INSOLE-BASED GAIT ANALYSIS

by

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STATEMENT OF THESIS APPROVAL

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ABSTRACT

Abnormal gait caused by stroke or other pathological reasons can greatly impact the life of an individual. Being able to measure and analyze that gait is often critical for rehabilitation. Motion analysis labs and many current methods of gait analysis are expensive and inaccessible to most individuals. The low cost, wearable, and wireless insole-based gait analysis system in this study provides kinetic measurements of gait by using low cost force sensitive resistors. This thesis describes the design and fabrication of two insoles and their evaluation with 10 control subjects and eight hemiplegic stroke subjects. The first insole used 32 force sensitive resistors and was used to determine the ideal locations of 12 sensors in the second insole. Linear regression was used on training data for each subject testing the second insole to determine ground reaction force, ankle dorsiflexion / plantarflexion moment, knee flexion / extension moment, and knee abduction / adduction moment. Comparison with data collected simultaneously from a clinical motion analysis laboratory demonstrated that the insole results for ground reaction force and ankle moment were highly correlated (all > 0.95) for all subjects, while the two knee moments were less strongly correlated (generally > 0.80). This provides a means of cost effective and efficient healthcare delivery of mobile gait analysis that can be used anywhere from large clinics to an individual's home. The two insoles also provide the means for further testing of force sensitive resistors in different applications.

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CHAPTER 1

INTRODUCTION

This section introduces the background and motivation for the instrumented insole systems with 32 sensors (prototype) and 12 sensors (subsequent design). The research and analysis that was done in preparation for the paper presented in Chapter 2 will then be discussed.

1.1 Background

People generally learn how to walk soon after they turn one year old. They use this skill daily for the rest of their lives. Unfortunately for some, the ability to walk is taken away or made difficult as a result of an accident or illness. Almost 10 million (5.2%) of adults between the ages of 18 and 64 in the United States are classified with a walking disability[1]. Gait analysis, the study of walking, can be an essential part in helping with rehabilitation and recovery. Physical therapists, doctors, surgeons, scientists, and many others can use the results of gait analysis to improve lives. It is becoming more important as orthopedics, rehabilitation, sport medicine and biomechanics fields continue to grow [2].

The most common gait analysis is done in a motion analysis laboratory, which usually contain infrared cameras and force plates. Computer software connected to the laboratory calculates several useful kinetic parameters such as ground reaction force, moments, and center of pressure. Along with kinematic data of the joints and body segments, these parameters can be used to calculate forces and moments on each joint. The result is a computer model of a complete gait cycle. Comparing results from healthy gait and abnormal gait can help identify functional problems and provide recommendations for treatment of those individuals.

A motion analysis laboratory depends on a variety of inputs to complete the calculations and produce the desired outputs. These inputs can be expensive and complex because of the equipment and software needed to run them. The lab also requires trained personnel to run them and are not easily accessible to clinics and hospitals where people would benefit from them.

Alternative methods of gait analysis have been the focus of research for several years. For example, the shoe or insole-based gait analysis system is a cost effective option. This option becomes much more available to people who do not have access to a motion analysis lab. However, the accuracy and complexity of the output parameters decreased because of the fewer and simpler inputs.

This thesis explores the research being done previously on shoe-based kinematic and kinetic gait analysis and presents the research and design done on a new insole-based system. This new design will be much less expensive, use fewer and simpler sensors, and be more accessible while maintaining an acceptable accuracy of the desired outputs.

One of the applications of this insole design is that it will be used in the development and modification of ankle foot orthotics (AFOs) used by those who have had a stroke. The data from the insole can be used in many parts of the rehabilitation process. It can be used to determine the stiffness setting of the AFO for individual patients. Stroke patients were used for testing the insoles in this study to provide results from a real application. Every year, 795,000 people suffer from a stroke. About 610,000 are their first stroke, while about 185,000 are recurrent attacks[3].

1.2 Previous Work

Currently, orthotists have limited resources available to them to analyze gait and obtain numbers, graphs, and values that will allow them to efficiently design or modify orthotics. To determine the usefulness of an ankle foot orthotic (AFO), orthotists use functional tests, but are sometimes left to rely more on eyeballing it and guessing.

Commercially, there are many ways to perform gait analysis. Using a clinical motion analysis laboratory (MAL), with infrared cameras and force plates, is one of the most common ways that gait is monitored. These labs have become a standard of measurement because of their accuracy; however, they are expensive and less accessible. Most systems are complicated to set up and run. They require a trained employee to maintain and run the system. They also place many limits on the actual analysis. They limit the number of measured steps in one trial and introduce inaccuracies due to the subject altering their gait to target the force plate. They do not replicate normal outdoor walking and make it difficult to measure the variability of walking situations. Subjects

also mask or exaggerate their walking problems when they are walking short distances [4].

Some companies, such as Tekscan (Boston, MA) and Pedar novel (Munich, Germany), offer single point force sensors as well as other insole shaped custom pressure sensors that have been used as a more mobile application in gait measurement. The various varieties of sensors can be used like flexible force plates or placed inside the shoe for continuous measurement. Most of these systems allow data to be gathered through USB connection or through wireless communications. These commercial systems opened the door for further research of wearable, insole-based gait measurement; however, they are too expensive to be used in a home or rehab environment.

A. Forner Cordero used the Pedar system to calculate the three components of the ground reaction force. His insoles collected data at 50 Hz and 50% of his trials were invalid due to a missing foot marker, the foot not landing on the force plate, and errors in the insole recordings[5]. This study shows that even with the use of more expensive sensor systems, there are still difficulties getting good trials. As will be seen, the two insoles in the current study had more than a 50% valid testing rate.

Brian T. Smith used the plantar pressure profiles determined using a Tekscan pressure sensitive membrane array to find ideal locations for FSRs in his system that detects gait events in children with cerebral palsy [6]. These examples show that commercial gait analysis systems are useful for validating and designing new and less expensive systems.

One of the first cost effective, insole-based systems was the "footswitch" system created by Hausdorff at Boston University in 1995. His study focused on providing a

simple, inexpensive, and accurate measurement of initial and end foot contact times. He was able to build his system for only $50 using force sensitive resistors (FSRs) [7]. This research led to other studies using the same insole and also using other types of sensors such as gyroscopes, accelerometers and EMGs [8-13]. At the University of Utah, there have been more studies that have focused primarily on using FSRs for mobile gait analysis, activity monitoring and functional electrical stimulation [14-18]. These insole systems typically have prototype costs on the order of a few hundred dollars, or two orders of magnitude lower than commercial systems.

The FSRs provide an inexpensive solution for force sensing; however, they are not as accurate because of their nonlinear nature [19]. This disadvantage can be overcome by using more FSRs as shown in a study at the University of Utah, where a new paradigm for designing medical instrumentation is given. This study proposes that quantity trumps quality in choosing sensors [15]. The current research shows that the use of FSRs can give an accurate measurement of ground reaction force and flexion/extension ankle moment. It even shows that FSRs located in the insole can give information about knee moments and opens the possibilities of many other measurements.

Wireless systems have been researched and have allowed subjects to have more freedom while walking. Without wires, the subject can walk normally. Many papers discuss the use of wireless systems [12, 20-23]. At the University of Utah, Christian Redd investigated a wireless system that also provided a feedback system for the user. This feedback system presented gait parameters in visual, audible and vibrotactile methods [16].

Real time gait analysis and feedback can be helpful in the rehabilitation and training process. Alpha Agape Gopalai used FSRs to study postural control in young adults. He found that the system was able to be used in real time as a qualitative tool for initial, on the spot assessment and as quantitative measure for postacquisition assessment [24].

Machine learning has proved to be an effective approach in training the sensors to more accurately measure desired parameters. Daniel Tik-Pui Fong used the Pedar system with 99 sensors covering the complete plantar area. He then used a stepwise linear regression method to identify the sensors sets that would predict the best triaxial ground reaction force [25]. Using machine learning has enabled systems to measure parameters to which they are not directly connected. As long as there is some correlation, machine learning techniques are able to train the sensor to measure with some percentage of accuracy. Benny Lo used Bayesian analysis to measure subplantar ground reaction force from a pervasive sensor attached to the ear [26]. Meng Chen used a system based on support vector machine regression for learning the relationship between eight FSR values and the corresponding mean pressure acquired by a Pedar insole system [27]. In another study, Meng Chen uses the Hidden Markov Model. However, he focuses only on toe in and toe out gait abnormalities [28].

There have been a few studies using wearable systems to gather data about the knee. Most of these studies use accelerometers, gyroscopes or cloth sensors placed on our around the knee [29]. Pete B. Shull built a system that gave the user feedback in order to reduce the knee adduction moment [30]. T. Liu's system measures three-dimensional lower limb kinematic and kinetic parameters, but requires sensors to be mounted on the

thigh, shanks and feet [23]. The current research used a linear regression method to calculate the knee flexion /extension moment and the knee abduction / adduction moment.

1.3 Contributions

This thesis has resulted in the following contributions:

- A 32 sensor insole (men's size 10) for lab based experimentation and evaluation of sensor locations.

