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Design and Development of a High Resolution Open-Source Plantar Pressure Sensing System

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Abstract

Diabetic neuropathy is a debilitating complication of diabetes that is often diagnosed long after irreversible damage has occurred in a patient. Diagnosis of the condition commonly occurs when examinations reveal physical changes in foot structure and ulcers of the foot; such damage reveals the condition but offers no chance at prevention. To better predict diabetic neuropathy and prevent damage, pressure sensors can be used to detect the onset of the condition before damage occurs. While several pressure sensing and foot mapping systems are commercially available, each is prohibitively expensive, requires specialized software, or maps a limited portion of the foot. We present an affordable, scalable, and high resolution pressure sensing system that maps the entire foot with a novel force sensor and sensor grid. The grid is based on a matrixed array of 704 individual sensors actuated by XactFSR resistive film. Each sensor consists of interlaced sensing fingers and measures 0.250 inch (6.35 mm) square. The device offers a solution that is several thousand dollars cheaper than other products, consists of commercially available boards and cables, and provides full mapping of the foot while operating with standardized and open source software packages. Our system provides emergent economies and regions presenting a high risk of diabetes with a predictive tool that can operate in nearly any environment.

Keywords: Diabetic foot ulcer • Flexible force sensor • Sensor grid • Foot worn sensor • Open source

Introduction

Diabetic Neuropathy has become a pervasive condition in the United States and abroad; over 200,000 cases are reported in the United States annually and global cases number in the hundreds of millions [1,2]. The condition is a complication of diabetes and is the most common complication stemming from diabetes mellitus [3]. Diabetic Peripheral Neuropathy (DPN) is a very common presentation of diabetic neuropathy with estimates that up to 50% of Type II diabetes sufferers will develop some form of the condition [2]. DPN, as the name suggests, affects peripheral nervous systems and is most commonly found to affect the lower extremities (feet and ankles). Elevated levels of blood sugar cause irreversible damage to nerve endings in extremities with the damage increasing in patients that poorly control their blood sugar levels [2].

Damage from DPN can be quite severe; loss of feet and legs is common in advanced cases. While the condition is initially fairly innocuous, DPN can lead to insensitivity of the limbs or a “pins and needles” feeling that dulls the patient’s awareness of damage [3]. The condition is rarely diagnosed without a neurological exam that occurs after physical examination of the patient reveals some combination of decreased sensation of the feet, weakened (or absent) ankle reflexes, or other neuropathic systems (e.g., tremors or abnormal gait) [2]. The presence of ulcers of the foot are commonly used to diagnose DPN, but an enormous amount of damage has already been sustained by the time ulcers emerge; damage up to that point is likely irreversible. More intensive tests, such as electromyography, and invasive tests (including biopsy) also serve to diagnose DPN. Considering the potential damage from DPN and the condition’s diabetic roots, it proves pragmatic to address the cause.

Unfortunately, diabetes is on the rise globally. The Centers for Disease Control (CDC) cite the rate of diabetes in the United States as over 10% of the population with a 5% increase annually [4]. Globally, more than 350 million people are predicted to suffer from diabetes by 2030 [5]. As many as 50% of diabetes patients will suffer from some effect of DPN with up to 30% of those DPN cases resulting in amputation or other debilitating condition [6-8].

Reduction of global obesity and the associated case of diabetes is an enormous undertaking that requires addressing poverty, reconsidering diet and lifestyle, and a number of items that would require a worldwide effort. Our focus is much narrower and aims to minimize future cases of DPN. While we have established that diabetic cases are unlikely to decrease as we move towards 2050, we also find that current diagnostic methods for DPN tend to be “posthumous” the condition is found due to damage already present. Our goal is to predict the onset of DPN before it can cause irreversibly damage to the patient. Risk factors for DPN are well known type 1 or type 2 diabetes must be present for the condition to occur [9]. Obesity, diet, lifestyle, and a number of other risk factors contribute to the chance that DPN will occur. Medicine has a clear understanding of the causes and underlying conditions leading to DPN all that is needed is a method to monitor areas of risk and provide a diagnostician with the clinical tools necessary to detect precursors to DPN.

