shown in Table 4.1 and Figure 4.2. Upon contacting the seller about these sensors, it was established that this product is still not finished and is currently aimed towards hobbyists rather than being aimed for mass production. It was recommended that the devices be calibrated within whatever unit they were being used for and that each individual device be tested. The first step involved upon receipt of the sensors was to determine the nominal resistance of each sensor. On first receipt it was determined that the nominal resistance for the first sensor (eventually set established as the PD sensor) was 43 KΩ and the nominal resistance of the second sensor (to become the IE sensor) was 35 KΩ. For ease of implementation a resistor between these two values was decided for use and based on availability, a 38 KΩ resistor was chosen. It should be noted that throughout the course of this project, the sensors were very erratic in behaviour. On different occasions, wildly different nominal resistances were determined, sometimes as high or higher than 100 KΩ. The sensors themselves eventually broke down and only worked in one direction, and only producing small changes in resistance in that direction, resulting in barely distinguishable and wildly inconsistent outputs from the circuit. This made calibration of the sensors impossible, although the calibration technique is described below. In order to demonstrate the proof of concept, two-way bend sensors were borrowed from the project supervisor, which functioned well enough to show the function of the design.

Length	114.300 mm
Width	6.350 mm
Thickness	0.508 mm

Table 4.1; Bend Sensor Dimensions

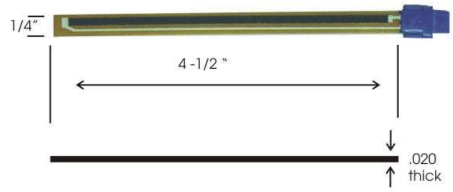

Figure 4.2; Bend Sensor Dimensions

The calibration initially intended for these sensors was attempted, yielding somewhat unusable results. This technique is described in the following paragraph, though it should be noted that it was not fully completed and the sample size used to demonstrate the concept was relatively small.

As mentioned previously, the sensors needed to be calibrated within the already built device, so calibration occurred after completing most of the circuit and ankle sleeve described in Sections 4.3 and 4.4, respectively. The assembled ankle sleeve was placed on the test-subject's ankle with the sensors embedded and hooked up to the circuit. The output of the circuit was attached to a voltmeter to display the output voltages. In order to calibrate the PD sensor, the test subject first bent his ankle in dorsiflexion to the maximum range he was able to reach, while staying neutral in inversion-eversion. This angle was measured using a goniometer and recorded with the corresponding output voltage of the circuit. The subject then bent his ankle five degrees less in dorsiflexion, measuring both the angle and corresponding output. This process is shown in Figure 4.3 below. This pattern continued for five-degree increments until there was no bend in dorsiflexion, and then the measurements proceeded into plantarflexion. This was repeated three times over for each angle. Since no outputs were received for most of the ROM outside of extreme dorsiflexion to around neutral dorsiflexion, and with the values yielded from these angles being fairly inconsistent, full calibration in this direction was not possible. Next, a similar process was attempted for IE. This was somewhat more difficult to measure since movement in IE is less linear than movement in PD. The method shown in [8] was used, involving having the test-subject lie horizontally, with a flat surface pressed against the bottom of his foot. The ankle was set to neutral position in both PD and IE and a straight line was drawn along the edge of the flat surface (in this case a book). The flat surface was then rotated slightly, causing the ankle to move in eversion. A straight line was drawn along the base of the flat surface again, and the angle between the two lines was measured using a protractor, with the output voltage of the circuit being recorded. This process, shown in Figure 4.4, was repeated in 5-degree increments three times over. In this case, no values were yielded for anything outside of a very slight, inconsistent signal at extreme inversion, so proper calibration in IE was also not possible.

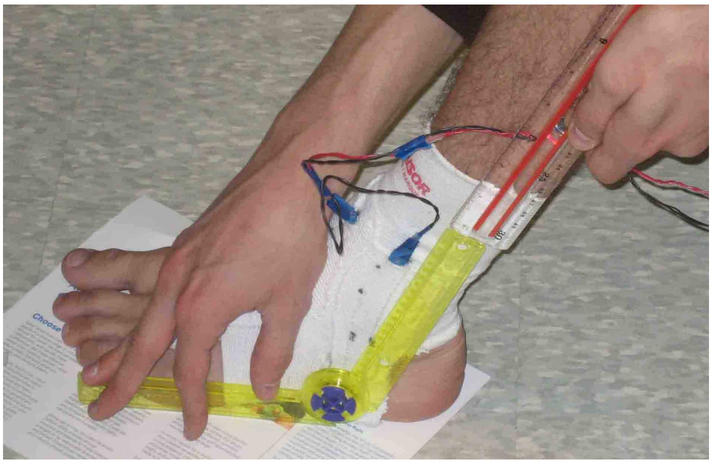

Figure 4.3; Calibration of PD sensor using goniometer

Figure 4.4; Calibration of IE sensor. Angle is measured using protractor on intersecting lines on the paper

The next step in the calibration was performed only for the few values yielded in dorsiflexion in order to demonstrate the process. First, the average of the three outputs recorded for each angle was taken and these averages were plotted in Matlab in a graph of Voltage Output vs. Bend Angle. Using the Basic Fitting function available in Matlab, multiple curves of best fit (each of a different order polynomial) were presented to fit these data points, along with the corresponding error associated with each curve. The fourth order polynomial was selected (This is shown in Chapter 5). It should be noted that in a real application, appropriate values at zero and five volts (the maximum and minimum outputs possible in the circuit) would have been added to the plots if not recorded in the calibration in order to allow for extrapolation outside of the ROM of the test-subject's ankle. This is important because higher order curves of best fit (offering less error) offer poor extrapolations from data sets. Another way to simulate angles outside of the test subject's range might have been determined if coherent values inside the range were obtained. This process would have been repeated with the data points from IE if they were available. These formulas, once obtained from Matlab, could then be plugged into the formula block functions used in LabVIEW to determine ankle angles from the corresponding voltage outputs of the circuit. This is described in Section 4.4.

4.2 Design of an Ankle Sleeve to Embed the Sensors

The purpose of the ankle sleeve is to hold the embedded sensors in a position along the ankle to measure each of PD and IE, while not hindering the movement of the patient and still being able to fit within normal athletic footwear. Two tracks were needed on the sleeve, one to hold a sensor along the anterior side to measure PD and the other to hold a sensor on the lateral side of the ankle to measure IE. The theoretical design of the sleeve is shown in Figure 4.7.

