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A Laboratory Insole for Analysis of Sensor Placement to Determine Ground Reaction Force and Ankle Moment in Patients with Stroke

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Abstract— **An insole system was constructed with 32 sensors inside a size 10 men's shoe. This system allows evaluation of the contributions of individual sensors spread throughout the surface area of the insole. The kinetic variables of interest in this initial study are ground reaction force and anterior-posterior ankle moment. Use of all 32 sensors are able to replicate the shape of the ground reaction force and ankle moment in a stroke patient who has regained a more normal gait, but less so in a stroke patient with impaired gait. Subsets of sensors can now be evaluated in order to ultimately identify an optimum set of sensors for determining kinetic variables necessary to classify presence or absence of a particular gait abnormality or other pathology.**

I. Introduction

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. 5.2% of adults between the ages of 18 and 64 (nearly 10 million) in the United States are classified with a walking disability [1]. Gait analysis, the study of walking, can be an essential component of rehabilitation and recovery.

Most commonly, gait analysis is done in a motion analysis laboratory, with expensive but highly accurate equipment such as infrared cameras and force plates. This allows determination of kinetic and kinematic parameters such as ground reaction force, moments, joint powers, etc., resulting in a computer model of a complete gait cycle. This allows results from healthy gait and abnormal gait to be compared to help identify functional problems and provide recommendations for treatment of those individuals.

Other methods of gait analysis have been the focus of research for several years. For example, a shoe- or insole-based gait analysis system allows the ability to analyze gait outside the motion laboratory. Some commercial systems (e.g. Tekscan (Boston, MA), novel (Munich, Germany), etc.) include insole-shaped custom pressure sensors that provide good measurements of kinetic parameters such as ground reaction force. However, these commercial systems generally cost more than $10K. This high cost generally precludes their use in a home or rehab environment. Alternatively, many researchers including ourselves (e.g. [2-8]) have developed inexpensive shoe-based systems to allow gait analysis away from the clinic at costs under $1K.

However, these inexpensive sensors typically come with a trade-off of lower accuracy and resolution. In addition, given a small number of sensors typically used in an insole, the placement of these sensors underneath the foot has generally followed the known biomechanical loading patterns of the foot, with sensors typically placed under the heel, metatarsals and great toe.

This paper presents the design of a lab-based insole system with 32 force sensitive resistors (Fig. 1), constructed in a size 10 men's shoe. This system was built to allow evaluation of the sensor performance and to identify the critical locations of sensors for estimating kinetic or kinematic variables of interest. In this paper, we will describe the design of this system, preliminary validation in control subjects and post-stroke patients, with an intention to identify sensor locations necessary to produce an estimate of ground reaction force and ankle moment within 10% of the measurements obtained by a clinical motion analysis lab.

Figure 1. 32 Sensor Insole

II. Materials and Methods

A. Insole Design

The insole system was constructed with 32 sensors inside a size 10 men's shoe (Fig. 1). The sensors used were Force Sensing Resistor, Model 402 from Interlink Electronics

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(Camarillo, CA). These sensors have been used by our lab and many others to build inexpensive insole systems. These sensors are very cost effective, thin, and robust. They do not require complex circuitry. They are limited, however, by their nonlinearity in loading.

The number of sensors (32) used was determined by the available National Instruments data acquisition module (DAQ), which allowed 32 analog input signals. In addition, this number of sensors was well suited to covering the footprint of the size 10 men's shoe. A Converse Chuck Taylor All Star (Converse, Inc., North Andover, MA) shoe was selected due to the flat nature of the footbed. The sensors were positioned (Fig. 2) so that they would cover the entire area of the footprint and particular attention was paid to ensuring that sensors were distributed under the heel, metatarsophalangeal joints, and the great toe.