Background

While pressure sensors represent a large market with numerous choices, they can be generally grouped as Insole system and or stationary pressure mapping system. Insole systems are typically limited by low sensor counts and the associated lowered sensor resolution [10]. Most of the In sole systems must be customized and fitted for each user. Stationary

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pressure mapping system utilizes a large sensor bed, thus providing a high density pressure mapping and do not require to be customized for individual subjects, but these types of systems must be used in a clinical setting. As with all commercially available gait analysis and pressure sensing systems, proprietary software must be purchased to analyze any collected data.

It is also worth noting that the focus of most modern pressure sensing efforts has been on insole technology. Such a focus assumes that the user has shoes that fit appropriately. In a broader consideration, it assumes that the patient has shoes. Weak economies possess populations with shoes that may affect collected insole data due to poor fit or being heavily damaged; many populations may simply lack shoes to place insoles into. Cultural considerations may also affect the reliance on insoles sandal wearing cultures (e.g., Myanmar and Southeast Asian tropical regions) and remote tribal groups (e.g., Uganda and Ethiopia) may not possess footwear that is compatible with insole technology. Moreover, the use of insoles may result in an unnatural gait that will skew any collected pressure data.

By reviewing the global needs and analyzing available commercial systems, we can summarize traits that are desired in a new foot sensor system and weakness of commercially available systems.

As shown in Table 1, with the above criteria in mind, we note that while many commercial systems satisfy some desirable traits, they all possess several drawbacks. Proprietary software requirements, prohibitive costs, and limited sensor densities are the most seen flaws in the commercial product pool [11]. As we consider diagnostic work in emergent economies and amongst indigenous populations, it is obvious that expensive systems and proprietary software will eliminate a system from consideration. Few sensors' densities limit a less affluent team's ability to properly diagnose and prevent DPN, since high sensor density is critical for a successful and timely diagnosis to prevent foot ulcer, as discussed by Ostadabbas, Saeed, Nourani, and Pompeo [12].

Table 1. System Trait Summary

Desired Traits of New System	Weaknesses Seen in Commercial Systems
Open-Source software	Proprietary software required
High sensor density	Limited sensor density
Affordability	System Cost
Reusable sensors/components	"Single-Use" Sensors
Robustness	Power/interface requirements
Able to be used in indoors / outdoors settings	Clinical setting requirements
Modular	Reliance on insoles
Maximum sensor resolution	Varied sensor resolutions

We seek to eliminate the limitations seen in current commercial systems and incorporate the aforementioned desired traits in a novel, affordable system. Open source software, modular components, and the ability to perform diagnosis without accoutrements common to the First World will drive the design of the new system.

Materials and Methods

While the concept of pressure sensing for diagnosis and prevention of diabetic ulcers and foot damage is not new, our approach and goals are both novel and ambitious compared to the previous efforts. We seek to provide a commercially viable system that incorporates academic research to provide a solution that is affordable while offering broader capability than standard products. West Texas A&M University, undergrad senior design students led the first design phase of plantar pressure device from January 2020 till August 2020, with a focus on developing a sensor matrix to map relative force. The project involved identifying the optimum size of each individual sensor, sensor selection, interface selection and prototype development [13].

Since high sensor density is critical to successful and timely diagnosis to prevent foot ulcer, we investigated the following: average size and minimum size of an ulcer wound, pressure spot developed in foot and orthotics discussed in publications [14,15]. As per our findings, we chose the size of individual sensor at 0.25 inch (6.35 mm) by 0.25 inch with an assumption that anything that covers 25% of the sensor surface can be measured accurately, which was tested later in the project and found accurate [13]. To match an average foot size, the sensor matrix consists of 44 × 16 sensor array of 704 sensors over a bed measuring 11 inches (279 mm) by 4 inches (102 mm). This configuration matched and, in many instances, far exceeded the highest resolution pressure sensors available on the market.

Generally, the sensor technologies used for pressure mapping are capacitive sensors, resistive sensors, piezoelectric sensors, and piezoresistive sensors [13]. Resistive sensors tend to be the simplest in design yet accurate enough for the application, so we chose the Sensitronics XactFSR force sensing resistor.

The sensor design itself is relatively straight forward with a single force sensor consisting of interdigitated, but electrically isolated fingers as seen in Figure 1. In the Figure, a single sensor is shown; the vertical columns of fingers are electrically isolated from their direct neighbors, so that "upward" fingers are on one net and "downward" fingers are on another. A single via can also be seen on the lower left hand side of the image that links each sensor to one of 16 columns on the reverse side of the sensor bed. These columns allow for the matrixed reading of the sensor bed's 704 individual sensors by the programmed software.