- A 12 sensor flexible insole (adjustable to two basic sizes, covering a range of sizes from women's size 7 to men's size 11) with wireless transmission of data to a laptop.

- Calibration routines used for calibrating the sensors. Sometimes the initial curve fitting function would not adequately fit the curve between the FSR and load cell data in calibration. These routines let the user decide how to manipulate the data to get a better fit. The user could add extra data points in regions where data points were scarce, so that the curve would align better in that region. The user could choose to add extra data points to the beginning of the data that complete the curve. The user could change the order of the curve fitting polynomial. These routines result in choosing the best calibration equation from given trials.

- Analysis software operated in MATLAB. This software allows the user to prepare and crop the data into steps. It runs a linear regression based on training and testing sets defined by the user and then calculates the RMS error and Spearman's correlation coefficient for each of the runs. It allows the user to define the

parameters such as maximum and minimum points in a step or the slope between two points. The software presents and saves the data in a plot for visualization and saves the data in a MATLAB structure to be opened and used later.

1.4 Hypotheses Tested

In order for the 12 sensor insole to be useful compared to the many other products that have been created, it has to have a certain level of accuracy. The MAL is currently accepted as a "gold standard" and can be used to validate other systems. The 12 sensor insole will be run concurrently with the MAL and the results for ground reaction force and anterior-posterior ankle moment will be compared.

Hypothesis 1: The 12 sensor insole can predict ground reaction force with RMS error < 10% and a Spearman's correlation coefficient over 0.95.

Hypothesis 2: The 12 sensor insole can predict anterior-posterior ankle moment with RMS error < 10% and a Spearman's correlation coefficient over 0.95.

1.5 Overview

The following chapters in this thesis have been submitted, or will be prepared for submission for inclusion in conferences and journals.

In Chapter 2, the design process of the 32 and 12 sensor insoles is described and will be organized for a conference publication later this year.

In Chapter 3, a conference publication is presented describing the design and testing of the 32 and 12 sensor insoles. This paper will be submitted for inclusion in the 2012, IEEE Transactions on Biomedical Engineering.

In Chapter 4, the main conclusions of the thesis will be presented, along with recommendations for future work

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CHAPTER 2

DESIGN, FABRICATION, AND TESTING OF INSOLE SYSTEMS

2.1 The 32 Sensor Insole

The first phase of creating an insole-based measuring system was to build a prototype with as many force sensing resistors that could fit inside a Converse shoe. This system used many more sensors than have been used in previous studies and allowed for the exploration of the importance of each sensor in the design based on its position. The results of the first insole were then used in the design of a second insole with fewer sensors. The sensors in the new insole were placed in the ideal locations to get the most accurate measurements.

The FSR 402 Round Force Sensing Resistors from Interlink Electronics (Camarillo, CA) that had been used in previous studies at the University of Utah were chosen because of their effectiveness in prior lab implementations of low cost sensor insoles. These sensors are very cost effective, thin, and robust. They do not require complex circuitry or integration. They are limited only by their nonlinearity in loading.

The first insole was connected to a National Instruments data acquisition module (DAQ) which allowed for 32 analog input signals. Thirty-two sensors were placed inside the footprint of a size 10 men's Converse shoe. The sensors were positioned so that they

covered the entire area of the footprint, and were more dense in important locations such as under the heel, metatarsophalangeal joints, and the great toe as shown in Figure 2.1. Each sensor had leads attached to it, so the sensor had to be oriented to allow the leads to go towards the outside of the insole with enough room to reach the outside of the shoe. The sensors were grouped into four quadrants so that they could be easily identified.
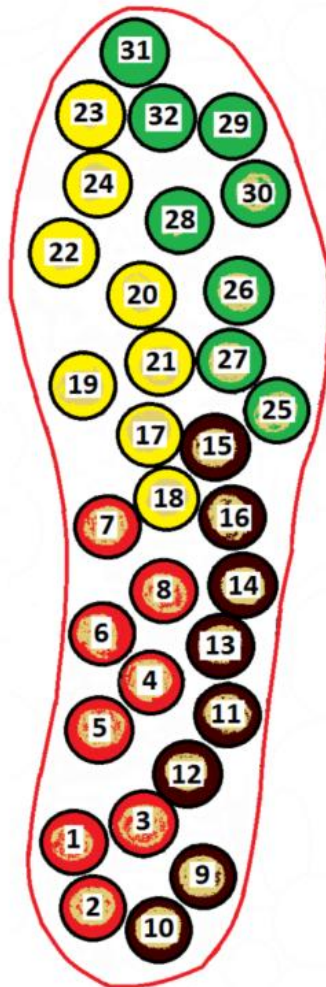

Figure 2.1 Sensor Locations on the 32 Sensor Insole.

Ecoflex 00-30 silicone rubber compound from Smooth-On (Easton, PA) was used to make the main structure of the insoles. A sandwich design was chosen to facilitate the correct placement of the high number of sensors. Previously, sensors were embedded during the process of pouring the silicone molds. As this is a time sensitive process, sensors would need to be placed quickly. A silicone layer approximately 4 mm thick was placed in the bottom of the Converse shoe on top of its normal insole. The sensor locations were traced onto a sheet of contact paper and placed on top of the first silicone layer. A dremel tool was used to create slits around the outer edges of the shoe to allow the leads of the sensors to exit the shoe. The leads of the sensors were then fed through the holes and the sensor was adhered to the contact paper using the adhesive backing. This layer was covered with another thin silicone insole, approximately 3 mm thick.

Ribbon cable was used to connect the sensors to a circuit board that was carried in a pack on the subject's waist. The shoe was divided into four quadrants to keep the wires from getting tangled together. Each quadrant was supplied with a 5 volt supply which was daisy chained through the leads on one side as shown in Figure 2.2. The other leads were grouped with their respective ribbon cable to connect to the board. Hot glue was used around the soldering joints to provide stability and stress relief. A voltage divider was built for each signal using a changeable resistor integrated circuit. This would allow the testing of diverse resistor values in different studies. In this study, the 1000 Ohm resistor was used because of its efficiency in previous studies. However, other resistor values were quickly tested to validate its use. The signal was then sent through a 5 meter ribbon cable to the DAQ from the subject's pack. The complete 32 sensor insole can be seen in Figure 2.3.
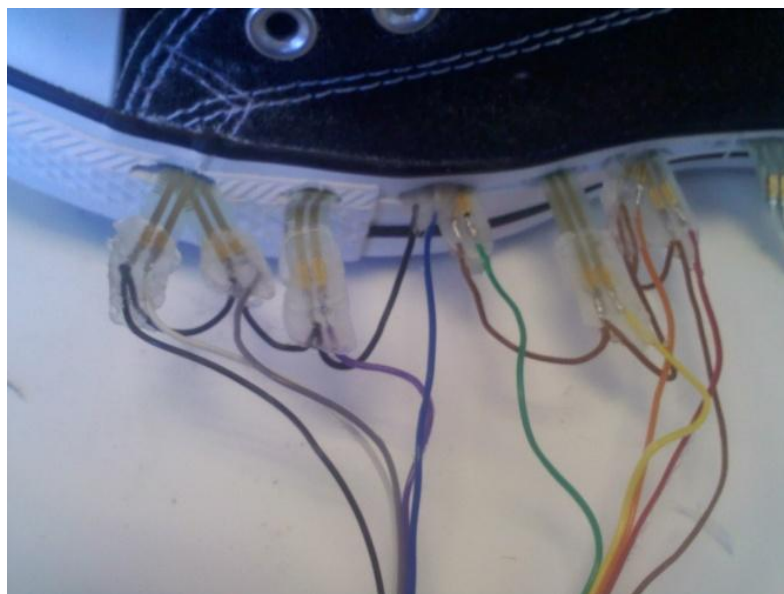
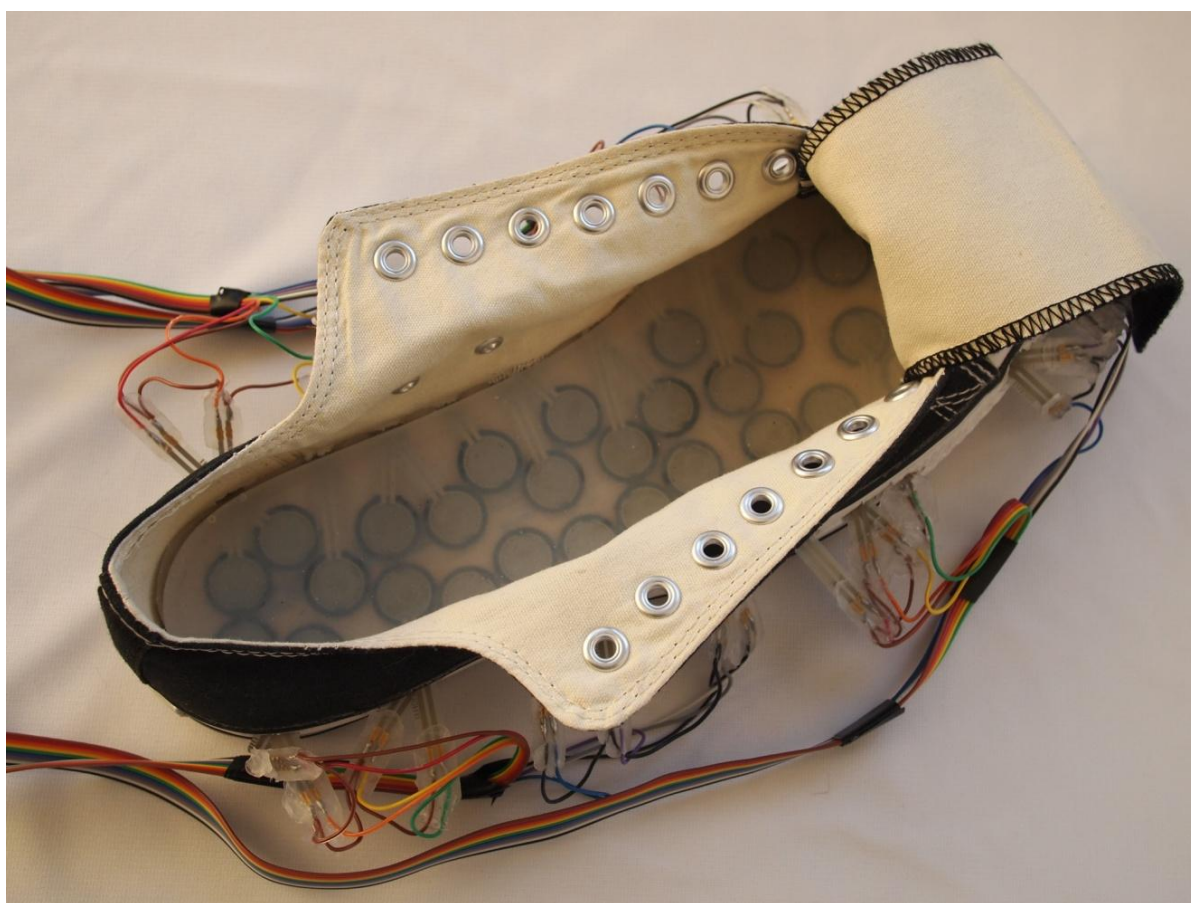Figure 2.2 Wiring the Leads for the 32 Sensor Insole