Silicone insoles were constructed from Ecoflex 00-30 silicone rubber compound from Smooth-On, Inc. (Easton, PA) and used to protect the sensors in a sandwich design. First, a silicone layer approximately 4 mm thick was placed in the bottom of the Converse shoe on top of its normal insole. Next, the sensor locations and outline were traced onto a sheet of contact paper, which was 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 and not restrict the space inside (Fig. 3). The leads of the sensors were then fed through the holes and the sensor was adhered to the contact paper using the adhesive backing of the sensor. Last, after all of the sensors were located, the sensor layer was covered with another thin silicone insole, approximately 3 mm thick.

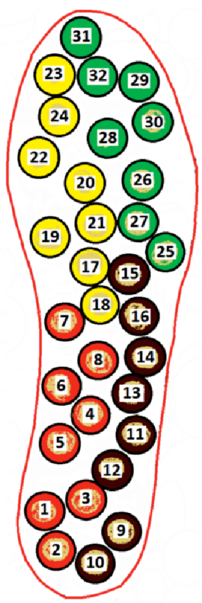

Figure 2. Layout of 32 sensors, with quadrants indicated by color

The shoe was divided into four quadrants (Fig. 2) for purposes of connecting the sensors to the conditioning electronics. Ribbon cable was used to connect the sensors to a circuit board that was carried in a pack on the subject's waist. Each quadrant was supplied with a 5 Volt supply from the DAQ that was daisy chained (as visible in Fig. 3) through the leads on one side. The other leads were grouped

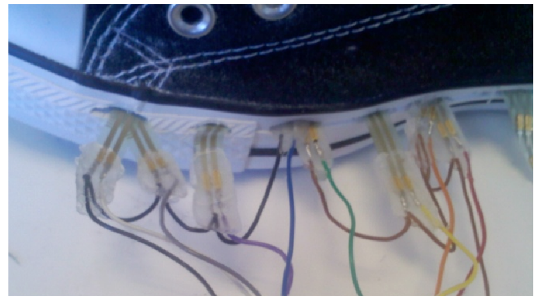

Figure 3. Close-up of sensor leads exiting the shoe

with their respective ribbon cable to connect to the board. Adhesive was used on the soldered leads to provide stability and stress relief (visible in Fig. 3). Each quadrant of sensors was connected to a voltage divider, using a resistor array with eight resistors. For this study, 1.0 KΩ resistors were used to match previous work in our lab; this circuit deign will facilitate future studies to investigate the performance of other resistor values. The ground of the voltage dividers was connected to the ground signal from the DAQ. The output from the 32 voltage dividers was transmitted to the DAQ using a 5 meter ribbon cable that exited the pack worn on the subject's waist.

The sensors were calibrated after construction of the insole using a load cell (as in [9]). Force was slowly loaded onto the sensor while both the insole and load cell stored data. The data was plotted with the FSR data in volts on the independent axis and the load cell data given in 1/1000 lbs on the dependent axis. Each run was curvefit with a polynomial equation that was used in the analysis to convert voltage readings on the FSRs to a force reading in Newtons.

B. Human Study

Control subjects and stroke patients with a shoe size close to size 10 men's were recruited to test the 32 sensor insole. The University of Utah's Institutional Review Board approved the study. 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 by the infrared cameras. Subjects were asked to walk on the force plates with the instrumented shoe. The MAL and the insole system captured the data simultaneously. 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, and so they were assisted as necessary in achieving this motion.

B. Data Analysis

Ground reaction forces and anterior-posterior ankle moments (corresponding to plantarflexion and dorsiflexion) for the MAL were determined using Vicon Bodybuilder and exported as text files. For the insole system, 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. For both systems, ground reaction forces were normalized by bodyweight in N