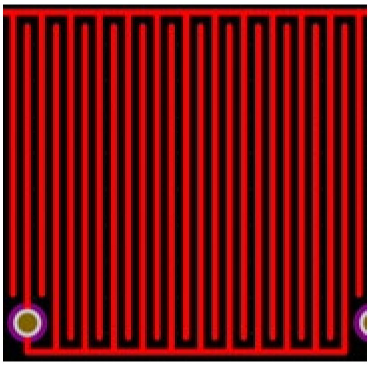

Figure 1. Single sensing cell

For the sensor to function, a resistive film is placed over top of the full sensor bed where it may contact the interdigitated fingers. When pressure is applied to the system, the resistive film contacts the sensor and the sensor traces allowing voltage to flow between them the mechanism is commonly referred to as shunting (shorting). As pressure changes, resistance between the two previously isolated traces changes as well; more force lowers resistance (to some threshold) and produces a higher signal from the sensor. When designing such a sensor, trace sizing and spacing are also critical as it has a direct relation to sensitivity of the sensor.

For the space and trace of the sensor, we examined recommendations from the manufacturer of our chosen resistive film (Sensitronics). In the previous work, a space and trace of 0.006 inch was chosen after some initial testing while Sensitronics suggests 0.007 inch for an all-purpose sensor [13-23]. For this work, a space and trace of 0.005 inch was chosen to provided better low end sensitivity as smaller space and trace sizes provide greater sensitivity to small forces [23]. When looking for subtle foot changes over time, greater sensitivity to small shifts should aid diagnosticians. ENIG (gold) finish was also chosen for the sensor traces as it was recommended by Sensitronics [23]. Lastly, half ounce copper was chosen for the production weight to ensure longevity of the sensors. The aforementioned decisions (trace and space, finish, and copper weight) also ensured high manufacturability and high sensor resolution goals we set to achieve from our conceptual design phase. The full sensor design along with a manufactured sensor can be seen in Figure 2 below.

Figure 2. Sensor bed traces (left), manufactured sensor bed (right)

Initial efforts focused on incorporating a Raspberry Pi for data acquisition and recording. Early on in the investigation, the Raspberry Pi proved to be problematic and an Arduino Uno proved much easier to work with for early efforts [18]. Ultimately an Arduino Mega was utilized as it had the number of pins needed to support the full sensor array and allowed for accurate, rapid recording of data [13]. To improve performance further Teensy 4.1 microcontroller was used in final prototype, which outclasses the Arduino Mega in every category relevant to the system design.

After researching a number or connectors, Samtec Edge Connector series parts were chosen as they possess a small footprint, robust, allow thousands of cycles, allow rapid data transfer (up to 56 Gbps), and are commercially available and an off the shelf part [24].

The controller board design consists of Teensy 4.1 Microcontroller, Samtec connector (for the sensor pad to connect), resistors to complete the sensor shunting circuit, multiplexers and associated hardware. Since some of the pins at Teensy 4.1 was not readily accessible, a Texas Instruments High Speed CMOS Logic, 16 Channel Analog Multiplexer (CD74HC4067M) was used. The device was an ideal candidate for this work due its low price, high speed (6 ns switching speed) 30, resistance to noise 30, robustness, and availability. A 3D Rendered Board was designed using Altium as shown in Figure 3.

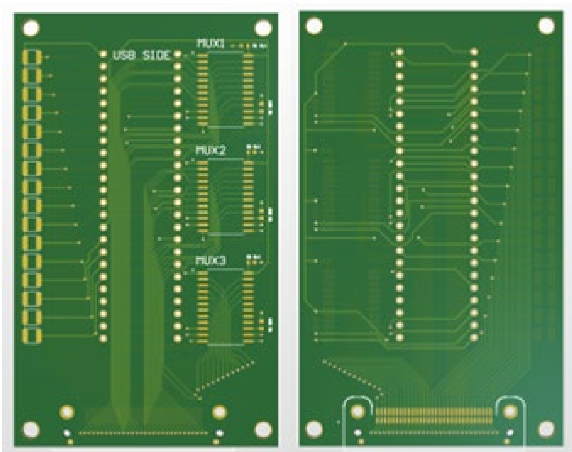

Figure 3. 3D Rendered board; top side (left), bottom side (right)

Portable chargers (or power banks) with a USB output have become extremely common and represent a durable, long lasting replacement for disposable batteries. Due to their decreasing cost and wide availability, a "lipstick" style USB charger was chosen to compliment the system.