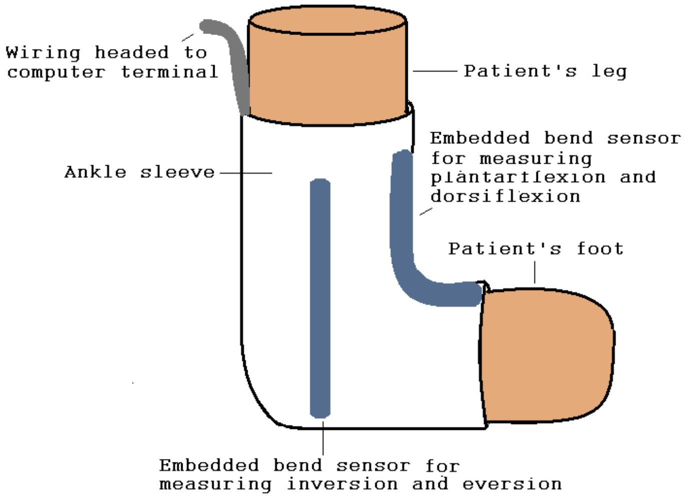

Figure 4.5; Theoretical ankle sleeve design

At first, a stretchy athletic sock was slated for use, but it was determined that this would not hold the sensors well enough since even tight socks allow for some sliding. It was surmised that a soft ankle brace could be used, since these hold fairly tight to the skin, while still allowing movement of the joint, although since it is designed as a brace, some hindrance is noticeable. Therefore a soft ankle brace one size too large for the test-subject was slated for use, since this would allow for more joint maneuverability, while still being tight enough to the skin to hold the sensors effectively. Two Tensor brand soft ankle braces were purchased, one for use as the sleeve and the other to use for modifying the sleeve. Ideal placement of the sensors was determined based on identifying a repeatable anatomic structure to align with while offering as much range for the sensors to detect as possible. Since the store bought brace was not long enough, either on the anterior section or the lateral portion next to the heel, small sections were sewed onto the sleeve to account for this. Tracks were then sewn onto the front and lateral sides of the sleeve, allowing for the sensors to be embedded while still being removable. Finally, testing was performed to see how maneuverable the device was with embedded sensors and to see how much bend was offered each sensor. The final ankle sleeve prototype is shown in Figure 4.8, with sensors embedded on the test-subjects ankle and by itself.

Figure 4.6; Ankle sleeve prototype with sensors embedded on the lateral and anterior sides

4.3 Design of Circuitry to Utilize Bend Sensors

As mentioned previously, the bend sensors function as a variable resistor, so changes in resistance can be shown using a voltage divider circuit. The resistor used in the voltage divider was determined to be 38 KΩ. It was also decided in concert with the Plantar Forces Sensor project associated with this project that a 5 V source should be used in the event that a microchip was implemented with the circuit instead of the planned use with the NI ADC available in the lab. The voltage divider circuits (one for each sensor) were then designed using the 38 KΩ resistors, a 5 V source, and an two Op Amps (which required a –5 V source) to be used as a buffer. The theoretical circuit diagram is shown in Figure 4.9.

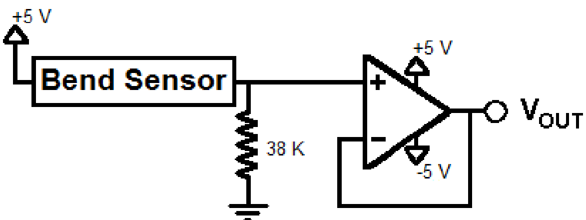

Figure 4.7; Circuit diagram for implementing bend sensors

This circuit was assembled on a breadboard with parts obtained from Tyler Ackland. The op amps used were LM741s and initially the sources available in the lab were used, with the intention of eventually using voltage regulators. This circuit was tested for functionality. A visit to Sayal Electronics was eventually made and a 78M05 +5 V and a 7905 –5 V voltage regulators were obtained for use in the circuit. The +5 V voltage regulator obtained functioned perfectly to specifications, but the –5 V voltage regulator purchased had a low frequency oscillation with an amplitude of ~1 $V_{pk\text{-}to\text{-}pk}$, making it unusable for application in the circuit. A method was determined in order to use another +5 V voltage regulator as the negative source, which involved hooking up the out pin of the 2^{nd} regulator (negative source) to the gnd pin of the 1^{st} regulator (positive source) and using the gnd pin of the 2^{nd} regulator as the –5 V source. A second 78M05 was purchased, and this method was determined to work, so the circuit was set up with these components on a temporary breadboard and a PCB was purchased in order to solder the components for the final implementation. For the soldered circuit, two 9 V battery holders were glued to the side of the PCB and the circuit components of the voltage regulators, acting as sources, the Ankle ROM Monitor and the Plantar Forces Sensor were soldered to the PCB, shown in Figure 4.10. For convenience, input pins were placed on the PCB with colour-coded wires to allow the long sensor wires to be removable. Output pins were also placed on the board with the same colour coding to allow convenient input into the NI ADCs in the lab. This circuit was tested as each component was added in order to ensure functionality.

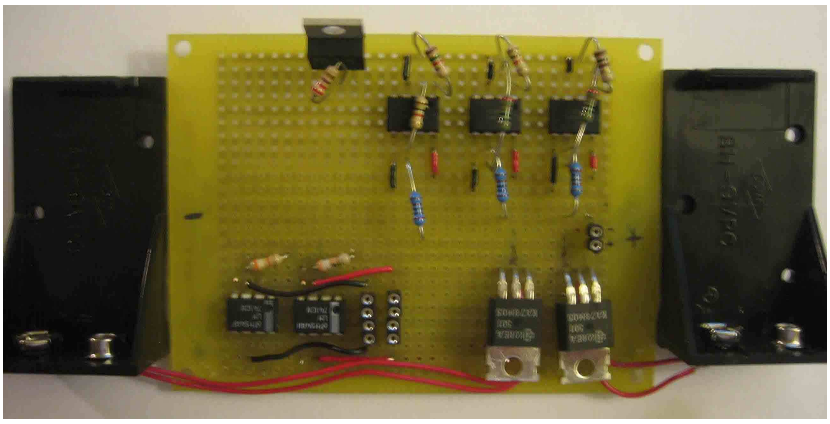

Figure 4.8; Soldered circuit. Bottom left is circuitry for bend sensors, bottom right are the voltage regulators and top is the circuitry for the Plantar Forces Sensor. Left and right sides are 9 V battery holders

4.4 LabVIEW Software Design for User Interface

The first step in the software design was ensuring proper data acquisition. The DAQ Assistant in LabVIEW was used in order to acquire the signals for both the Plantar Forces Sensor and the Ankle ROM Monitor from the in-lab ADCs. A method was needed to convert the incoming voltages to the corresponding angle values and the initial intention was to use a look-up table with the ability to interpolate for inputs in between the LUT input values. Unfortunately, the LabVIEW LUT function did not work properly in real time, so this was sacrificed in favour of using the Formula function, where the input voltage is placed into a pre-defined formula to define the output angle values. This formula was determined in the calibration portion of Section 4.1. The output angles were then displayed on a graph vs. time for the user to analyze, with many features available, explained below.

Following is an explanation of the design of the user interface (UI) for the software designed. The full block diagram of the software with explanations is available in Appendix A.

The user interface was designed primarily for ease of use, since this project is targeted towards a clinical market where the user will not necessarily have any kind of background with computers. Push buttons, switches and simulated LEDs were used on the interface in order to clearly present the options in available or in use to the user. The graph displaying the angles was made basic and clear, allowing the user to disseminate from it whatever information they need. A pause button and scroll bars were included with the graphs, allowing the user to go back and review data obtained earlier in the acquisition. Adjustable scales were also included, allowing the user to view all the data at once, or to have a set amount of data flow through the graph over the defined amount of time. The scales can also be adjusted very small in order to review a specific peak or section of information (in concert with the pause button if needed). There is also a reset button available in order to clear all the information on the graph.