Figure 2.3 Finished 32 Sensor Insole

The sensors were calibrated using a load cell as shown in Figure 2.4. Force was slowly loaded onto the sensor while both the insole and load cell stored data. The two sets of data were then plotted with the FSR data given in volts on the independent axis and the load cell data given in 1/1000 lbs on the dependent axis. Three pairs of data was collected for each sensor and the best of the three was chosen based on the slope of the curve and the maximum voltage reached. A polynomial equation was fit to each of the curves which could be used to convert voltage readings on the FSRs to a force reading in Newtons. These polynomial equations were created for each of the 32 sensors and entered into the MATLAB code.

Control subjects and stroke patients with a shoe size close to size 10 men's were recruited to test the 32 sensor insole as approved by the University of Utah's Institutional Review Board. Testing took place in the motion analysis laboratory(MAL) in the Department of Physical Therapy of the University of Utah. The PlugInGait marker system was used, which includes 18 markers placed on the lower limbs and tracked

Figure 2.4 Calibrating the Sensors with a Load Cell

by the infrared cameras. Subjects were asked to walk on the force plates with the instrumented shoe on while both the MAL and the insole systems captured the data. The two systems were synchronized by having the subject tap their heel twice on the force plate before walking. This worked well to line up the data, but it was difficult for the stroke patients to tap their foot.

The ground reaction force was calculated by summing the force from each of the sensors. The ankle moment was calculated by multiplying the force of each sensor by its anterior posterior distance to the ankle joint center. The results demonstrated that the 32 sensors performed well for the kinetic calculation of ground reaction force and the kinematic calculation of ankle moment. Comparing plots of the insole data on top of the motion lab data showed that the insole picked up many of the same trends and curves as the lab data although the scaling was not exactly right. The sensors on the insole only picked up a proportion of the weight of the subject because they did not cover the whole area of the footprint. Tests were run with the subject standing statically on the insole and force plate simultaneously, and showed that the sum of all the forces on the insole typically around 50% of that measured by the force plate.

In preparation for the next insole design, analysis was done to isolate which sensors were most valuable in the ground reaction force and ankle moment calculations. First, subsets of the 32 sensors were defined for use in the calculations. These subsets were defined based on their anatomical location of the sensors, their loading trends, and the number of sensors in certain areas of the footprint.

Subsets containing sensors around the heel, metatarsophalangeal joints, and toes were used in different combinations to see which set most closely matched the MAL

data. Each subset group was given a scaling factor which was varied to see how much the subset of sensors affected the calculations. It was found that many sensors could be eliminated because they did not have as much effect as the sensors in those anatomical positions. It was also found that some of the sensors in the arch region of the foot played a bigger role in one of the subjects, where in the other subjects these played a minimal role.

The data from the sensors were analyzed to show how much they were used on average with all of the subjects. The maximum, minimum, and average values of the force on the sensors during gait showed which sensors were getting loaded. This allowed other sensors to be eliminated from a final insole design because they were used minimally when the subject walked. It was found that in some places, the sensors would be saturated to their limit and it was hypothesized that this would affect the resulting force. The sensor location would need to be in a place where it would get loaded, but not get saturated.

The last analysis done on the data was to divide the footprint into seven area sections. These areas were based on anatomical position and locations where sensors were most used or showed other importance. The number of sections (seven) was selected by a limitation of the number of analog inputs available on the desired wireless solution. A sensor was chosen from each of the sections. The ground reaction force was then calculated by multiplying the sensor force by a factor based on the percentage of the total footprint area that its section covered before the sum was taken. The ankle moment was calculated by multiplying those scaled forces by the distance from the ankle joint center to the center of the area section. The area sections are shown in Figure 2.5.
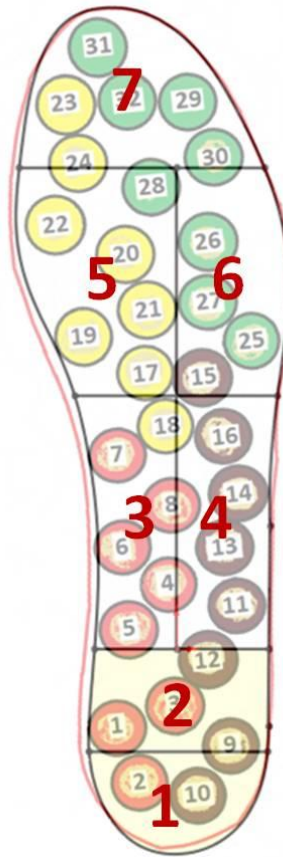
Figure 2.5 Area Sections of the 32 Sensor Insole

2.2 The 12 Sensor Insole

Ultimately, no set of seven sensors to predict force and moment data well were

identified. Thus, for the second insole, it was decided that it would contain 12 sensors

because more sensors on the first insole meant better output data. Two wireless

transmitters would be used to transmit the five extra sensors. One of the restrictions was

that the leads of the sensors needed to remain inside the insole so that it could be used in

the subjects' shoes. To plan the location of these sensors, the information learned from

the 32 sensor insole was taken into consideration as well as a diagram showing the

pressure distribution of the foot. The pressure distribution was taken from force plate data and showed a gradient of where the most force was located. As expected, the high pressure zones were the heel, metatarsophalangeal joint, and under the great toe. For the 12 sensor insole, sensors were placed on the edges of these pressure zones to decrease the possibility of saturating the sensors. Two sensors were placed in the area of the arch to provide complete coverage and also account for subjects with different shaped feet.

The comparison of the 12 new insole sensor locations and the 32 sensor locations are given in Figure 2.6. The figure also shows where those 12 sensors are in relation to the pressure diagram of the foot.
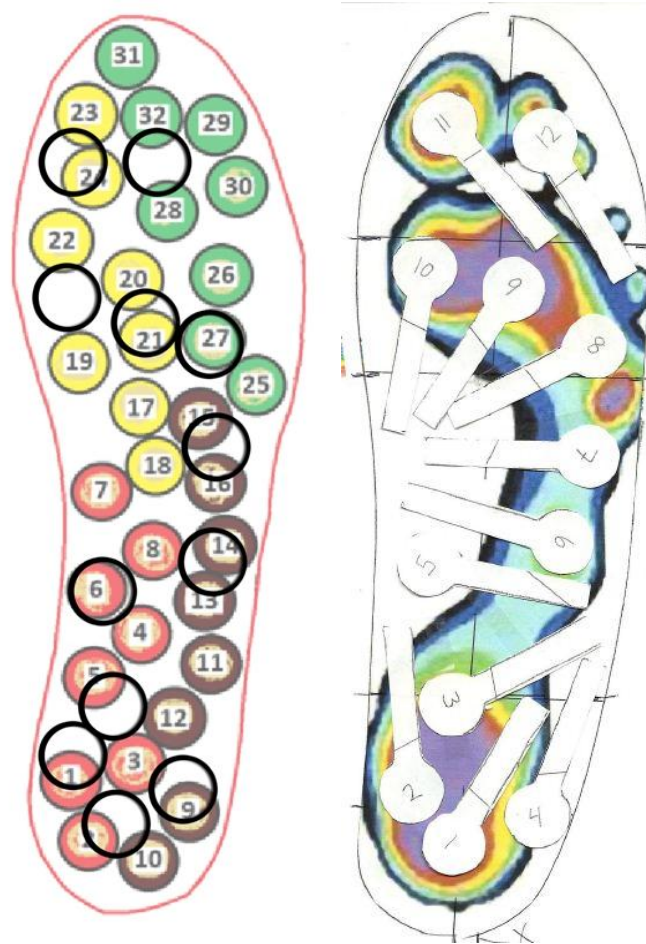

Figure 2.6 Location of the 12 Sensors Based on 32 Sensor Insole and Pressure Diagram.