From the literature review, it was inferred that plantar pressure must be measured when standing or walking [19-22]. By understanding the general motion of the human foot during locomotion, we came up with a sensor platform, both flexible (to match the contours of the moving foot) and rigid (to ensure accurate pressure readings). A "linkage" design, similar to tank treads was used in the sensor platform that can conform to individual foot curvatures. It also provides an unyielding sensor backing for the required platform. Figure 4 shows the push fit tread design as a single tread and as a small assembly.

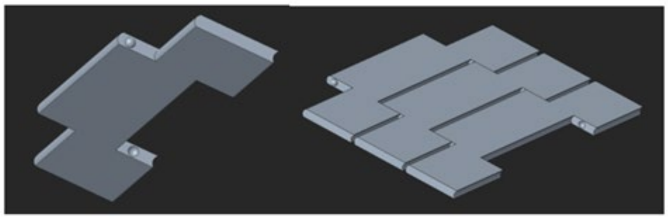

Figure 4. Push-Fit tread design; single tread (left), assembly (right)

For prototyping and initial testing, pieces were 3D printed. VeroClear, a proprietary material designed to mimic PolyMethyl MethAcrylate (PMMA) [25-27], was the material used to print the prototype pieces of sensor bed and circuit enclosure. Though Vero Clear is reasonably flexible at 0.125 inch, the material provides the needed rigidity and unyielding surface desired for a sensor base. Stratasys touts the strength and rigidity of the Vero Clear [27], citing its relevant physical properties as shown in Table 2.

While these values must be understood to ensure we produce treads that are rigid, unyielding, and can withstand the expected forces, we also consider them as alongside plans for the future product. The 3D printing of these treads is likely not a commercially viable solution; polyjet printing is time consuming, materials are expensive, and large volumes are impractical. Our design has the capability of bulk production. Though we must be able to utilize additive manufacturing for prototyping, we will look towards injection molding for bulk production. With injection molding, the material of choice will be Acrylonitrile Butadiene Styrene (ABS) plastic due to its strength, rigidity, and durability. Its physical properties are shown in Table 2.

Table 2. Physical Properties

Material	Tensile Strength	Modulus of Elasticity	Flexural Strength	Flexural Modulus	Rockwell Hardness
VeroClear	50 MPa–65 MPa	2000 MPa–3000 MPa	75 MPa–100 MPa	2200 MPa–3200 MPa	Scale M, 73 – 76
ABS	43.6 MPa	2030 MPa	70.5 MPa	2070.0 MPa	Scale R, 68 - 118

As seen from the table above, VeroClear is a reasonable surrogate for ABS. While ABS is slightly weaker more easily deformed and also less dense. The lower density will allow for a lighter sandal in the final product. As ABS is slightly weaker, treads with an overall thickness of 0.250 inch will also be designed and tested to ensure that no issues arise during the transition to injection molding [28,29].

Results and Discussion

Loading simulations

Although initial physical proofing of the prototype pieces worked well (a single tread could bear a load of up to 200 lbs), we chose to examine the potential failure of the tread pieces using Creo Simulate 4.0. Our primary focus was on the load of 1000 lbs as it represents an extreme scenario either a 1000 lbs. individual standing still, or the maximum expected pressure generated by a 500 lbs. load during walking [30].

Case I: We first examined a flat loading of the entire foot over a tread assembly in Figure 5. With 1000 lbs of load in the z-direction, the maximum Von Mises stress seen was just over 260 psi (note that units in this section will be in psi unless otherwise specified). Notice that the heel piece sees very little stress due to its larger size; as the heel piece does not experience stress in the way the smaller pieces do simulation efforts were not focused on small pieces. Expected yield strength of ABS plastic is roughly 7000 psi making additional simulation efforts of a full foot not worth pursuing [29].