Along with these graph options, the user was provided with the ability to record incoming data and store it in a user-specified file on the hard drive. A “Record” push button with a red indicator light are included on the UI to allow the user to easily initiate and terminate the recording of data, with a red indicator light visible on the screen to indicate to the user when data is being recorded.

At the top of the user options, there is a switch available allowing the user to choose between the incoming data from the Ankle ROM Monitor or reviewing previously saved data. All of the same options, including but no exclusive to scale adjustment, pausing, resetting the graph and scrolling through data, are available both for incoming data and for reviewing saved data. There is a light associated with the switch to tell you when incoming data is being displayed or when saved data is being displayed. There is also a “Stop” button located next to the saved data if the user wants to cease displaying a particular file.

Also available to the user is a control box titled “Comparative Values”. In this box, there are two different buttons available allowing the user to place a comparative input onto the same axis as the data currently displayed (either incoming or reviewed saved data). Two options are available for this, one being a live comparative input, which allows the user to have a second live input coming in in real time from a second device. A second device was not assembled, so this concept has been simulated in order to prove it works, but if a second device were built the software need not be changed (other than calibration formula) in order to implement a second device. This is important for clinical settings since physiotherapists will typically use a healthy ankle as a basis for comparison with the unhealthy joint being rehabilitated. The user need only press a large button labeled “Live Comparative Value ON” in order to initiate the display of a second comparative value, and the corresponding light will go on when this utility is in effect. The other comparative value option available is for recorded data to be displayed on the same axis as the current data, which may be a live input or pre-recorded as well. This allows the user to compare data to previously acquired data, such as from a previous session in order to monitor rehabilitation. Again, a light will go on when this functionality is in effect and a “Close File” button is available if the user would like to cease displaying the saved data before the file has reached its end. Both of the comparative values can be displayed at the same time (when a live input is being used), meaning that up to three data sets, each shown in a different colour, can be on the graph at once. Including one comparative saved input option shows that this function can exist, meaning that if more than one recorded value were desired for comparative display, these could be added.

Included in this project is a second program available that merges the data from the Plantar Forces Sensor with the data from the Ankle ROM Monitor. All of the same graph manipulations are available to the user as for the Ankle ROM Monitor software discussed above, but the ability to display comparative inputs on the same axis has been left out since this software is used to relate ROM with plantar forces. Instead pre-recorded comparative data can be displayed on a graph located directly below the incoming data display. Tabs are available on the UI allowing the user to view the Plantar Forces Sensor data, the Ankle ROM Monitor data, or both data sets at the same time on the same axis. Displaying this information on the same axis allows events, such as taking a step, to be isolated so that all relevant information in that instant can be analyzed.

Below is the LabVIEW front panel that serves as the user interface, in Figure 4.9. The block diagram is included in Appendix A with more detailed explanations of the programming.

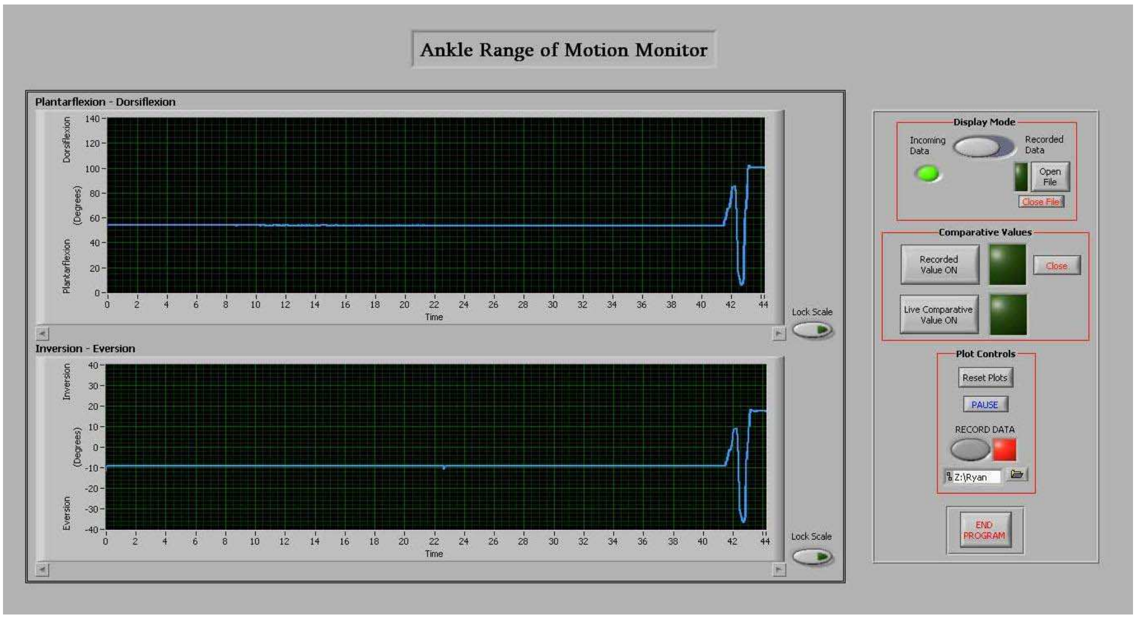

Figure 4.9; User interface, designed in LabVIEW

CHAPTER 5

RESULTS & DISCUSSION

5.1 Bend Sensors

One of the first steps in the project was deciding on a type of sensor to use and designing the rest of the device accordingly. It was determined that using a bend sensor would be the most straight-forward method of determining bend angle for a patient's ankle while managing to fit in a device that can comfortably fit inside of a shoe since these sensors are only 0.508 mm thick. The dimensions of the bend sensors used in this project are available in Table 4.1. Most available bend sensors can't differentiate between bend in either direction. They either increase or decrease their resistance when bent in both directions, but the sensors acquired for this project are an experimental product that are bi-directional, meaning that the direction of bend can be determined based on the change in resistance. When the sensors are bent one way, the resistance increases and when bent the other way, the resistance decreases.

When these sensors were first acquired, some testing was done in the lab to see how well they functioned. They were found to be overly sensitive to touch and consistent readings were hard to come by. Furthermore, the neutral position of the sensor was not straight, so in order to determine nominal resistance, the sensors had to be manually held straight, making it difficult to get a consistent value. After taking the average of multiple readings, the nominal resistance for the PD sensor was determined to be approximately 43 KΩ and the nominal resistance of the IE sensor was determined to be approximately 35 KΩ. Thus it was determined that a 38 KΩ resistor would function well enough in the voltage divider circuit to work with both sensors.

Throughout the project, whenever testing was done on the sensors, different, inconsistent readings were found for the nominal resistance. The above values were found in November, and throughout the course the values read from the sensors changed, including finding nominal resistances of greater than 100 KΩ in March. The circuit's resistor was replaced with a 100 KΩ resistor, which only worked for a very short period of time. Shortly after this, the sensors stopped working in both directions, producing outputs from the circuit that were greater than 0 V only for angles of bend in one direction, several degrees off from the neutral position. The outputs produced were also highly inconsistent. An example of inconsistent data obtained from movement of the ankle with the device placed on it is shown in Figure 5.1. During this test, the peaks were a result of random movements, and more often from perturbing the sensor or wire rather than bending it. The IE sensor failed to pick up any movement.