The sandwich design for the first insole worked well; however, over time the different layers shifted slightly on top of each other. Because of the prototype nature of the design, the insole needed to be nonpermanent to allow troubleshooting and modification of sensors that might stop working. This idea would carry on to the 12 sensor insole, but in the future a more permanent insole structure would need to be made that would fix the sensors in the correct position without shifting.

One of the changes in the design of the second insole from the first was the use of a flexible circuit board. This would allow for the insole to be much thinner than previous insoles and take out the complexity of planning routes for individual wires. The flexible circuit board would allow for faster manufacture of the insole. In this design, the ribbon cable would be attached to the medial side of the foot in the arch area. This would be the area least likely to receive stress from the loading of the foot and would allow room for the wires to be soldered onto the flexible circuit. The flexible circuit was designed using Cadsoft Eagle PCB Design Software. Two different sizes were made to accommodate a more diverse testing population. The large size was designed towards a men's size 10 shoe while the small size was a women's size 7.Another change was that each silicone layer was made to be 2 mm thick to result in an even thinner final insole.

The leads of the sensors were first soldered and then the sensors were adhered to the flexible circuit. The silicone layers were then placed on either side it. On the outside, contact paper was used to provide stability for the insole and electrical tape held it all together. Because there was a lot of stress on the wire connection to the flex circuit, the wires were soldered, bent over themselves and then hot glued to keep the connection. This insole could then fit into the subject's shoe on top of their regular insole.

The transmitter box was designed to contain the voltage divider circuit as well as the wireless transmitters, TI EZ430-RF2500T. This circuit was powered with 3.3 V and ran off two AA Batteries. A voltage regulator maintained the 3.3 Volts. A wireless receiver, also a TI EZ430-RF2500T, was connected through USB to a computer where the data was collected via a custom MATLAB gui. The transmitter box can be seen in Figure 2.7.
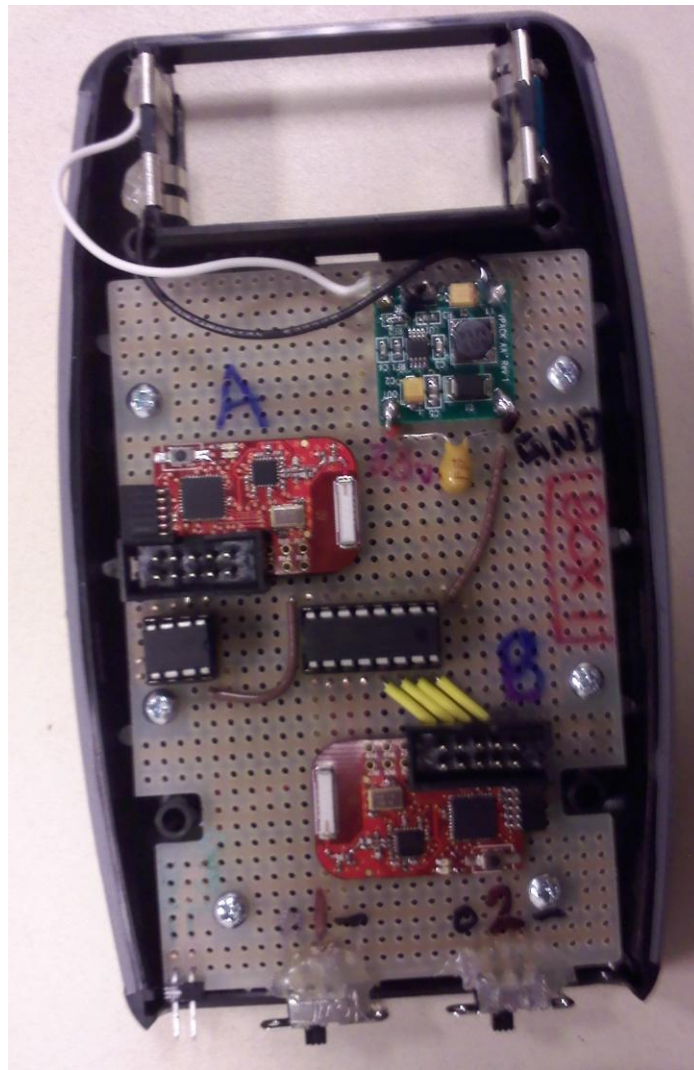

Figure 2.7 Wireless Transmitter Box Circuit Board for the 12 Sensor Insole.

A new set of control subjects and stroke patients were recruited to test the new insole. Again, the testing was done in the MAL at the University of Utah.. The goal was to collect steps from about 10 walking trials. For the stroke patients, this could result in about 20 collected steps if they stepped on each of the two forceplates. For the control subjects, it also could result in 20 steps because for each trial, they walked over the force plates, turned around, and walked back across the force plates. This time, the two systems were synchronized using the forceplate, and using an FSR that was connected externally from the insole. This FSR plugged into the top of the transmitter box and could be either left there and held by a control subject as they walked, or unplugged by an assistant before a stroke patient walked. The systems were synchronized by tapping four or five times on the FSR as it rested on the force plate.

In order to compare the insole data to the MAL data, each step was extracted from the data. Initially, the data was analyzed as in the 32 sensor insole, with the sum of the sensor outputs used for ground reaction force and the product fo the sensor locations multiplied by their forces to calculate ankle moment.

The root means square (RMS) error was calculated between the two sets of data. For the ground reaction force, the RMS error was divided by the maximum value of the MAL data to obtain the percent RMS error. For the ankle moment, the RMS error was divided by the difference between the maximum and minimum MAL moment data to obtain the percentage. The RMS error indicates how much error is between two sets of data. The original target specifications were for an RMS error below 5% for ankle moment and below 10% for ground reaction force. However, after the initial 32 sensor insole test, it was clear that the ankle moment was similarly difficult to determine as the

force, and the target specifications were revised to obtain an RMS error below 10% for both ankle and moment.

The Spearman's correlation coefficient was also calculated between the two sets of data for each step. Spearman's correlation coefficient indicates how similar the shape of two sets of data are, and ignores any offset. The target specification was to achieve a Spearman's correlation above 0.9 for both force and ankle moment.

Using this initial approach, the features in the insole data matched some of the features of the MAL data, particularly with control subjects. However, it was apparent that these results were insufficient on their own to meet the target specifications.

Next, a linear regression technique was implemented. The theory of this technique is to take the inputs to the system (the 12 insole force values) and match them up to the desired outputs (the MAL ground reaction force, ankle moment, or other parameters). The equation: $B_{(1-12)}F_{insole} + B_{13} = F_{motion\ lab}$ can be used as a relation between the inputs and outputs. F_{insole} is an array of each of the 12 input sensors, $F_{motionlab}$ is the output ground reaction force for the MAL, B_{1-12} are 12 scaling coefficients, each one matching up with a sensor, and B_{13} is a shifting coefficient.

The data gathered from the steps were separated into groups of training data and testing data. The training steps were lumped together and fed through the linear regression formula, solving for the coefficients. The testing data was then put into the equation with the coefficients, solving for the ground reaction force. The details of this analysis and the corresponding results are discussed in the paper presented in Chapter 2.

CHAPTER 3

KINETIC GAIT ANALYSIS USING A LOW COST INSOLE

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Kinetic Gait Analysis Using a Low-Cost Insole

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Abstract—Abnormal gait caused by stroke or other pathological reasons can greatly impact the life of an individual. Being able to measure and analyze that gait is often critical for rehabilitation. Motion analysis labs and many current methods of gait analysis are expensive and inaccessible to most individuals. The low-cost, wearable, and wireless insole-based gait analysis system in this study provides kinetic measurements of gait by using low-cost force sensitive resistors. This paper describes the design and fabrication of the insole and its evaluation in six control subjects and four hemiplegic stroke subjects. Linear regression was used on training data for each subject to determine ground reaction force, ankle dorsiflexion / plantarflexion moment, knee flexion / extension moment, and knee abduction / adduction moment. Comparison with data collected simultaneously from a clinical motion analysis laboratory demonstrated that the insole results for ground reaction force and ankle moment were highly correlated (all > 0.95) for all subjects, while the two knee moments were less strongly correlated (generally > 0.80). This provides a means of cost-effective and efficient healthcare delivery of mobile gait analysis that can be used anywhere from large clinics to an individual's home.