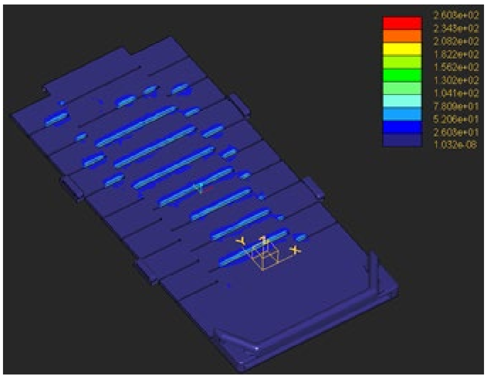

Figure 5. Simulation of entire foot on treads at 1000 lbs. Z-loading

Case II: We now begin investigating worst case scenarios where loading is isolated to only a part of the foot. In the first case, we examine loading on the forefoot and toes during the "push off" phase of locomotion. We once again consider the case of 1000 lbs. of loading in the Z-direction in Figure 6. The Von Mises stress is this scenario is much more dramatic, reaching a maximum of 2830 psi (note that Figure 6's units are in ksi). However, psi levels over 1130 occur along edges and corners in the design. These features can be engineered out in future iterations. Even in this scenario, we are far below the 7000 psi yield strength of ABS.

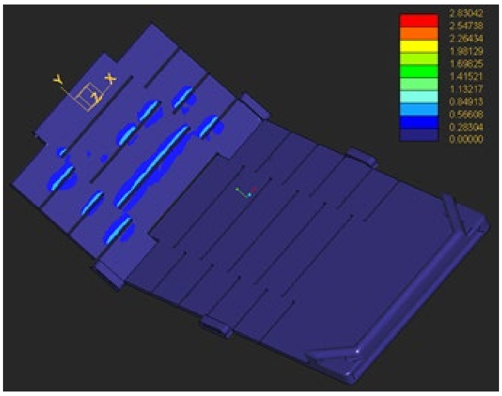

Figure 6. Simulation of forefoot on treads at 1000 lbs. Z-loading

Case III: For our final set of simulations, we move to the most extreme case where all loading is on the big toe of an individual. It is admittedly unrealistic that a patient would have the entirety of their weight on a single toe as by the time only the toe is in contact, the opposite foot is bearing the bulk of the body's load, but this scenario is simply designed to determine if the material will fail under any possible condition (no matter how remote).

Case IV: Figure 7 shows 1000 lbs. of loading in the Z-direction at only the toe. At last, we have a peak Von Mises stress that is almost exactly the failure loading of ABS plastic. However, the peak stress is again found on the edges and corners that can be engineered out of future versions.

We did more simulations on several scenarios and have produced results where breakage is possible, for 500 lbs. One such scenario is that the person is putting their entire weight on a single toe. By the time the weight would be focused on a single toe, the opposite foot is bearing the bulk of the body's weight [21,22]. Since such a scenario is not credible, we find the design and choice of ABS suitable for our needs (with the current design and at 0.125 inch thick). If we were to make a very conservative estimate, the tread design could be limited to users below 350 lbs. Realistically, the design will support patients of up to 500 lbs.

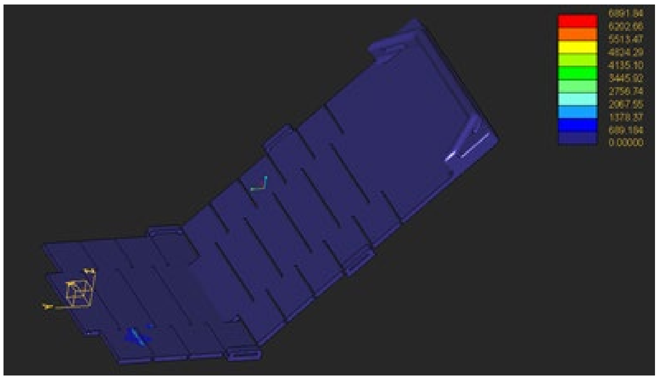

Figure 7. Simulation of toe on treads at 1000 lbs. Z-loading

Testing

For our work, testing was focused on two areas:

1. Characterizing the response of individual sensors
2. Testing the final product

To test individual cells, a 6 inch diameter aluminum platform was fabricated. The platform was then mated to a steel rod with a 0.400 inch diameter head; such a configuration allowed weight loaded on the platform to be transferred to the face of the steel head. After initial testing, being able to place a cushioning material on the face of the steel piece would prove critical. We should also note that the 0.400 inch diameter head was purposefully larger than the 0.250 inch square individual cells to allow for installation of materials at the face and manual shaping of the interface. Testing results can be found in Figure 8. Results were very consistent between the three runs and matches with the manufacturers data sheet, providing confidence that the sensor bed will provide data that can eventually be used for absolute loadings rather than just relative values. The consistency found in this initial data also provides some confidence that results will be consistent over time.

