Intelligent Sensor Floor for Fall Prediction and Gait Analysis

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Intelligent Sensor Floor for Fall Prediction and Gait Analysis*

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Abstract—Falls account for a significant number of injury and death in the elderly. Risk factors for falls include increase in muscle weakness, change in gait and balance abnormalities. Detecting these changes are an important part of a smart health and aging environment. Because fall assessments do not occur very often, usually once a year in a patient’s annual physical, there is a need to develop a system that can monitor and assess gait and balance more periodically. This will allow older adults to continue living in an independent setting and reduce the need for expensive care facilities.

The objective of this project is to use gait analysis to accurately predict and detect anomalous behavior and help predict fall and health issues particularly in elderly persons. This paper describes the investigation and development of the Intelligent Sensor Floor used for fall and balance analysis. It can be used to monitor and assess fall risk in the home environment in an unobtrusive way. We will describe the hardware, circuit and floor construction process, sensor calibration, and initial data collection.

Index Terms—Intelligent sensor floor, gait, balance, pressure sensors, fall prediction, aging in place

I. INTRODUCTION

Falls are the leading cause of injury-related hospitalizations and account for 70% of accidental deaths in adults 75 years and older [1]. In addition, changes in gait, muscle weakness, dizziness and vertigo are common precipitants of falls, with the majority of falls occurring in or around a patient’s home [2]. Recovery for the elderly after a fall is markedly slower and a significant number experience functional decline in activities of daily living and in physical and social activities [1]. Fall assessments do not occur very often (i.e., usually once a year in a patient’s annual physical), and many falls do not come to the physician’s attention due to the patient withholding information [1].

To allow older adults to continue living longer in independent settings and thus reduce the need for expensive care facilities, low-cost systems are needed to detect not only adverse events such as falls but to assess the risk of such events, in addition to the early onset of illness and functional decline. Continuous, ongoing assessments of physical function would help older adults live more safely in independent settings, while also facilitating targeted medical interventions when needed. Ideally, such measurements would be obtained passively, in the course of normal daily activity [3].

In this paper, we describe the development of the Intelligent Sensor Floor and its functionality. More specifically, the floor should be able to:

1) detect and learn balance and gait patterns that is unique to each person
2) detect any anomalies in a person’s gait or balance
3) inform the patient, caregiver and/or physician about increased fall risk or a fall incident if it has occurred.

Furthermore, the Intelligent Sensor Floor should be least obtrusive to the patient as possible, (i.e., the sensors should be integrated into the patient’s environment rather than worn on the patient). The system can be constructed at a relatively low cost by using pressure sensors and floor tiles that are commercially available.

II. BACKGROUND

Gait is simply a particular way or manner of moving on foot. Every person has his or her own way of walking. There are several human factors, such as aging, injuries, operations on the foot, etc. that may change a person’s style of walking either permanent or temporarily. It also appears from early medical studies that there are 24 different components to human gait, and that if all the measurements are considered, gait is unique [4]. This has made gait recognition an interesting topic to be used to identify individuals by the manner in which they walk. This basic idea of gait analysis is the driving force behind the Intelligent Sensor Floor.

A. Gait Analysis

Gait analysis is the systematic study of human motion. The analysis involves the measurement, description, and assessment of quantities that characterize human motion [5]. Through gait analysis, gait phase can be identified, kinematic and kinetic parameters of human gait events can be determined, and musculoskeletal functions can be quantitatively evaluated. Gait analysis is used in the field of biomedical engineering, to characterize human motion. A major advantage of gait recognition is that it is unobtrusive. It can be measured at a distance, without the knowledge or cooperation of the subject, whereas current methods would require physical touch or close-range sensors [6].

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B. Gait Phases

Generally, human walking is a periodic movement of the body segments and includes repetitive motions. To understand this periodic walking course better, gait phases must be used to describe an entire walking period. In the past, normal events were conventionally used as the critical actions of separated gait phases. However, this practice only proved to be appropriate for amputees and often failed to accommodate the gist deviations of patients impaired by paralysis or arthritis. For example, the onset of stance has customarily been called the heel strike [7], [8]. However, the heel of a paralytic patient may never be in contact with the ground or may do so significantly later in the gait cycle.

Similarly, initial floor contact may be made by the entire foot (flat foot), rather than having forefoot contact, which occurs later after a period of heel-only support. To avoid these difficulties and other areas of confusion, the Rancho Los Amigos gait analysis committee developed a generic terminology for the functional phases of gait [9].