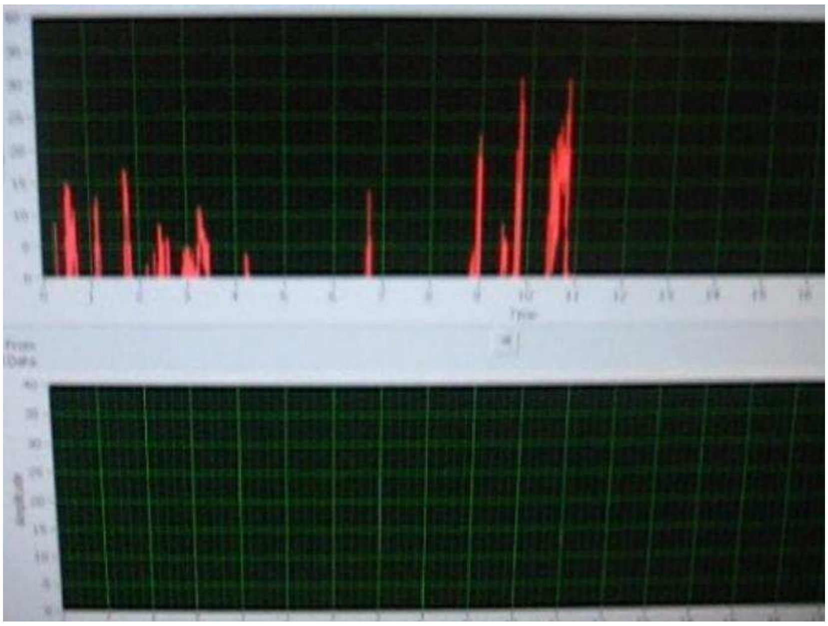

Figure 5.1; Actual recording of ankle movement. PD sensor shows spikes instead of smooth tracking of data, IE sensor failed to pick up any movement

Due to the failure of these sensors, a different method for determining bend would have to be considered for use. Two-way bend sensors could be used in the same way, but with some other method, such as Hall effect sensors or an accelerometer, to determine which way the ankle is bending. Other methods than bend sensors might also be considered for studying ankle angle measurement. Such a product might involve aligning a rod along the ankle that would run up the leg towards another rod, with the ends of the two attached to a Hall effect sensor used to determine how close together the rods are, and from that determining the angle of the ankle. Another method that could be used would be lining accelerometers on the ankle to determine its position. However, both of these applications would be considerably bulkier than the current model, making them difficult to implement in a clinical setting. Fibre optic bend sensors are a product that can accurately determine bend, however these products are generally used for civil engineering applications, which involves much larger implementations like bridges or buildings. Furthermore, these fibre optic sensors are very expensive and any project similar to this one that attempted to use these types of sensors would be far too expensive for the targeted clinical application discussed here.

5.2 Ankle Sleeve Design

The design of the ankle sleeve initially involved using an athletic sock as the basis. When the bend sensors were acquired some testing with spare socks was performed, involving hold the sensors to the sock with tape and testing how well the sensors stayed on the foot and how well they could be used to deduce bend. During these stages, it became clear that socks slide around on a foot to a degree that would have been significant enough to make most sensor readings inconsistent over the course of an exercise. Markers also had to be used a lot in order to ensure that the sock had the same placement on the foot each time it was put on. Eventually, this idea was dropped in favour of a method that could more consistently keep the sensors in place. Similar testing was performed with a soft ankle brace that was on hand and it was determined that using a soft ankle brace could solve these issues. This idea had originally been ignored since it was decided that ankle braces were more likely to hinder ankle movement. However, if a flexible brand were chosen and a larger size were used, this problem could be mostly alleviated and the brace would be stiff and tight enough to hold the sensors in place on the person wearing it. As an added, unexpected plus, it was found that the brace ended up in the same placement, pretty much, each time it was removed and replaced, making significant use of anatomic markers less necessary.

The ‘Tensor’ brand soft ankle brace purchased was found to allow significant ankle maneuverability and stayed on the foot extremely well, meaning that consistent values could be garnered from the sensors. However, it was found that the sleeve would be too short on both the anterior and lateral sides to properly fit the sensors, so modifications were made to the purchased brace to accommodate. These modifications were somewhat flawed, lacking a seamless profile where the brace ended and modification began, as would be expected from hand-sewing this kind of product. However, the seam was functional enough that the sensor was relatively unaffected.

Another issue that came up while designing the ankle sleeve was the fact that when the ankle moved around, the sleeve used and the skin both stretch and contract depending on the movement of the ankle, while the sensors are rigid from end-to-end meaning that they twirled or bunched instead of staying flat against the sleeve/skin of the test-subject. This problem was addressed using a track, where the top end of the sensor was fixed to the sleeve and the bottom end was kept in a track where it could slide along the proper line when the sleeve contracted or stretched. This functionality worked quite well, although tape had to be used to round out the tips of the sensors to ensure the end of the sensors did not catch on the sleeve.

The final build of the sensor worked very well for repeatable placement on the test subject's ankle and allowed for excellent mobility of the ankle, even while wearing running shoes and with the sensors embedded. The PD sensor moved with a very good range and functioned quite well, although it was difficult to try and prove this quantitatively due to the failure of the sensors. The IE sensor fit in the device well, but it was hard to determine how much movement the sensor actually saw. Since IE involves a smaller range and in a more difficult plane to measure, finding a location for the sensor that works well was a difficult task, and there are concerns that the sensors would need to be more sensitive to differentiate the angles with such a small range of values. This concern could be neither proved nor disproved due to the failure of the sensors to function entirely.

5.3 Circuitry

The circuit was initially built on a breadboard with the voltage sources in the lab being used to produce the +5 V and –5 V sensors needed. Using the 38 KΩ resistor discussed above, the circuit functioned very well in its task, producing an output that varied relative to the sensor, with values around 2.5 V being produced at the output from a nominal resistance at the sensor (no bend) and values decreasing towards 0 V when the sensor was bent in one direction and increasing towards 5 V when bent the other way. The erratic behaviour of the sensors prevented consistent readouts from the circuit, however, this was not a result of the circuit's functioning and all test inputs that were not from the sensor produced predictable outputs. It was therefore concluded that the circuit functioned perfectly to the design specifications.

A –5 V voltage regulator was initially acquired for use as the negative voltage source for the final implementation. However, on testing this sensor, it was discovered that the output of the regulator was sinusoidal in nature, oscillating at a very low frequency, with an amplitude of nearly 1 volt. Since this feature was unsuitable for the project’s needs, this component was discarded and a method of using a +5 V voltage regulator was determined, since the +5 V voltage regulator functioned quite well. This method, described in Chapter 4, functioned well when tested on the breadboard, producing the requisite –5 V voltage using a battery and the +5 V source’s ground input as the positive terminal of the regulator.