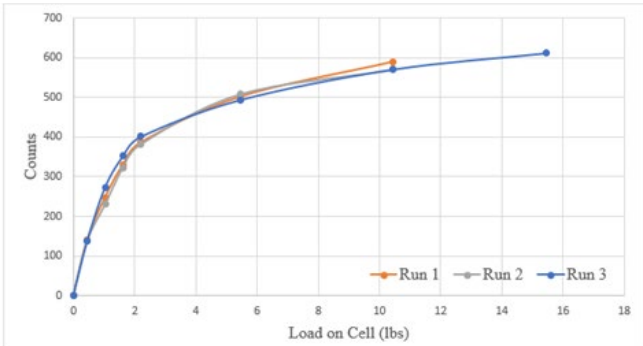

Figure 8. Single cell testing; 0.005-Inch Thick, 450 kOhm/Inch^2 Resistive Film

Looking at the data, relatively linear behavior is apparent in two regions: from 0 to 1.5 lbs and from 2.5 lbs to 15 lbs. The shape of the curve and relatively sharp transition between regions is somewhat expected and it is the characteristic of the resistive film [23]. For future work, a curve can be fitted to the data that allows for calculation of weight from the counts of a sensor.

Having verified the system's functionality and characterized single sensor behavior, we moved to examining performance of the full system. For our test cases, two subjects (one male and one female) stood on the sensor bed with the system in its ultimate configuration. The system's button was momentarily actuated, and data was captured. Data was sorted and analyzed via a MATLAB routine. Figures 9-11 show the single foot system with parts connected and in place and results of the analysis.

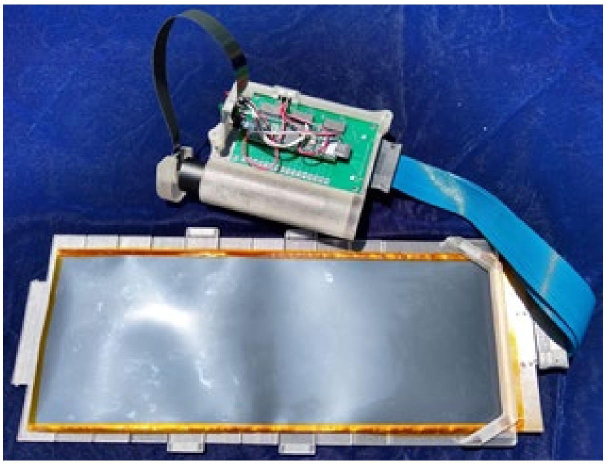

Figure 9. Single foot system with parts connected and in place

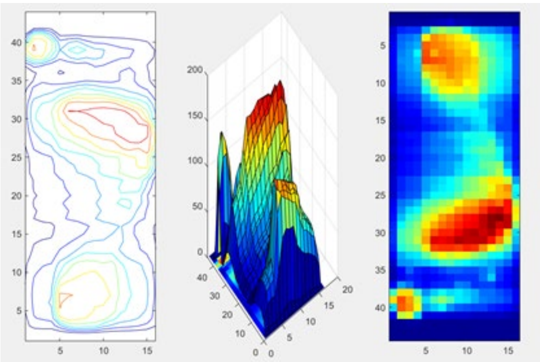

Figure 10. Male subject analysis results

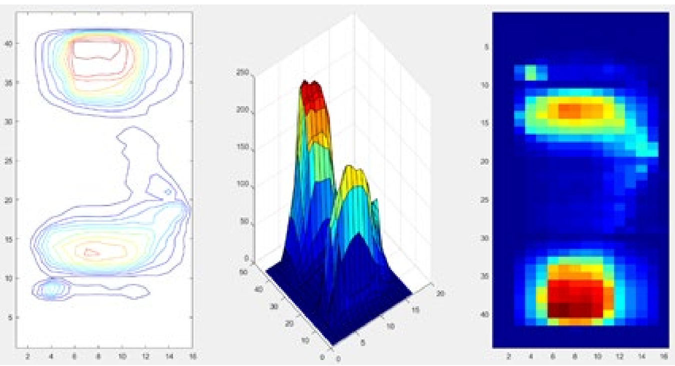

Figure 11. Female subject analysis results

Results presented here show the current output of the Matlab based analysis routine for a single reading. Numerous readings were taken, but only a single snapshot of each analysis is presented here for illustrative purposes. Matlab's native visualizations allow for generation of contour plots, color maps, and 3 dimensional mappings for each reading, allowing users to choose the visualization method that they prefer.