Analysis of the human walking pattern by phases more directly identifies the functional significance of the different motions generated at the individual joints and segments. Normal walking gait cycle is divided into eight different gait phases, i.e., initial contact, loading response, mid-stance, terminal stance, pre-swing, initial swing, mid-swing, and terminal swing as shown in Fig. 1 [10], [11].

![Gait phases](image)

Fig. 1. Gait phases in a normal gait cycle. (a) Gait phases of the stance period; (b) Gait phase of the swing period.

- **Initial contact:** This phase comprises the moment when the foot touches the floor. The joint postures present at this time determine the limb’s loading response pattern.
- **Loading response:** This phase is the initial double-stance period. The phase begins with initial floor contact and continues until the other foot is lifted for swing. Using the heel as a rocker, the knee is flexed for shock absorption. Ankle plantar flexion limits the heel rocker through forefoot contact with the floor.
- **Mid-stance:** This phase is the first half of the single-limb support interval. In this phase, the limb advances over the stationary foot through ankle dorsiflexion (ankle rocker), while the knee and hip extend. Mid-stance begins when the other foot is lifted and continues until body weight is aligned over the forefoot.
- **Terminal stance:** This phase completes the single-limb support. The stance begins with the heel rising and continues until the other foot strikes the ground, in which the heel rises and the limb advances over the forefoot rocker. Throughout this phase, body weight moves ahead of the forefoot.
- **Pre-swing:** This final phase of stance is the second double-stance interval in the gait cycle. Pre-swing begins with the initial contact of the opposite limb and ends with the ipsilateral toe-off. The objective of this phase is to position the limb for swing.
- **Initial swing:** This phase is approximately one-third of the swing period, beginning with a lift of the foot from the floor and ending when the swinging foot is opposite the stance foot. In this phase, the foot is lifted, and the limb is advanced by hip flexion and increased knee flexion.
- **Mid-swing:** This phase begins as the swinging limb is opposite the stance limb and ends when the swinging limb is forward and the tibia is vertical (i.e., hip and knee flexion postures are equal). The knee is allowed to extend in response to gravity, while the ankle continues dorsiflexion to neutral.
- **Terminal swing:** This final phase of swing begins with a vertical tibia and ends when the foot strikes the floor. Limb advancement is completed as the leg (shank) moves ahead of the thigh. In this phase, limb advancement is completed through knee extension. The hip maintains its earlier flexion and the ankle remains dorsiflexed to neutral.

Each gait phase has a functional objective and a critical pattern of selective synergistic motion to accomplish its goal. The sequential combination of the phases also enables the limb to accomplish three basic tasks, namely, weight acceptance, single-limb support, and limb advancement. Weight acceptance begins the stance period through initial contact and loading response. Single-limb support continues the stance through the mid-stance and terminal stance. Limb advancement begins in the pre-swing phase and continues through initial swing, mid-swing, and terminal swing. Based on the above analysis of the gait phases and basic tasks of limb movement, the gait phases may be detected effectively after orientations of the leg segments are accurately obtained.

To further complement the gait based health diagnostic system, another key parameter, pressure, is also taken into focus. As a matter of fact, in our modern living era, pressure sensing is one of the most performed measurements to enhance quality of life, encompassing a variety of specific applications. Measurement of interface pressure between the foot and floor underpins a number of important applications in quality of life and health monitoring, and abnormal pressure readings may indicate instability in gait.

III. MATERIALS AND METHODS

The main goal in the design of the Intelligent Sensor Floor was to make it affordable and be able to construct a system that can be easily implemented. Thus, the following materials and methods were selected with this goal in mind.

A. Microcontrollers

The prototype was completed using the Arduino Nano (V3.0) microcontroller. Eventually, we would like to replace the Arduino with the PIC18F452.
1) **Arduino Nano (V3.0):** The Arduino Nano (V3.0) is a microcontroller with an operating voltage of 5 volts, and a clock speed of 16MHz, with eight analog input pins and 14 digital I/O [12]. The Arduino was used to perform the analog to digital conversion. It was connected to the computer via USB and the digital output signals to the computer were in real-time.

2) **PIC18F452:** The PIC18F452 is an enhanced 16-bit program word and a family of high performance, CMOS, fully static MCUs with integrated analog-to-digital (A/D) converter. It has enhanced core features, a 32 level-deep stack, and multiple internal and external interrupts sources. A total of 77 instructions (reduced instruction set) are available [13].

### B. Pressure Sensor

The FlexiForce A401 sensor was chosen for the project because of its relatively low cost and ease of usage. The next subsections will discuss the pressure sensor’s nature and circuitry, loading area, as well as the conditioning process and calibration methods used.