After sufficient testing of the circuit on the breadboard took place, the components were soldered to a store bought PCB. Colour-coded wires were used in concert with pins allowing the sensor wiring and outputs of the circuit to be removable for ease of use and testing. The same testing done with the breadboard was performed to ensure proper functioning of the soldered circuit, which was found to function just as well as the breadboard circuit. The colour-coded wires and input pins for the sensors and outputs made testing easy. The final circuit implementation worked perfectly to design specifications.

5.4 Software Design

The software design is discussed in significant detail in both Appendix A and in Chapter 4. The final implementation of the software functioned extremely well, with all the design specifications met. The software is very user friendly, allows the user multiple functionality options as far as recording data, displaying incoming data, displaying recorded data, and displaying multiple inputs on the same axis, all with easy-to-use push buttons and LEDs displayed to inform the user that a particular function is in use. This user-friendly functionality is important, since the target market for the product does not necessarily involve knowledge of computers. This makes ease of use paramount.

Several types of actions that would normally produce errors were performed with the software to test error handling, and the software was found to run quite smoothly and continuously despite the simulated errors. The graph functionalities such as pausing and scrolling back to view information, resetting the graphs, adjusting the scales, etc. all performed to their desired functions, as well.

The joint software implementation, designed from a combination of the above software and the software from the Plantar Forces Sensor, did not include all the specific comparative value functionalities of the above software, but allowed the user to show the ROM data alongside the plantar forces information. This software also performed quite well.

The only limitation seen with both software designs was that the function included in the Formula block of the software (discussed in Appendix A) did not contain a meaningful formula due to the results of the calibration, discussed below. A formula that normalized the input voltage (known to be between 0 and 5 volts) and multiplied it by the range of each direction of motion was used to demonstrate the software functioning while using some simulated input such as a variable voltage source in the lab or borrowing working sensors in the lab from the project supervisor. This allowed for all the software functionalities to be demonstrated.

5.5 Calibration

The calibration process discussed in Chapter 4 could not be fully performed based on the failure of the sensors, discussed earlier in this chapter. The calibration process was attempted near the end of the project, once the other components had been assembled. At this time, the sensors only worked in one direction and produced very little change in resistance for small angles of bend. The results of the attempted calibration are shown in Table 5.1 below. The averages of these data points were plotted on a graph in Matlab, shown in Figure 5.2, and using Matlab’s Basic Fitting algorithm, shown in Figure 5.3, a formula for finding an angle from the input voltage was determined. The formula is also shown in this figure.

Ankle Bend	100°	90°	85°	75°	70°	60°
Voltage output (1st trial)	0	0	0.24	0.45	0.45	0.5
Voltage output (2nd trial)	0	0.15	0.21	0.28	0.4	0.6
Voltage output (3rd trial)	0	0.22	0.27	0.3	0.38	0.55
Average	0	0.123	0.24	0.27	0.41	0.55

Table 5.1; Calibration of the Sensors (90° corresponds to neutral position, while 60° corresponds to dorsiflexion). Some angle values were left out due to poor functioning of the sensors and ability to obtain values

Since the IE sensor is relatively straight, ROM in IE did not correspond with the ranges of bend that produced an output greater than 0 V from the circuit once the sensors were broken, so a sample calibration of the IE sensor could not be performed.

It should be noted that this calibration technique performs well in interpolation, but not extrapolation. Since the output of the circuit is limited to between 0 and 5 volts, this problem can be alleviated by simulating maximal input values. Simulated maximum possible ROM values could also be used for this.

Another issue with this technique is the fact that the PD sensor moves in IE and vice versa, so proper calibration of the sensors would need to have a formula that accounts for motion in both planes to determine angle in each of the sensors.

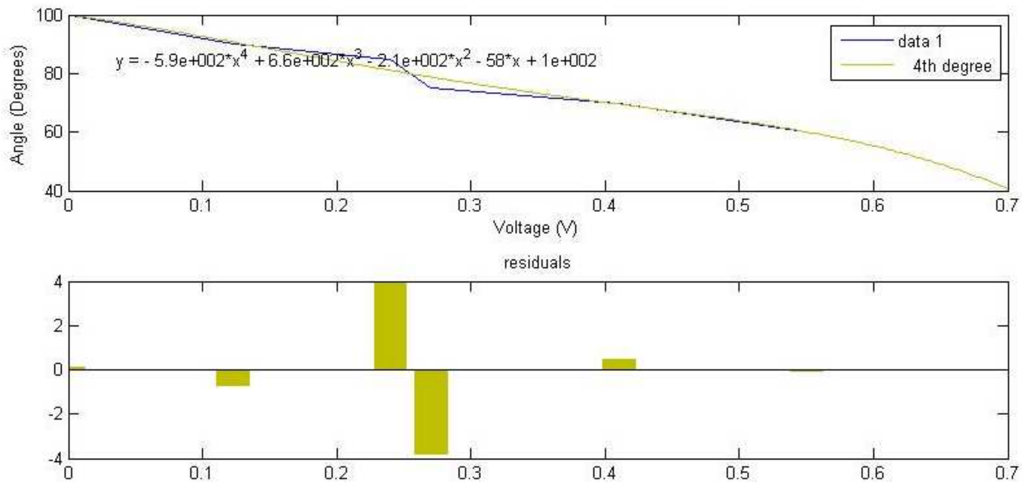

Figure 5.2; Graph of voltage output of circuit vs. bend angle during calibration. Top graph, blue data set is the actually obtained values; yellow data set is the 4^{th} order curve of best fit. Lower graph is the residuals for this curve of best fit

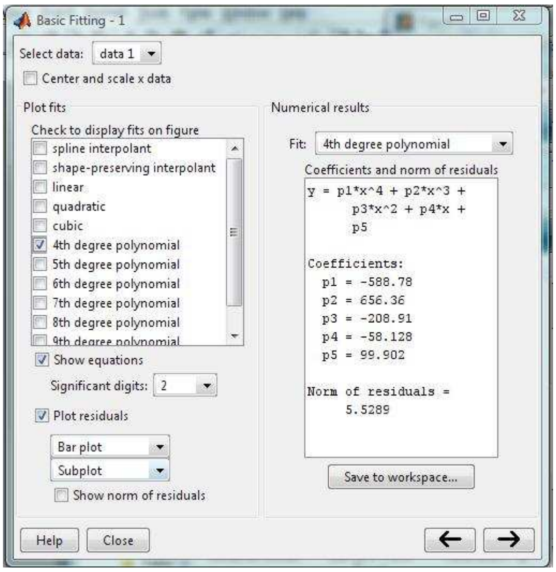

Figure 5.3; Matlab's curve fitting function, 'Basic Fitting'. 4^{th} Order polynomial for found curve of best fit is shown (y)

CHAPTER 6

CONCLUSIONS & RECOMMENDATIONS

6.1 Conclusions

Based on the outcome of this project, it is clear that the bend sensors are the limiting factor in this being a viable, marketable product. Given the research performed and discussions with physiotherapists, this product, if made fully functional and commercially saleable, would be marketable. The theoretical design of the project would allow health care professionals to quantitatively compare ROM results of patients to enhance their rehabilitative process. Given more time and professional resources, the majority of the limitations seen in this project could be overcome. The calibration technique used could be enhanced such that it accounts for most factors involved in ankle motion and a professionally tailored sleeve with better IE sensor placement could be made to properly determine bend in IE, despite the limited motion apparent on the side of the ankle. However, this project, as it stands cannot work without a sensor that can properly determine bend at the ankle. The bend sensors discussed in this project, when they were working, were not precise nor accurate enough to do the job and broke down rather quickly, meaning that even if they had functioned properly, they could not necessarily be used long term. Two-way bend sensors are also not precise enough to be used in this application and their long-term durability is also questionable. In the research done, another sensor implementation that could easily fit inside of a sleeve would likely be too cumbersome or expensive for this type of implementation. However, as the bend sensor technology used is relatively young, it is very possible that this product will improve in time to the point where the project would be able to be implemented successfully.