For a medical professional, a number of things can be gained from this analysis with no other medical history. More importantly, and in the context of our project motivation, these images provide a baseline pressure mapping for each subject. With each subject's unique footprint, we now have a basis for comparison over time. By testing at regular intervals, changes to pressure distribution can be used to diagnose subtle changes in the foot structure and stance, allowing diagnosticians to see the onset of DPN well before irreversible damage occurs. More broadly, the system will detect any changes in pressure mapping, allowing changes in the feet due to injury or other conditions to be diagnosed.

Final system features

In the early sections of this work, we stated the general considerations and design of an affordable, open source plantar pressure sensing system. While the need for this work, design considerations, and functionality are apparent, we feel it is important to summarize the key features of this pressure sensing system in terms of both our initial goals and existing, comparable systems.

Scalability: Though the system is designed to be nearly one size fits all or customizable by the user, the sensor itself is designed to be of singular size and width. (A) For larger feet, a larger sensor may be needed. In that case the sensor bed can be made larger and the sensors themselves of greater area. By increasing the size of individual sensors, the number of channels needed is unaltered and the same circuit board/architecture can be used. (B) For smaller feet, a smaller sensor bed may be needed. The sensor bed can simply be decreased in size without changing connector footprints, board design, or any other feature. The sensors themselves can be scaled up or down (by increasing or decreasing spacing between fingers) and the sensor bed footprint can be scaled as needed. With 704 sensors in its native configuration, the sensor density is such that the increasing sensor area represents a miniscule change in sensor density.

Affordability: The system is extremely affordable. For a full system that includes a pair of sandals, sensor beds, housings, batteries, and cabling, the cost is only $ 600 ($ 300 for one unit).

Costs could be reduced for the Samtec cabling and connectors. As quantities scale up and a supplier is prepared for larger runs, costs for individual units will decrease.

Sensors and sandal pieces represent the items that will most likely require replacement from wear and use. Some consideration should be given to budgeting for replacements parts, but the costs of sensors and sandal pieces are still very low compared to comparable systems.

Both the battery and associated USB cable could be replaced with cheaper components without much loss of quality. The components presented here were chosen for their quality and durability, but cheaper replacements can be found.

Low resource requirements: A computer of minimal processing power and an SD card are the only barriers to entry for this product. We plan to provide the necessary SD cards as part of the system package. Keeping resource requirements low drove, us to avoid the more commonly used wireless communications seen in other commercial systems. To run the software required to interpret data from the system, a computer must support Java, have 256 K of RAM, and a CPU more advanced than the Pentium 4. When considering visualizations, the compiled MATLAB routine requirements are much more nebulous; no graphics cards are required, but at least 2 GB of RAM are necessary. Computers in the $ 100-$ 200 range are more than capable of interpreting data from our pressure sensing system. For most users, no additional hardware will be required.

Interchangeability: Our data cables are designed to be easy to replace, with replacements available from Samtec with a minimal lead time. Sensors are also fairly easy to replace; a two layer design was retained to ensure lower costs and lead times.

Custom programming: The Teensy 4.1 and Mega 64 boards both rely on the Arduino architecture and programming language, allowing users the opportunity to use the board itself as a platform for multiple diagnostics. With a minimal knowledge of programming and limited time, users can change the function of the microcontroller and board from a plantar pressure sensor to anything the user can obtain a program for. Though sensors and other various hardware would be needed, the open source nature of the microcontroller allows the multiplexed design to serve as a springboard for other analysis.

More advanced users can program their own routines and have a 16 analog/44 analog pin platform at their disposal.

Sensor number: With 704 individual sensors, our platform far outperforms insole systems that possess 9-15 sensors. Our system matches many stationary pressure sensor densities while allowing for dynamic measurements. We provide the same density as static systems while allowing locomotion.

Minimizes changes to gate: We seek to preserve a user's natural gait during measurements. Weight of the system is kept low with lower density, plastic components at the foot and a fairly lightweight container at the ankle or calf. Beyond that an unyielding, flat surface is maintained underneath the foot so that natural foot shape and weight distribution is preserved during analysis. While measurements taken in a patient's preferred footwear are of great use for athletes and patients with acute conditions, monitoring weight distribution on a stable platform over time is of greater value for preventing diabetic damage to a foot.