Using the A401 FlexiForce sensor, we are able to record and analyze pressure readings as you walk across the floor. Furthermore, to make the system unobtrusive, the sensors were placed in the environment that is on the underside of the floor as shown in Fig. 2.

1) **Circuitry:** The A401 sensor acts as a force-sensing resistor in an electrical circuit as shown in Fig. 3. The sensor consists of a variable resistor whose resistance varies inversely to the force or pressure applied. When the force sensor is unloaded, its resistance \( R_S \) is around 1M\( \Omega \) which is very high. When a force is applied to the sensor, this resistance decreases. For this application, a voltage measurement corresponding to the resistance was evaluated in order to gauge and measure the force applied by unknown weights [14].

The circuit above uses an inverting operational amplifier to produce an analog output based on the sensor resistance and a fixed reference resistance \( R_F \). The inverting amplifier’s gain relationship is shown in (1). Equation (2) shows the results of the initial reference resistor value using a -3.3 drive voltage \( V_T \).

\[
V_{out} = -V_T \cdot \frac{R_F}{R_S} \quad (1)
\]

\[
V_{out} = -V_{in} \cdot \frac{R_F}{R_S} \Rightarrow 5 = -(-3.3) \cdot \frac{R_F}{30000} \Rightarrow \quad (2)
\]

From the equation, a decrease in \( R_S \) results by applying a force to the sensor and thus \( V_{out} \) increases. For a fixed reference resistance, \( R_F \), 22k\( \Omega \) was picked because it had the desired quality of sensitivity. To get a positive value for \( V_{out} \), it is required to provide -5V supply to the inverting op-amp.

2) **Sensor Loading:** According to the FlexiForce data sheet, the active sensing area of the sensor should be treated as a single contact point, with the load distributed evenly across the sensing area. For best results, the loaded area should cover 70-100% of the sensing area and apply a minimum of 1PSI. The load was centered in the sensing area and not near the edges if the load was smaller than the active sensing area.

Since the load (a tile) was larger than the active sensing area, we started by using a “puck”, which is a piece of rigid material smaller than the active sensing area that is placed between the object and the sensing area. Later, we used a piece of EPDM rubber roll cut to size for the sensor. The EPDM rubber has excellent stretching capabilities while maintaining an average of 1,000PSI tensile strength. This caused the load path to go through the EPDM rubber and was evenly distributed across the entire sensing area, resulting in more accurate measurements.

3) **Conditioning:** Before using the FlexiForce sensor for the first time or after long periods of inactivity, it is highly recommended that it be conditioned in order to stabilize the output resistance. The output resistance can fluctuate over the first few tests, thus conditioning the sensor ensures repeatable results. To condition the FlexiForce sensor, 110% of the maximum test weight was placed on the active sensing area for a few seconds. This process was repeated another four to five times in order to guarantee accurate results.

4) **Calibration:** The first step of obtaining accurate sensor data from the FlexiForce is by calibration. This was performed according to the steps provided by the Tekscan website [15].
At any given time a person can be standing on a single sensor, therefore each sensor should be able to read the person’s full weight. An average adult’s weight is assumed to be 100 pounds or greater. As a result, several different calibration tests were performed to determine the best feedback resistor to use in the circuit. We started with a resistor value of 22kΩ, and applied pressure weights ranging from 10 pounds to 30 pounds. The resulting calibration plot can be seen in Fig. 4 below. The maximum pound force each sensor could hold was about 22.5 pounds, and a maximum total of 77 pounds was measured when the tile was placed on all four sensors. This limitation was a result of the 22kΩ resistor that was used in the recommended circuit shown above in Fig. 3.

![Fig. 4. Calibration graph for a 22 kohm resistor](image)

In order to achieve the desired maximum force that each sensor should be able to withstand when underneath the floor, we had to further reduce the value of the input resistor to 1kΩ, which is the minimum resistor value allowed. With this new setup, we were able to measure pound force values of up to 125 pounds, as shown in the calibration plot in Fig. 5. As a result, this resistor was chosen over the 22kΩ resistor. We were not able to experiment further with heavier weights due to the limitation of the testing environment.