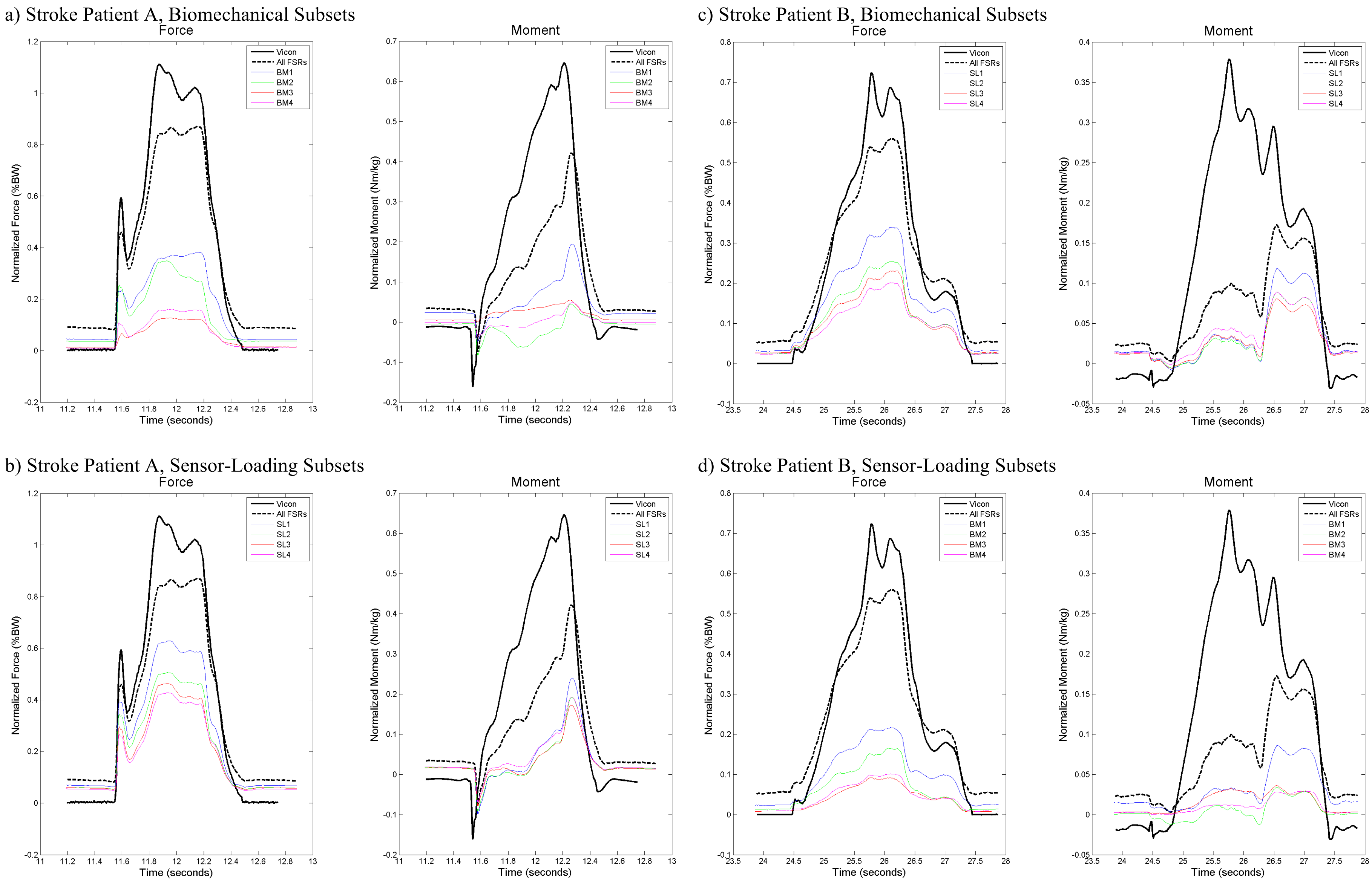

Figure 4. Representative plots showing normalized results of ground reaction force and ankle moment calculations for all 32 sensors in the instrumented insole, as well as various subsets of sensors (as detailed in Table 1 and Figure 1). The normalized ground reaction force and ankle moment determined by the Vicon system and AMTI forceplates in the motion laboratory are shown for comparison. Each trial consisted of one step on the force plate.