Our preference for an unyielding platform eliminates potential changes to foot pressure and weight distribution caused by a patient's footwear or the insoles required by other systems.

Mobility: While we have avoided wireless communication for cost and equipment requirement reasons, our decision to include an SD card allows users to walk and move freely. Our system provides both standing and walking foot pressures to the diagnostician. With the SD card, users can walk at their own pace and capture data without excess cabling or concerns about wireless transmission range. With the data capture system being entirely located on the user during use, patients can walk on any surface, indoor or outdoor, and provide data in their preferred environment.

Ease of use: Using the system requires a minimal amount of training or skill. If one can tighten a Velcro strap and monitor an LED, one can use the system. Data capture is fully automated, and the software easily pulls in data from the SD card with minimal user input.

Sampling and sensor density: With 704 sensors each measuring roughly 0.250 inch square, the reading density is already quite high.

Data simplicity: While the enormous sensor density and rapid read/record rate provides the diagnostician with a wealth of data to analyze, the data itself can be easily understood by the patient. No special analysis or interpretation is required for the layman. With a few mouse clicks, data goes from an array on the SD card to an easy to understand display showing a pressure map of the foot during testing. Simple color coding shows the patient areas of high, low, and zero pressure. The data is simple to read directly after a session and, as the pressure readings are relative during each session, easy to interpret over time; from session to session, the patient can easily spot differences in pressure distribution and whether their gait is changing.

Allows motion: The system allows for both static and dynamic foot pressure measurements. Measurements while standing are commonly taken, but the system also allows for various "poses", where the foot and a patient's weight are placed into a chosen configuration for analysis. This allows for "snapshots" that complement dynamic measurements. Walking is what the system is presently designed for, but the electronics and sensor would function during running, jumping, or other athletic activity. If measurements of plantar pressure during activities more intense than walking are desired, refinements would be required for the sandal and attachment portions of the design as the brittle, unyielding plastic would hamper rapid movement and likely break under heavy impulse. Our testing indicates that while the design will hold up during walking, but more intense movements will cause pressures that the plastic will not be able to handle.

Ruggedized/Interchangeable parts: Within the motive of low cost and open source, replacement parts for the system are not proprietary (save for the microcontroller board and the sensors themselves). As the board and sensors are of a custom design, they are not available from any other source. Cables, batteries, data recording, and board components are all readily available from commercial sources. Users are able to service their own units easily and affordably continued support of the systems would not be a source of revenue by design.

Conclusion

In this work, we have presented a low cost, rugged system for plantar pressure sensing, recording, and analysis. The system focuses its efforts on a high sensor density, overall durability, and open source nature of both software and firmware. While the system is not as sleek as modern wireless units, it is elegant in the unique sensor design, high sensor density, and rapid refresh rate. Our units offer a high throughput solution at $ 600 versus the cost of most modern systems of comparable design ($ 20000 or more); in addition, our sensor density far outmatches similar units.

The system presented allows for data recording from a stationary position (whether standing or with the foot in a specific position) or during a walking gait. Recorded data is easily transferred via SD card to a separate system for analysis where density maps can be generated for stationary data and with a high sampling rate during motion. Mapping is easily understood by patients with no medical training and can be interrogated by physicians to draw broader conclusions about foot condition and weight distribution. When data for a single patient is observed over time, subtle changes become apparent before permanent damage occurs or conditions produce major effects in the patient. Data recording is accomplished via an open source Arduino code and analysis through an open source Matlab executable; both the recording and analysis software are designed to be provided at no cost and run by users with only very basic computing capabilities. The collected data can also be processed for visualization using several open source software's.

Each system offers a full solution to plantar pressure sensing, requiring no additional tools or parts from the patient or medical provider. Footwear, attachment materials, sensors, cables, and hardware are all part of the package. For the impoverished or lesser equipped clinic, only the presented system and a cheap computer are required. Such a system allows for developing countries, charities, and poorly funded medical establishments to offer plantar pressure analysis at minimal cost. Beyond the indigenous target audience, our system allows any clinic to offer a low cost, rugged plantar pressure analysis to a broad audience with a minimal investment.

As more and more of the world grows susceptible to diabetes, the need for a low cost plantar pressure sensing system is apparent. With half of the United States at risk for diabetes and much of the world becoming obese, diabetic foot damage is a growing concern. We have presented a preventative solution that is affordable to a broad audience.

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