![Fig. 5. Calibration graph for a 1 kohm resistor](image)

Equation (3) and (4) show the cutoff frequency for the 22kΩ and 1kΩ resistors respectively, and (5) and (6) show the resulting linear equations from the calibration tests of the 22kΩ and 1kΩ resistors respectively, where \( y \) represents weight in pounds and \( x \) is the sensor reading in volts.

\[
X_c = \frac{1}{2\pi R F C} = \frac{1}{2\pi \times 22000 \times 10^{-6}} = 72.3 \text{Hz} \quad (3)
\]

\[
X_c = \frac{1}{2\pi R F C} = \frac{1}{2\pi \times 1000 \times 10^{-6}} = 15.9 \text{Hz} \quad (4)
\]

\[
y = 6.4685x - 0.7026 \quad (5)
\]

\[
y = 140.76x - 2.655 \quad (6)
\]

C. Design Specification

The prototype construction as shown in the schematic in Fig. 6, consists of a voltage regulator (MCP1827S), which regulates a 5V input to 3.3V. A voltage converter (TL7660) was used to convert the positive 3.3V to -3.3V drive voltage needed at the negative terminal of the rail-to-rail instrumentation operational amplifier (MCP6004) as well as on each sensor input. Analog sensor inputs into the operational amplifier were fed into an 8-1 analog multiplexer (HTC4051) and later on into the Arduino or PIC18F452. The three select inputs from the multiplexer were also connected to the Arduino/PIC18F452 digital output pins D1, D2 and D3. The Arduino was configured to output the analog result on the serial monitor with the maximum baud rate of 115200, and later analyzed using MATLAB to get the center of pressure. For easy interfacing to a PIC’s 0-5V analog input range, the rail-to-rail instrumentation operational amplifier was used. This instrumentation op amp subtracts its two inputs, multiplies the difference by the gain, and adds on the reference voltage.

![Fig. 6. Prototype schematic with the PIC18F452 with 16 analog inputs, but not all inputs on the multiplexer were connected due to the lack of space to include a second set of 8 op-amps on the schematic.](image)

The PIC18F452 or Arduino is in the center. Above the processor you can see a crystal and two capacitors giving us the required 20MHz reference frequency. It turns out that
low-speed USB is more flexible with frequency accuracy than the faster variants and a ceramic resonator would have been sufficient and a little cheaper. Just to the right of the processor is a decoupling capacitor and a resistor holding the chip’s reset pin high.

This is everything you need to get the device up and running. You can see the prototype connected to the computer in Fig. 7 and the completed installation of the floor in Fig. 8.

**IV. Data Collection & Analysis**

Data was collected both from a Wii Balance Board and the Intelligent Sensor floor once it was fully installed.

**A. Wii Balance Board Data**

To determine what rate the floor should take pressure readings, we used a Wii Balance Board as reference. An open source Flash Wii library provided the interface between the Wii Balance Board and a Mac using its built-in Bluetooth [16]. In addition, an example driver was provided that gave the total weight of objects set on the Balance Board [17]. This driver was edited to give readings for each of the four force load sensors built into the Balance Board. These readings along with timestamps were printed to a file. The data collection consisted of recording subjects stepping onto the Balance Board, standing for approximately 10 seconds, and then stepping off. This process was performed three times for each subject to get data for standing on each foot as well as on both feet.

From this data, it was calculated that the Wii Balance Board takes readings at approximately 100 Hz. In addition, a fast Fourier transform (FFT) was performed using MATLAB on the total weight and each of the four sensors for each data set. The resulting periodogram of the total weight when standing on one foot indicated a unique peak in the 10-15 Hz frequency range for each of the three subjects from which data was collected. Fig. 9 shows this peak, which suggests that there is a common frequency component for people standing on one foot. However, more data needs to be collected from a larger pool of people to confirm this indication.

![Fig. 9. FFT of total weight for Wii Balance Board data.](image)

Furthermore, the center of pressure (COP) was calculated using the formula for calculating the center of mass given in (2).

\[
COP = \frac{m_1x_1 + m_2x_2 + \ldots + m_nx_n}{m_1 + m_2 + \ldots + m_n} \quad (2)
\]

The pressure readings from each sensor in the Balance Board were used as the masses \(m_i\) and the dimensions of the Balance Board used as the distance \(x\). The equation was used twice for each pressure reading to calculate the center of pressure in the x-axis and the y-axis, and a two-dimensional graph can be seen in Fig. 10. From this, the x- and y-coordinates were plotted with time in MATLAB to give a three-dimensional graph, shown in Fig 11.

**B. Intelligent Sensor Floor Data**

Once the sensors were installed underneath the floor, the center of pressure was calculated and plotted similar to how it was calculated with the Wii Balance Board data. Fig. 12
Fig. 11. Three-dimensional plot of the center of pressure from the Wii Balance Board.

shows a 2-D plot of COP and Fig. 13 shows a 3-D plot of COP.

Fig. 12. Two-dimensional plot of the center of pressure from the Intelligent Sensor Floor.

Fig. 13. Three-dimensional plot of the center of pressure from the Intelligent Sensor Floor.

that analyzes changes in balance and gait patterns to detect anomalies will have to be developed to help assess fall risk among the elderly.

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