Index Terms—gait analysis, ankle moment, ground reaction force, force sensitive resistor, insole, orthotic

I. INTRODUCTION

NEARLY 10 million (5.2%) of adults in the United States between the ages of 18 and 64 are classified with a walking disability [1]. Stroke is one of the many causes of these disabilities. Every year, 610,000 people experience their first stroke, and 185,000 more have a recurrent stroke attack. In other words, every 40 seconds, someone in the United States suffers a stroke [2]. Ankle-foot orthotics are designed to assist stroke patients in walking. Measurement and analysis of ground reaction force and ankle moment in gait can provide important information regarding the rehabilitation process of stroke patients. Ankle moment has a significant influence on trunk acceleration, propulsion, and balance while walking [3].

There are many ways to perform gait analysis. Using a clinical motion analysis laboratory with infrared cameras and force plates is one of the most common ways that gait is monitored. These labs have become a standard of measurement because of their accuracy; however, they are expensive and not easily accessible. The direct kinetic analysis is limited to only those steps that land on a force plate. Inaccuracies may be introduced if the subjects alter their gait to target the force plate. Instrumented treadmills are available but treadmill gait differs from normal gait in that it does not require torso motion. Some companies (e.g. Tekscan (Boston, MA), novel (Munich, Germany), etc.) offer single point force sensors as well as other insole-shaped custom pressure sensors that have been used as a more mobile application in gait measurement. However, these commercial systems are too expensive (on the order of $10K or more) to be used in a home or rehab environment.

In 1995, Hausdorff et al. demonstrated a "footswitch" system, with a focus on providing a simple, inexpensive, and accurate measurement of initial and end foot contact times [4]. This low-cost system has been used in a number of studies, particularly investigating stance times (e.g. [5]). It inspired many groups (e.g. [6-9]), including our own (e.g. [10-13]) to build instrumented insoles using inexpensive force sensitive resistors (FSRs) for a variety of applications, including mobile gait analysis, activity monitoring, and functional electrical stimulation. These insole systems typically have prototype costs on the order of a few hundred dollars, or two orders of magnitude lower than commercial systems. However, the low-cost of the FSR sensors is accompanied by a highly non-linear response [14] that makes finding parameters such as ground reaction force challenging. Other groups have investigated alternative means, such as Tomizuka *et al.* with air coils to measure pressure in the shoe [15], or textile sensors [16-18]. Accelerometers and gyroscopes (alone or in combination with in-shoe senors) have been used for applications ranging from activity monitoring, fall detection or gait event detection to joint center estimation (e.g. [10, 19-25]).

Other types of insole sensors may offer advantages of linearity and reliability not achievable by force sensing resistors (FSRs). However, the low cost and slim form factor of inexpensive FSRs can be exploited with calibration [26], for instance to estimate center of pressure with mean RMS error of 11 mm in the medial-lateral direction and 17 mm in the anterior-posterior direction [13].

Realizing ground reaction force and other kinetic measures outside of a motion analysis lab may require machine learning techniques to capitalize on the many small, local

measurements inherent to in-shoe measurement [27-30]. Chan *et al.* used a 99 sensor pedar (novel, Munich, Germany) insole in combination with linear regression analysis to determine ground reaction force, with 5% RMS error compared to a motion laboratory [30]. Aminian *et al.* added an inertial measurement unit to the pedar sensor, and also used non-linear techniques to achieve 4% RMS error in ground reaction force [28].

This paper presents the design and validation of an instrumented insole with twelve FSRs mounted to a flexible circuit insole. Each FSR has been individually calibrated with a load-cell. Control subjects with no known gait abnormalities and hemiplegic stroke patients who regularly used an ankle-foot orthotic were evaluated, with simultaneous data collection in a clinical motion lab. Training data from the insole and corresponding motion lab results were used in combination with linear regression to develop models of each subject's gait. The training data from the insole was then used to predict the ground reaction force, ankle dorsiflexion / plantarflexion moment, and knee moments in flexion / extension and abduction / adduction.

II. METHODS

A. Hardware

An insole with twelve sensors was designed to fit into the subject's own shoes, as shown in Fig. 1). The sensors were FSRs, model 402 (Interlink Electronics, Camarillo, CA), and are inexpensive, highly durable, and simple to implement with a voltage divider. FSRs decrease in resistance when an increase of pressure is applied to its 1.27 cm (0.5 inch) diameter surface. The FSRs were calibrated using an iLoad Mini 50 pound miniature load cell (Loadstar Sensors, Inc., Fremont, CA). To determine an appropriate layout of the FSRs on the insole, a 32 sensor insole was built and tested. This 32 sensor insole showed that sensors placed in areas under the heel, metatarsal-phalangeal joints, great toe, and the lateral arch provided the most influence in the analysis [31]. These areas correspond to the known biomechanical areas of loading in a typical plantar pressure distribution.

A flexible circuit board was designed and fabricated to facilitate rapid and repeatable construction. Longer copper pads for several of the heel and toe sensors allowed for the implementation in two sizes of insole by allowing the position of these sensors to be altered along the length of the insole. The small size was used for sizes men's 5-9 (women's 7-11) and the large size for men's 10-13 (women's 12-14).

The sensors were first soldered to the flexible circuit, and then attached with the adhesive on the back of the FSR. A sandwich was constructed by placing the flexible circuit between two slim (2 mm (0.078 inch) thick) silicone insoles that were custom made from a cured silicone rubber (Ecoflex 0030, Smooth-On Inc., Easton, PA). The insole was reinforced with contact paper on each side, and adhesive was used to hold the sandwich together.

A ribbon cable connected the flexible circuit to a box containing the microcontroller, wireless transmitter, and two AA batteries. The box was placed on the medial side of the subject's lower leg. The wires were soldered to the flexible circuit and then strengthened using a thin layer of hot glue. The prototype costs of the entire insole and accompanying electronics were under $150.

Fig. 1. The kinetic insole, including 12 force sensitive resistors mounted on flexible circuit board. Heel and toe sensors have longer copper pads to enable multiple sizes of insoles.

B. Study Procedure

The studies were approved by the University of Utah's Institutional Review Board, and all subjects provided informed consent. Testing was carried out in the Motion Analysis Lab (MAL) at the Department of Physical Therapy, with simultaneous collection from the insole and the motion analysis equipment. The motion analysis equipment includes 10 infrared motion capture cameras (Vicon, Oxford, UK) and 2 OR6 series multi-axis force plates (Advanced Mechanical Technology, Inc. Watertown, MA). The standard lower body plug-in-gait marker set was used. Both marker and analog force plate data were collected at 200 Hz. Vicon Nexus software was used for marker labeling, modeling, analysis, and export of marker and analog data from the MAL. The insole data was transmitted wirelessly to a receiver plugged into a laptop, with a frequency of 118 Hz. Data from the two systems were synchronized via a concurrent tap on a force sensitive resistor and the force plate.

Subjects were asked to walk on the force plates at a self-selected pace. Individual steps were retained for analysis if the footfall was completely on only one force plate. Each subject walked until a minimum of ten steps were obtained on the force plate (if able, some subjects performed more walks). Steps were later excluded (resulting in fewer than ten for some subjects) if one or more markers were occluded and affected the MAL data output.

C. Subjects

Control subjects with no known gait abnormalities were recruited from the general University of Utah population. There were six control subjects (five male, age: 23.3±3 years, mass: 69.2±6.2 kg, height: 1.7±0.1 m, shoe sizes: 10, 10.5, 11.5 men's and 7 women's), and all wore shorts during testing to allow markers for the cameras to be placed directly on the

skin of the legs. Stroke subjects were recruited from the University Health and Wellness Center. There were four stroke patients (two male, age: 55.8±10.9 years, mass: 105.4±14.6 kg, height: 1.8±0.2 m, shoe sizes: 10, 13 men's and 7.5, 8 women's). Characteristics of the participants are detailed in Table 1. The stroke subjects were permitted to use the attire they were wearing in the clinic that day for their comfort, which resulted in all stroke subjects wearing pants; markers were adhered to the outside of the pants. Only one stroke subject (H) was able to walk with and without his ankle-foot orthotic. When using an orthotic, the insole was inserted between the orthotic and the shoe insole.

TABLE 1
CHARACTERISTICS OF PARTICIPANTS

	ID	Shoe Size	Insole Size	Orthotic	Total steps	Test group sizes
Control Subjects	A	10.5 M	L	No	8	3, 3, 2
	B	11.5 M	L	No	50	17,17,16
	C	10 M	L	No	39	13,13,13
	D	10 M	L	No	20	7, 7, 6
	E	10M	L	No	23	8, 8, 7
	F	7 W	S	No	24	8, 8, 8
Stroke Patients	G	8 W	S	Yes	6	2, 2, 2
	H	10 M	L	Yes	8	3, 3, 2
	H	10 M	L	No	7	3, 2, 2
	I	7.5 W	S	Yes	10	4, 3, 3
	J	13 M	L	Yes	17	6, 6, 5

D.   Analysis

Data from the MAL and the insole were exported for subsequent analysis in MATLAB (Mathworks, Inc., Natick, MA) and Excel 2010 (Microsoft, Inc., Redmond, WA). Forces and moments from the MAL were normalized by dividing by the bodyweight in N and in kg, respectively. A spline fit was used to resample the MAL data at the time points corresponding to the insole data.