(resulting in units of %bodyweight), and ankle moments were normalized by bodyweight in kg (resulting in units of N·m/kg). All analyses of insole data and comparisons with MAL data were performed in MATLAB (Natick, MA). The MAL data was normalized by the total bodyweight reported by the patient, while the insole data was normalized by the bodyweight measured by the sensors (determined during quiet standing with each foot on a separate force plate).

For this preliminary investigation into optimum placement of sensors for analysis of ground reaction force and ankle moment, eight different subsets of sensors were analyzed. The calculations of ground reaction force and ankle moment described above were repeated for each subset. Four of these were inspired by typical biomechanical loading, e.g. sensors under the heel, the metatarsals (primarily first and fifth), and the great toe. Four other subsets of sensors were selected based on the mean and maximum loads encountered by each sensor during the gait tests with four control subjects and two stroke subjects. The sensors used in each subset are listed in **Table 1**.

Table 1. Subsets used for initial evaluation of sensor placement

Inspiration	Name	Sensors Included (locations: Fig. 1)
Biomechanical	BM1	1, 5, 9, 10, 19, 20, 22, 23, 25, 26
	BM2	2, 3, 10, 22, 25, 31
	BM3	3, 22, 25, 31
	BM4	5, 10, 14, 19, 22, 26, 31
Sensor Loading	SL1	1, 2, 3, 9, 10, 12, 20, 21, 23, 25, 26, 27
	SL2	2, 3, 9, 10, 20, 23, 25, 27
	SL3	1, 2, 3, 10, 23, 25, 27
	SL4	2, 3, 10, 20, 23, 25, 27

III. Results

The ground reaction force and anterior-posterior ankle moments for representative steps from the two stroke patients are shown in Figure 3. The results from the MAL are displayed, along with the results for all 32 sensors and the eight subsets, grouped by subset type. Stroke Patient A's ambulation pattern was similar to control subjects, while Stroke Patient B still suffered from impaired gait, and had a noticeably different gait pattern (as demonstrated in Fig. 3).

IV. Discussion

Inexpensive force sensitive resistors are typically considered useful only as switches, i.e. to indicate the presence of loading, but not to measure loading. The representative plots shown in Figure 3 suggest that a large number of insole sensors are able to measure many of the same trends and curves as the clinical motion analysis equipment in the MAL, although the scaling was not exactly right. Specifically, for ground reaction force, the use of a large number (32) of these sensors replicates the shape of the force curve well. For Stroke Patient A, the shape of the anterior-posterior ankle moment is also captured well by the 32 sensors. However, for Stroke Patient B, the ankle moment shape is quite different. Upon investigation of the motion lab data, it was clear that this subject's reduced plantarflexion resulted in increased loading through the arch.

Using 32 of these sensors in a wireless wearable insole is not very feasible, because of the difficulty in fitting the sensor tails in the insole. Instead, the purpose of building this wired insole was to provide a tool to identify optimal sensor locations corresponding to specific measurement goals. For instance, subsets BM1 and SL1 (which have 7 sensors in common) result in shapes similar to the 32 sensor results with 10 and 12 sensors, respectively. The magnitudes are smaller, but could be scaled appropriately. Neither subset improves on the 32 sensors for the ankle moment measurement in Stroke Patient B.

The next steps are twofold. First, these qualitative comparisons between the 32 sensors and the subsets need to be quantified across the stroke and control subjects. We plan to use percent change in root mean square (RMS) error to quantify absolute change at each time step, and Spearman's correlation to quantify how well the shape of the curves match. Second, the results for the ankle moment measurement in Stroke Patient B suggest that techniques other than direct calculation of ground reaction force and ankle moment may be necessary. That is, the contribution of forces under the arch is underrepresented and results in missing the peak ankle moment in Stroke Patient B. We plan to implement machine learning techniques such as least squares (e.g. [10]) to 'learn' the appropriate scaling for each sensor, as well as to determine the optimal sensor locations to allow classification of pathological and abnormal gait.

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