6.2 Recommendations & Future Direction

As mentioned in the conclusions section, the IE placement of the device and overall design of the ankle sleeve will need to be improved in order to properly determine bend in IE. The calibration process will also need to be improved such that more factors in ankle bend are taken into account to properly determine the bend in each direction from information provided by the bend sensors. In clinical settings, multiple sizes of sleeves would likely be needed to allow for fitting with different patients. Keeping the sensors removable allows for having multiple sleeves without having added cost of multiple sensors.

Provided accurate angular information is available, enhanced analysis of the ankle function would be provided in future implementations, allowing for measurement of torque or stresses at specific points on the body.

Future applications of this project will undoubtedly be wireless. This was an aspect of the project that was considered for implementation if given enough time. A wireless implementation of this device will allow the subject greater movement and mobility to perform their activities or sports actions for a health care professional or trainer. Furthermore, combined with some internal memory, this device could be used during an actual sporting event, to have the data analyzed later for study of the entire ROM throughout the course of an activity. This information could be used for training or fitness studies, or could be used for custom designing of ankle braces or safety equipment specific to an athlete or sport's needs.

APPENDIX A

LABVIEW DESIGN PROCEDURES

Section 4.4 discusses the user interface and functions available to the user in the software implementation of this project. This section discusses the details of how the block diagram, shown in Figure A.1, was designed in LabVIEW.

The DAQ Assistant was used in order to obtain output voltages from the circuitry via the NI ADC in the lab. The DAQ Assistant requires a while loop, which allows for continuous acquisition of data and it is set to continuously sample at a rate of 1000 Hz. The DAQ and most of the programming is placed inside the true condition of a case structure, which is controlled by the stop button that stops the loop from running. There is also a reset button outside the case structure that enables the user to zero all graph data.

On the left side above the leftmost Read from Measurement File and below the DAQ Assistant block are two sets of blocks; the top one controls the scale adjustments for the charts and the bottom one allows for graph scrolling. The bottom set of blocks involves a True constant hooked into the input of each of the graph's "ScrollVis" option. This enables the ability for the user to scroll through the previously acquired data on the graph when the scale is such that not all the data is visible at once on the axis. The top set of blocks dictating the ability to adjust the graph scale involves two switches (one for each graph) labeled "Lock Scale" on the UI whose Boolean outputs are linked into the "XScale.Editable" input of each graph. This allows the user to adjust the time scale on the graphs, meaning they can display data over a defined interval of time. These Booleans are also split off and complemented and converted into an 8-bit integer, which is multiplied by 2 and hooked into the input of the "XScale.ScaleFit" option of each graph. This allows for auto scaling of the graphs when the user does not choose to define a scale.

The data acquired by the DAQ Assistant, which is visible in the top left of the large case structure housing the majority of the code (visible in the block diagram in Figure A.2), is passed through a relay block, which allows the data to be paused when the pause button is selected. The enable of this relay block is hooked into the output of the binary pause button. This is the same button that controls the enable on all of the relay blocks throughout the block diagram, meaning that all data is paused at the same time when this button is pressed. The data is then split into 4 separate signals, the first two being the two live input signals from the device, the second two being the live input signals that would be coming from a second, comparative device. These 4 data streams are passed through formula blocks, which hold the formulas for obtaining an output angle value from an input voltage based on the calibration technique discussed in Section 4.1. The outputs from these four formula blocks are four different sets of angle data.

After being conditioned into the requisite angle values, the data branches out, with the two inputs from the device going to a Save Measurement File block, which allows the user to save the data to file. This block is controlled by a push-button that has an indicator light associated with it, informing the user that data is being recorded. The block is set to open or create a new file with the file name being specified by the user in the File Path box available on the screen. If no file is specified, the Save to Measurement File block has been set to store the data in a default file. The other branch of data, which includes all 4 data streams acquired from the DAQ and conditioned in the formula blocks, are sent to be displayed on the graphs through a series of case structures, with the first set of angle data also being merged with the data stream from a Read from Measurement File block, discussed below. The first case structure controls whether or not the second live incoming data set is shown on the axes along with the first set of live incoming data. The true condition is set to merge the two PD data sets and merge the two IE data sets, one from the live input, the second from the comparative live input, so that both sets of data (for both PD and IE) are displayed on the same axes on the two graphs) when the push-button associated with this case structure (the “Live Comparative Value ON” button) is selected. When this is selected a green indicator light next to the button will also go on. When this button is not selected, the condition is false, in which case the case structure only passes the input from the first device to the outputs, resulting in only the PD and IE data from the first live input being displayed. The switch for this case structure is passed through an AND condition before it reaches the condition of the case structure. The other condition that must be true is that the switch at the top of the UI must be set to “Incoming Data”. This makes it so the user can’t have the primary input being from a file while the comparative input is live from a device.

The output of the above case structure then leads into another case structure, which is controlled by the “Incoming Data – Recorded Data” switch at the top of the UI. When “Incoming Data” is selected, the output of the case structure is the output of the previous case structure discussed in the paragraph above, the live incoming data, which has a corresponding green light to let the user know that live data is being displayed. When the switch is set to “Recorded Data”, the live inputs are not output from the case structure, but rather the outputs of the Read From Measurement File blocks (discussed below) are passed through the case structure and go into the graph allowing previously recorded data to be displayed on the graph.

The error output terminal of the DAQ Assistant is linked into the Relay block following it, whose error output is linked to the Write to Measurement File block. The error output of this block leads into an error handler, which manages any errors that occur from any of these blocks. This error handler is set to not output any dialog box when an error occurs and allows the program to ignore errors from these blocks and continue to run. Errors seen by the Save to Measurement File are most likely, since the user has more control over this block. If the user inputs an incorrect file name or generates some other type of error, the program continues to run and the data can still be saved. In the event of an error that prevents the data from being saved, the user can choose to reset the program using the “End Program” button at the bottom of the UI and initiating the program again.