A least squares linear regression was used to weight the sensor forces to match the motion lab data for normalized ground reaction force and moments. Three moments were analyzed: ankle dorsiflexion / plantarflexion (d/p) moment, knee flexion / extension (fl/ex) moment, and knee abduction / adduction (abd/add) moment. The linear regression model was implemented in MATLAB to determine the weighting coefficients and a shifting constant. For example, the ground reaction force was modeled by:

$$\sum_{i=1}^{12}(B_i F_{insole}) + B_{13} = F_{MAL} \tag{1}$$

where B_{1-12} were the twelve coefficient weights for the 12 sensors, B_{13} was the shifting constant, F_{insole} was the vector of input data from the 12 sensors, and F_{MAL} was the MAL ground reaction force. For each subject, the data was cropped into steps aligned at heel strike and ending at toe off. The steps for each subject were divided into three groups (see Table 1). Two of the groups were used as a training set while the third group of steps was used as a testing set. Each of the three groups

were used as the test set while the other two were used as training sets until all of the steps were tested.

The insole data for all of the steps in the training group were concatenated and used as inputs to four linear regression models. The MAL data for the ground reaction force and the three moments were each used as an output to one of the models. This allowed the training group to solve for the coefficients and shifting constant. Next, the steps from the testing group were used with the coefficients and shifting constants to obtain the ground reaction force and moments measured by the insole. Essentially, for each subject, the sensor values were used to train a model to produce the desired output.

To validate the insole measurements, the results from the steps in the testing groups were compared to the corresponding results from the motion lab. The root mean square (RMS) value and the Spearman's rank correlation were calculated to evaluate how well the shape from the insole matched the MAL for each force and moment variable. The percent RMS value (%RMS) was determined by:

$$\%RMS = \frac{RMS}{MAL_{max} - MAL_{min}} \tag{2}$$

where MAL_{max} and MAL_{min} are the maximum and minimum values, respectively, of the MAL value for that step. In addition, the maximum value of the ground reaction force, the ankle d/p moment (i.e. the maximum plantar flexor moment), the knee fl/ex moment (i.e. the maximum extensor moment), and the knee abd/add moment (i.e. the maximum adductor moment) were extracted from the MAL and insole data for each step to evaluate how well the insole found these changes. The % Errors of these maximum values were calculated by

$$\%Error = \frac{MAL - Insole}{MAL} \tag{3}$$

where *MAL* and *Insole* represent the maximum values of each parameter for that step.

III.   RESULTS

A total of 164 steps were obtained for the six control subjects, and 50 steps for the four stroke patients, one of whom repeated the test without use of his orthotic. The results (%RMS error and the Spearman's correlation coefficient) of the correlation of the insole shape to the MAL shape are summarized in Table 2. The mean (± standard deviation) values are given for each individual subject, with overall mean (± std. dev.) for the group of control subjects (A-F) and the group of stroke patients (G-J). Stroke subject H was able to walk with and without an orthotic, and both results are presented.

The %RMS error gives an average error for the given set of data points, indicating how well the insole curve corresponds to the MAL curve at each time point. The %RMS error only indicates whether the data compared have similar values, resulting in a larger %RMS error if the shape of the curves are similar but shifted horizontally. The Spearman's correlation

TABLE 2
PERCENT RMS ERROR AND SPEARMAN'S CORRELATION COEFFICIENT FOR FORCE AND MOMENT CURVES OF THE INSOLE COMPARED TO THE MAL

	ID	Ground Reaction Force		Ankle Dorsi/Plantar Moment		Knee Flex/Ext Moment		Knee Abd/Add Moment	
		%RMS Error	Spearman's Corr. Coeff.	%RMS Error	Spearman's Corr. Coeff.	%RMS Error	Spearman's Corr. Coeff.	%RMS Error	Spearman's Corr. Coeff.
Control Subjects	A	4.3 ± 2.0	0.98 ± 0.01	4.5 ± 2.3	0.99 ± 0.00	10.4 ± 5.6	0.94 ± 0.02	13.9 ± 3.1	0.89 ± 0.05
	B	5.7 ± 1.8	0.96 ± 0.02	7.1 ± 3.4	0.98 ± 0.02	10.2 ± 3.2	0.91 ± 0.07	a	a
	C	5.1 ± 1.2	0.97 ± 0.01	5.5 ± 1.7	0.96 ± 0.06	13.8 ± 6.3	0.91 ± 0.05	15.4 ± 7.5	0.82 ± 0.10
	D	7.8 ± 3.0	0.95 ± 0.04	7.5 ± 3.4	0.97 ± 0.01	13.9 ± 5.2	0.85 ± 0.07	18.2 ± 6.5	0.84 ± 0.09
	E	4.4 ± 0.8	0.98 ± 0.01	4.1 ± 0.8	0.98 ± 0.01	7.7 ± 3.6	0.95 ± 0.02	18.4 ± 3.4	0.85 ± 0.05
	F	4.5 ± 1.5	0.97 ± 0.01	4.6 ± 1.1	0.98 ± 0.01	9.1 ± 3.3	0.93 ± 0.03	15.4 ± 2.3	0.86 ± 0.03
	Overall	**5.4 ± 2.0**	**0.97 ± 0.02**	**5.9 ± 2.7**	**0.97 ± 0.03**	**10.7 ± 5.3**	**0.89 ± 0.17**	**16.4 ± 5.7**	**0.84 ± 0.08**
Stroke Patients	G	9.0 ± 3.4	0.97 ± 0.01	a	a	24.7 ± 9.5	0.81 ± 0.05	19.1 ± 13.7	0.61 ± 0.4
	H_orthotic	4.1 ± 1.0	0.97 ± 0.02	6.6 ± 2.6	0.95 ± 0.03	11.3 ± 3	0.93 ± 0.03	14.5 ± 4.0	0.89 ± 0.07
	H_no orthotic	6.1 ± 3.3	0.96 ± 0.03	10.6 ± 5.1	0.94 ± 0.03	10.8 ± 3.5	0.90 ± 0.05	20.3 ± 7.7	0.84 ± 0.06
	I	8.2 ± 4.9	0.95 ± 0.03	14.9 ± 12.2	0.92 ± 0.06	14.1 ± 6.3	0.89 ± 0.06	22.9 ± 12.5	0.79 ± 0.14
	J	5.9 ± 2.9	0.97 ± 0.03	8.6 ± 5.5	0.96 ± 0.05	12.1 ± 5.4	0.94 ± 0.06	13.5 ± 7.4	0.88 ± 0.08
	Overall	**6.4 ± 3.5**	**0.96 ± 0.02**	**9.8 ± 7.2**	**0.95 ± 0.05**	**13.7 ± 7.0**	**0.91 ± 0.07**	**17.1 ± 9.4**	**0.82 ± 0.18**

a The corresponding moments were not available from the processed MAL data for these subjects.

oefficient is a unitless number between 0 and 1 that compares the shapes of the two curves instead of the actual values. A Spearman's correlation coefficient close to 1 indicates that the shapes are highly similar, but does not indicate whether the curves are shifted horizontally.

The percent errors of the maximum forces and moments and are summarized in Table 3. These provide an indication of how well the insole found the total change in each parameter.

Representative plots showing the ground reaction force and the three moments from a single step are shown for a control subject in Fig. 2, and for a stroke patient in Fig. 3.

TABLE 3
PERCENT ERROR OF MAXIMUM FORCES AND MOMENTS FOR THE INSOLE AS COMPARED TO THE MAL MEASUREMENT

	ID	Ground Reac. Force	Ankle D/P Moment	Knee Fl/Ex Moment	Knee Abd/Add Mom
Control Subjects	A	-1.5 ± 2.6	-1.4 ± 6.1	-0.4 ± 17.9	4.9 ± 24.2
	B	-1.9 ± 6.6	-1.5 ± 13.2	5.2 ± 11.5	a
	C	1.1 ± 3.6	0.3 ± 6.8	-11.2 ± 28.8	8.8 ± 16.1
	D	-2.7 ± 6.9	-5.5 ± 14.8	0.1 ± 23.7	5.4 ± 49.3
	E	-2.9 ± 4.7	-1.8 ± 3.5	0.8 ± 12.2	-3.9 ± 44.2
	F	-0.1 ± 2.5	0.8 ± 5.2	-2.4 ± 13.8	2.9 ± 21.2
Overall		**-1.1 ± 5.3**	**-1.3 ± 9.9**	**-1.5 ± 19.8**	**4.2 ± 32.0**
Stroke Patients	G	3.1 ± 9.5	a	-34.7 ± 32.4	15.5 ± 19.5
	H	2.5 ± 2.5	-2.2 ± 9.4	10.5 ± 25.9	9.5 ± 16.5
	H	2.8 ± 6.1	0.4 ± 11.6	8.0 ± 14.8	-7.2 ± 29.7
	I	4.9 ± 11.2	-6.1 ± 26.1	19.0 ± 18.6	-3.9 ± 41.6
	J	1.4 ± 8.2	-3.7 ± 11	3.0 ± 20.3	9.1 ± 21.4
Overall		**2.6 ± 7.8**	**-3.1 ± 14.6**	**2.9± 26.5**	**5.2 ± 26.4**

a The corresponding moments were not available from the processed MAL data for these subjects.

IV. DISCUSSION

Overall, the test results from inexpensive insole provided quantitative data that was comparable to the results obtained in a clinical motion analysis lab, the MAL. The best performance was achieved with the ground reaction force, which is not surprising since the sensors used in the insole measure force. For both control and stroke subjects, the mean plus one standard deviation of the RMS error were under 10%, which is close to that achieved by others with the pedar insole [28, 30], which is two orders of magnitude more expensive. The mean Spearman's correlation coefficient was above 0.95 for all subjects, indicating that the overall shape of the curves were highly similar. Figs. 2a and 3a, which demonstrate that the shape is similar, with minor deviations at the two peaks in the ground reaction force. In addition, the mean error plus one standard deviation of the maximum ground reaction force (Table 3) were under 10% for both control and stroke subjects.

The ankle d/p moment performed similarly well in the control subjects. In stroke subjects, the shape (measured by the correlation coefficient) was similar, but the %RMS error was higher. The lower magnitude ankle moments generated by the stroke subjects result in a smaller denominator and a larger numerical value of percent error. The stroke subjects typically generated an ankle moment that was about half that generated by control subjects, as seen by comparing Figs. 2b and 3b.

Simply comparing the numerical results indicates that the two knee moments (fl/ex and abd/add) were less correlated with the MAL than the ground reaction force and the ankle d/p moment. However, the mean plus one standard deviation of the RMS error were generally under 20-25%, with correlation coefficients generally over 0.80, which is remarkable given that the knee moments were determined without any knowledge of the position (and thus moment arm) of the knee.

Tables (hidden text)
T3.2 %RMS Error and Spearman's for Force and Moment Comparison

T3.3 Percent Error for Compared Maximum Forces and Moments.

Fig. 2. A representative plot from a control subject showing motion analysis lab (MAL) and trained insole results for a) ground reaction force, b) ankle d/p moment, c) knee fl/ex moment and d) knee add/abd moment.
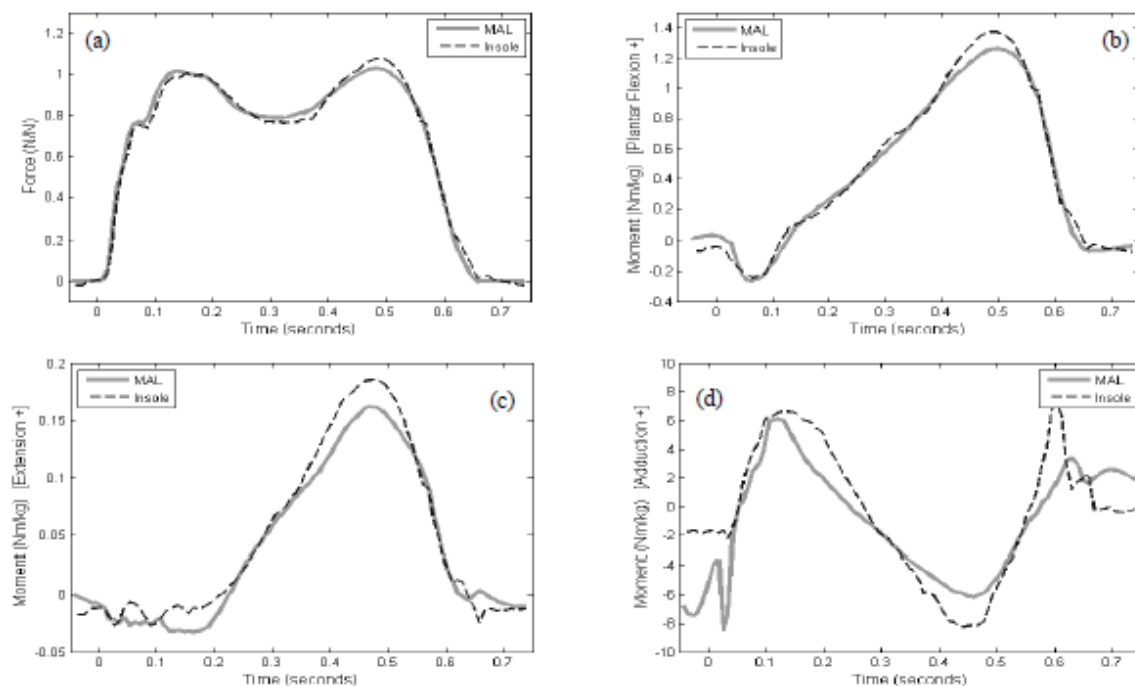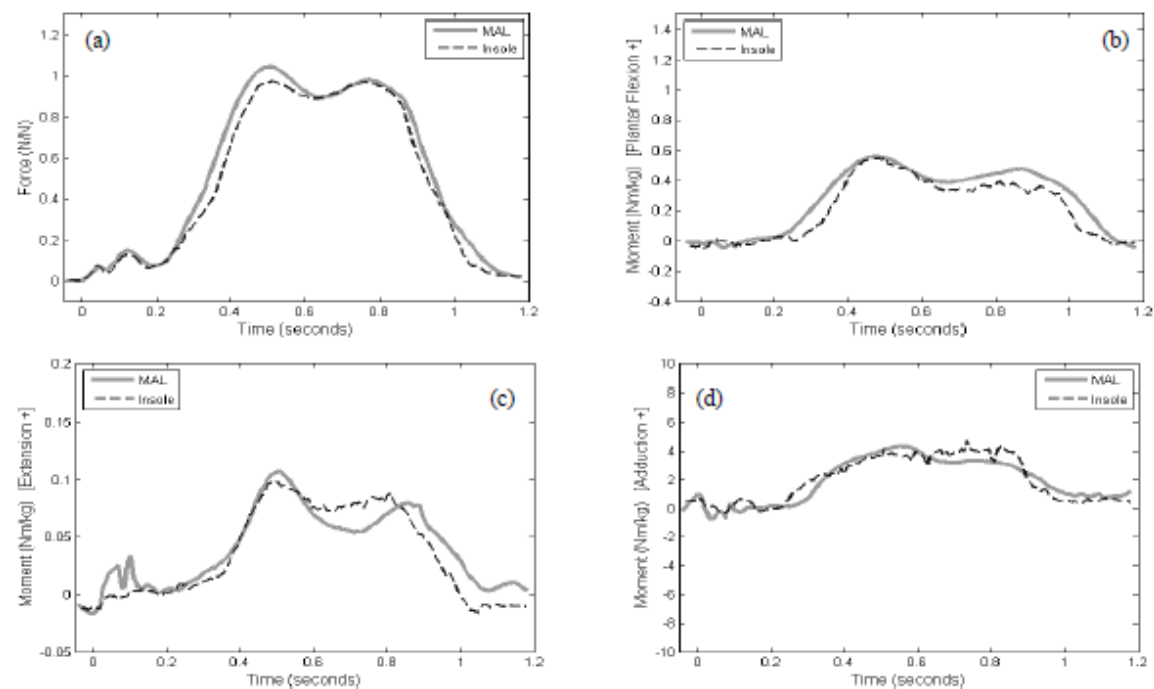

Fig. 3. A representative plot from a stroke subject showing motion analysis lab (MAL) and trained insole results for a) ground reaction force, b) ankle d/p moment, c) knee ex/fl moment and d) knee abd/add moment.

Inspection of the representative plots in Figs. 2c, 2d, 3c, and 3d indicates that the general shape is measured surprisingly well by the insole for both the control subjects and stroke patients. It is likely that the results can be improved by providing the system with an input corresponding to motion of the knee. Our next system will include a triaxial accelerometer and triaxial gyroscope in the ankle box that is worn on the leg. These outputs will then be used as additional inputs to investigate whether improved measurement of knee moments is possible.

Indeed, for all four parameters, the visual shape captured by the insole generally captures the magnitude and shape of the four kinetic variables as analyzed by the expensive clinical motion analysis equipment in the MAL. Differences captured by the MAL in stroke patients as compared to controls are similarly evident in insole data, as seen in Figs 3 and 4. The inexpensive insole can therefore be used to provide a quick estimation for kinetic gait analysis at the foot and ankle. These visual plots could be used to guide therapy and rehabilitation. For instance, the slope to the peak on the ankle d/p moment (as seen in Fig. 3b) corresponds to how well the subject can control dorsiflexion and plantarflexion. This information can be used to determine the appropriate stiffness of an orthotic. Alternatively, these results could be incorporated into a smartphone system, e.g. to monitor stroke patients throughout the day to complement the work of Sazonov *et al.* in monitoring the acitivity of stroke patients [6] or to provide feedback in real-time for gait modification [12].

In addition to generally reduced magnitudes, the stroke patients had a wider variety of shapes in the ground reaction force and the three moments. This was compounded by the fact that the reduced physical abilities of the stroke patients meant that less data was available for training and testing groups. Future studies will be designed to allow subjects appropriate rest periods until a larger number of steps are obtained. That will result in larger sets of training data that encompass a more of the various step shapes encountered.