Read from Measurement File blocks are used to allow the user to open up saved data (data that was previously recorded using the Save to Measurement File blocks, discussed above) for display on the graphs. There are two separate case structures that dictate the behaviour of the Read From Measurement File blocks, one allowing for a comparative input to be displayed on the same axis as the data (either live incoming data or pre-recorded data). The case structure at the bottom right of the block diagram is controlled by two different buttons on the UI, and both binary conditions must be met for a file to run. The first is the “Incoming Data – Recorded Data” switch discussed above, which allows the user to select between displaying live data and displaying pre-recorded data. When the switch is set to “Recorded Data”, the user can then select the “Open File” button (labeled “Recorded Input” on the block diagram), which will open up a dialog box allowing the user to select a file with pre-recorded data to display on the graph. An indicator light will go on informing the user when there is a file opened and being displayed, and the user also has a “Stop” button there, allowing him/her to close the file before it has reached the end. These user options are located at the top of the UI, under “Recorded Data” in the “Display Mode” box. The output of the “Recorded Input” Boolean (controlled by the “Open File” button) is put into an AND block with the “Incoming Data – Recorded Data” switch discussed above, meaning that this will only work when the “Recorded Data” is selected for display by the user. There is also a feedback loop associated with the True condition of this case structure, which is fed by the output of the “EOF?” option on the Read from Measurement File block. This output is a Boolean that indicates when the end of the opened file has been reached and is passed through an OR gate with the output of the “Close File” button inside the case structure. This feedback is complemented and led into an OR gate with the “Recorded Input” Boolean controlling the case structure so that when the end of file is reached or the stop button is pressed, the case structure will be turned off. An issue that was considered when dealing with the Read from Measurement File block was that on first opening a file, the “enable” input of the block must be true in order to open a file (this setting is a result of the settings in the Read from Measurement File properties). However, on subsequent uses of the block, (for example, if the user wants to reopen the file or select a new file) the “Reopen File” Boolean input of the Read from Measurement File block must be selected in order to close the first file. This functionality was made possible by utilizing a register that takes an input and passes it to the next clock cycle. The register was hooked into the aforementioned feedback loop, at the output of the OR condition involving the “EOF?” output and “Close File” buttons, passing this value to the next clock cycle. The output of the register (at the next clock cycle) is hooked into the “Reopen File” input so that when the case has been run once and either the file ends or is closed by the user, a true is passed to the following clock cycle allowing the next opening of the case structure to compensate for having to close the previously opened file. When the case structure is false, a true constant is hooked into the register so that the “Reopen File” input will be true the next time the case structure is set to true. There is also a false constant linked to the enable and a true constant linked to the feedback loop discussed previously. The data output from this case structure is passed through a relay block with its “Enable” input hooked into the pause button discussed earlier. This data is then passed into the case structure controlled by the “Incoming Data – Recorded Data” discussed above, such that when “Recorded Data” is selected, the output from the Read from Measurement File block discussed in this paragraph will be displayed on the graphs.

Another case structure containing a Read from Measurement File block exists, visible in the bottom left corner of the block diagram. The Read from Measurement File block and case structure here are set-up in the same way as the ones discussed in the previous paragraph, but are controlled by different buttons and are not affected by the “Incoming Data – Recorded Data” switch. This case structure is controlled by the “Recorded Value ON” button, located in the “Comparative Values” box. When this button is selected, a file is opened which is displayed on the same axis as the current value, either one from a saved data file opened in the case structure discussed in the paragraph above, or the live input discussed at the beginning of this appendix section. When the “Recorded Value ON” button is selected, a dialog box opens allowing the user to select a file to display on the axes as a comparative data set. When the file is opened, the associated indicator light will be turned on and when either the file ends or the user selects the stop button, this file will close. These functionalities are programmed in the same way as the same functionalities discussed in the paragraph above. The output of this Read from Measurement File block is passed through a Relay block and the data stream is merged with the data stream going to the graphs from either the live input or the pre-recorded data file (depending on the “Incoming Data – Recorded Data” switch).

The “Error Out” output of the Read from Measurement File block discussed above is passed through the associated relay block and into the “Error In (No Error)” input of the other Read from Measurement File block. The “Error Out” of this block is passed through its associated Relay block and into the input of a second error handler. This error handler is set the same as the previously discussed error handler, in that it allows the program to continue to run when an error is detected without displaying a dialog box to the user. The Boolean output of this error block, which is set to True when an error is detected, is passed into the input conditions of the two case structures discussed in the previous two paragraphs. This Boolean is passed through an OR gate with each of the feedback loops used in the case structures such that when an error is detected, the file is closed. When this happens the user can still continue to run the program and open another file.

On the following pages are full-page displays of the UI and block diagram, in Figures A.1 and A.2, respectively. These images have been rotated to best fit onto the page.