A few problems were encountered in the course of the subject testing. A transceiver occasionally failed to complete data transmission for short periods of around 20 ms, possibly because the transceiver was temporarily obscured by a body limb or because there was interference from other equipment in the MAL. When a transmission was not received, the receiver duplicated the data from the previous time step. This resulted in visible error in the ground reaction force and ankle moment plots for those time frames. Subject A's data was re-analyzed with the two trials in which this occurred excluded from the eight steps collected. This resulted in insignificant changes (a small increase in %RMS error for ground reaction force from 5.35% to 5.74% and a small decrease in %RMS error for ankle d/p moment from 5.13% to 4.66%, both of which were within the standard deviations, and no change in the mean Spearman's correlation coefficients). This suggests that occasional data loss does not significantly affect the results, however, prior to our next study, we will investigate how to improve the wireless data transmission.

Some error was introduced in the MAL data for the stroke patients because the markers were placed on the outside of long pants allowed for comfort of the subject during testing in the winter. This may have contributed to the wide variety of shapes encountered in stroke subjects (i.e. if there was marker movement during or between trials). Also, all control subjects and stroke patients were tested wearing shoes, and therefore the markers usually placed on the toe and heel of the foot were placed on outside of the shoe, which may have introduced some error and/or reduced repeatability of the MAL data. The plug-in-gait marker set used in the MAL testing does not produce accurate data for the ankle pronation / supination moment, so that parameter was not tested with the linear regression. Adding another marker on the fifth metatarsal of region of the shoe would increase the accuracy of that parameter and allow for the insole to be trained to measure the ankle pronation / supination moment.

There are many ways to implement the testing and training data for the linear regression models. For this study, data was tested on trained data from the same subject. The long-term goal is to have enough data collected from a wide variety of patients to result in a set of linear regression models that can be used for any subject. Such a configuration will likely need a large number of subjects in order to train linear regression models that are related to the shoe size of the subject and whether the insole is placed between the foot and the shoe or under an orthotic. Stroke patient H performed the testing with and without his orthotic, and in most cases, the orthotic results were moderately better. The stiff surface of the orthotic may help distribute the loads under the foot to reach the small sensors, and may suggest that it is worth considering a stiff layer on top of the insole sandwich.

The normalized AP ankle moments in the control subjects were generally above 1.6 Nm/kg, which is high compared to an expected value of approximately 1.2 Nm/kg. However, the walking distance before the subject reached the force place was about 1.5 meters [32]. This may have resulted in the subject still being in acceleration phase when they reached the force plate, which would result in a larger moment than if the subject had attained a constant walking speed.

V. CONCLUSION

We have developed a low-cost insole-based kinetic gait analysis system. The ground reaction force and ankle dorsiflexion / plantarflexion moment measured by the insole were highly correlated with the motion laboratory measurements, and the %RMS errors were under 10%. In addition, the knee extension / flexion moment and the knee abduction / adduction moments showed strong promise as parameters that may be measured with the insole. Future work will include triaxial acclerometers and gyroscopes to provide additional inputs to the model. Other models, especially non-linear ones will be explored. In addition, a large array of subject data will be obtained to create a large set of training data. Our goal is to provide as system to study the kinetic aspects of gait when access to a clinical lab is not available. As indicated by the representative plots shown in Figs. 2 and 3, the shapes and magnitudes of the ground reaction force and

the three moments are biomechanically different for stroke patients as compared to the control subjects. For instance, the shape of the ankle dorsiflexion / plantarflexion moment in a stroke patient can be compared to a normal curve to investigate whether the orthotic that is being used is appropriate for the stroke patient's needs and capabilities.

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CHAPTER 4

CONCLUSIONS AND FUTURE WORK

4.1 Conclusions

This thesis documented the design, fabrication, and validation of an insole-based gait analysis system. A 32 sensor insole was developed and gave information about where the sensors should be located to be most effective. A 12 sensor insole was then developed with a linear regression model for each of the subjects on which it was tested.

Hypothesis 1 stated that the 12 sensor insole can predict ground reaction force with RMS error < 10% and a Spearman's correlation coefficient over 0.95. Experimental results demonstrated that the insole met the specifications. The average RMS error was 5.4% for the ground reaction force and for the control subjects and 6.4% for the stroke subjects. The Spearman's correlation coefficient was an average of 0.97 for the control subjects and 0.96 for the stroke subjects.

Hypothesis 2 stated that the 12 sensor insole can predict anterior-posterior ankle moment with RMS error < 10% and a Spearman's correlation coefficient over 0.95. The average RMS error was 5.9% for the control subjects and 9.8% for the stroke subjects. The average Spearman's correlation coefficient was 0.97 for control subjects and 0.95 for stroke subjects.

The average RMS error for knee flexion and extension moment was 10.7% for the control subjects and 13.7% for stroke patients. The average Spearman's correlation coefficient was 0.89 and 0.91 for the control subjects and stroke patients respectively.

The average RMS error for knee abduction and adduction moment was 16.4% for the control subjects and 17.1% for the stroke patients. The average Spearman's correlation coefficient was 0.84 and 0.82 for the control subjects and stroke patients respectively.

The low error and high correlation between the insole and MAL values validates that the 12 sensor insole system can be used for gait analysis.

4.2 Future Work

The 12 sensor insole performed well; however, there are still improvements that can be made in the immediate and more distant future. The insole design, the wireless transmission, testing procedures, and analysis techniques have areas where they can be improved.

Previous work on wearable gait measuring systems has given ideas about new things to try in the construction and design of our systems. For example, Dragoljub Surdilovic created a system where the sensors were placed on the exsole of the shoe rather than on the insole because he argued that there was too much variability in the shoes and how they fit the foot to provide the level of accuracy needed. He was measuring center of pressure and gait phases in robots, subjects with artificial limbs and patient rehabilitation [1]. Future work could include making a more robust system to be

worn inside the shoe that correctly measures the desired parameters for a variety of shapes of feet.

There are a few papers that discuss the FSR circuitry, implementation, and calibration procedure. N. Maalej characterized the FSRs and tested an amplification circuit to use in gait analysis [2]. Stephen Urry presented a study that discussed the lack of understanding of the relationship between the sensor characteristics and the associated data. He focused on the difference between pressure and force measurement and how that affects the results [3]. J.A. Florez discusses the time dependency of the FSRs in his research. This can lead to difficulty in calibration. He suggests that the best way to calibrate is to closely simulate the loading conditions that will be applied in the use of the FSR. He discussed the calibration of the FSR in static and dynamic applications and using a mechatronic device for that calibration [4]. In the future calibration of FSRs, it would be good to look into calibration techniques and how to get the most accurate data from them.

In our insole system, there were times when the transmission of data failed for one transceiver and there was a loss of data for about 20 ms. Work should go into improving the consistency and robustness of the wireless transmission of data. Alf Johansson did a study that included an investigation on the coexistence of WiFi and Blue Tooth signals with the wireless system. In his research, he saw loss of data packets when receiving signals from two insoles, one on each foot [5]. This is comparable to our system because there were two transceivers. Experimentation should be done to identify the reasons for the transmission error and to find ways to fix it.

Now that the system has been built and validated in the MAL, it should be tested in a variety of walking situations. Varying subjects, terrain, walking speed, and other conditions will provide much more needed information of how well the insole measures real life applications. As stated before, the MAL greatly limits the amount of steps that can be collected and also restricts normal walking conditions. As Bijan Najafi showed in his brief study of varying walking conditions, there are many possibilities to analyze with our system [6]. Guillaume Chelius tested his sensor network by doing a six day experiment of running through the desert [7]. S.M.N. Arosha Senanayake did a study that focused on the gait patterns of soccer players with FSRs located in their shoes [8]. There are many possibilities for testing our system now that we aren't confined to the motion analysis lab.

Further analysis will need to be done to create a built in model for linear regression that fits different sizes of the insole and different levels of abnormality in gait. These models would be calibrated with the MAL beforehand so that new subjects would not have to be tested in a MAL. Daniel Tik-Pui Fong's study in 2008 using a stepwise linear regression identified that different subjects, motions, footwear and floor conditions affected the accuracy of the training. He planned to look into those effects further but no publications from him on the subject can be found [9]. H. Rouhani tested two training strategies, one using intrasubject and another intersubject testing. He was testing on healthy subjects and subjects with ankle disease. Unhealthy subjects got higher error when trained on different subjects but that error improved when the training was applied on the same patient [10]. H.H.C.M. Savelberg carried the research further as he varied the

walking speed from 0.9 m/s to 2.3 m/s and investigated both intrasubject and intersubject training. He saw that speed was a factor in the learning methods [11].

The biggest area for future work is to expand the use of linear regression to make the knee moment calculations more accurate. Linear regression could also be applied to train the FSRs to measure kinematic parameters of the ankle and knee joints. The insole should also be able to measure the varus / valgus ankle moment, but will need to be validated with a new marker set that makes it possible for a more accurate calculation in the MAL.

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