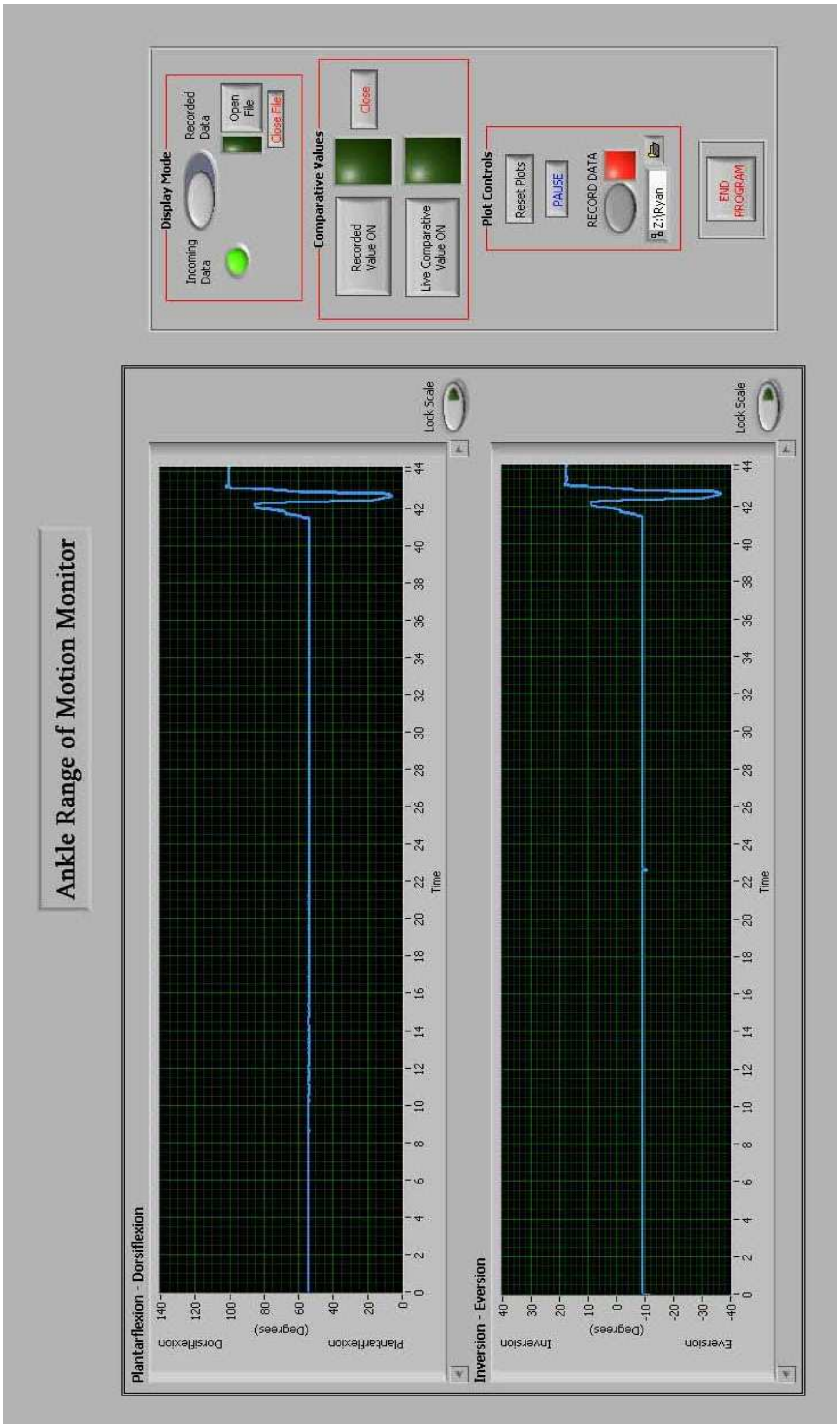

Figure A.1; Software User Interface done on LabVIEW front panel

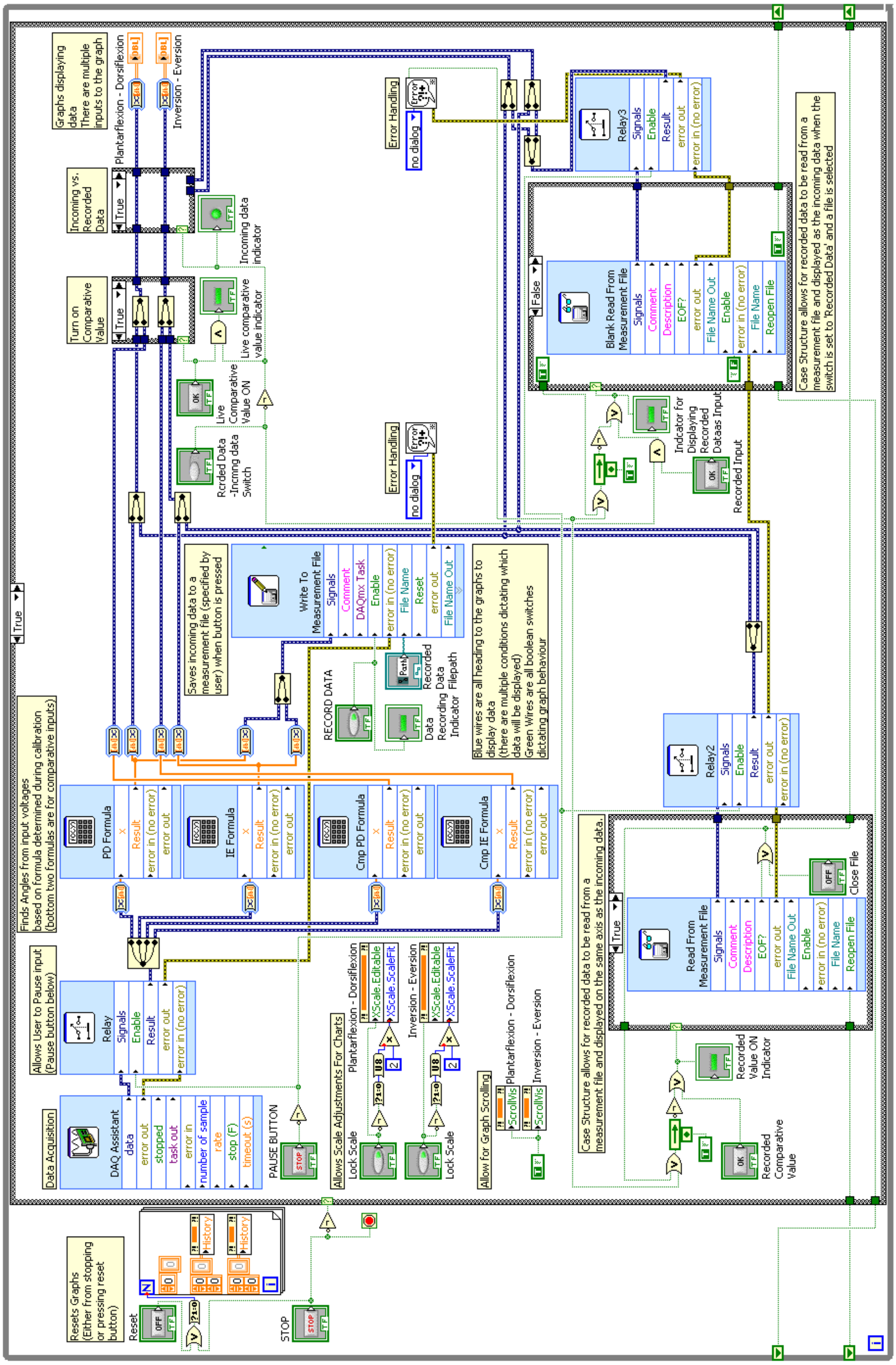

Figure A.2; Block diagram for project software done in LabVIEW

APPENDIX B

DEVICE SCHEMATIC

Figure B.1 contains a block diagram schematic of the hardware involved in the project. Theoretically, the resistance in the bend sensor varies between infinite and zero resistance, resulting in a voltage output between 0 and 5 volts. At infinite resistance, there is no current, while at zero resistance, the sensor is a short circuit so the current is 5 V / 38 KΩ, resulting in a maximum current of 0.132 mA. Power at nominal resistance of the sensor (estimated at 38 K) is equal to voltage squared over resistance, which is $(2.5\ V)^2$ / 38 KΩ = 0.164 mW.

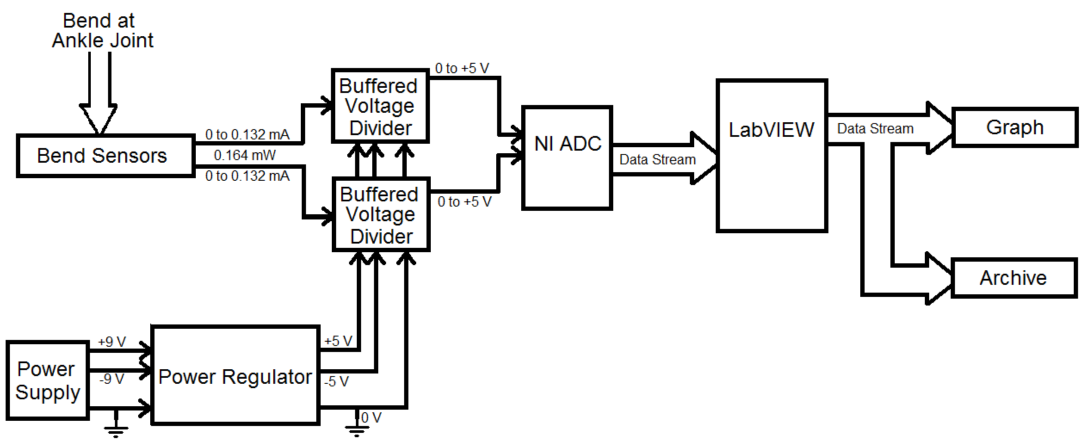

Figure B.1; Block diagram schematic of hardware with estimated voltage, currents, and power in the circuit

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