# A Piezoelectric Energy-Harvesting Shoe System for Podiatric Sensing

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*Abstract*— **This paper provides an energy-harvesting, shoe-mounted system for medical sensing using piezoelectric transducers for generating power. The electronics are integrated inside a conventional consumer shoe, measuring the pressure of the wearer's foot exerted on the sole at six locations. The electronics are completely powered by the harvested energy from walking or running, generating 10-20 $\mu$J of energy per step that is then consumed by capturing and storing the force sensor data. The overall shoe system demonstrates that wearable sensor electronics can be adequately powered through piezoelectric energy-harvesting.**

## I. Introduction

The continued improvement in size, cost, and power of integrated circuits has enabled entirely new classes of wearable devices. Unfortunately, these new devices are still limited by current battery technology, which in many cases is heavy, expensive, and unable to store sufficient power for long-term biomedical sensing. Hence, wearable devices powered by an alternative method, such as energy-harvesting, are desirable.

Of all the locations on a human's body, the feet experience the highest levels of mechanical and kinetic energy during normal use. Therefore it makes sense to embed energy-scavenging within a conventional shoe. Furthermore, due to the large numbers of injuries and conditions associated with high-intensity athletic activities, podiatric sensing is a natural application for this harvested energy sensor system.

Previously, there has been a limited amount of piezoelectric, energy-harvested, shoe-mounted bio-sensing systems. However, there has been research on energy-harvesting using piezoelectrics, including [1], [2], [3], and [4]. Most notably, the Responsive Environments Group at the MIT Media Laboratory created a system that utilized the energy harvested from two piezo transducers to broadcast RFID signals [1]. Their system utilized custom PZT (Lead zirconate titanate) and PVDF (Polyvinylidene fluoride) transducers to produce 1.3 $m$W continuously (at a 0.9 Hz pace). While their system generates greater power, our system utilizes limited, commercial off-the-shelf (COTS) harvesters to achieve similar goals.

There has also been some previous work on mobile, podiatric sensing (such as gait analysis). Stacy Morris of MIT developed a wireless, shoe-based, gait analysis system [5]. The system utilized Force sensitive resistors (FSRs) and PVDF transducers for sensing both static and dynamic forces. In our case, we utilize only FSRs to record gradual changes in pressure, since our sampling frequency is low. Similarly to their implementation, we use a heel-strike sensor (PZT) to determine when to start sampling the FSRs.

While our solution uses COTS piezoelectric harvesters, there are many other methods for energy harvesting. [6] provides a survey of some different methods, including electromagnetic, electrostatic, and more. Additionally, [7] presents a shoe-based energy-harvesting method using dielectric elastomers achieving $\tilde{1}20$ $m$J per step. We chose piezoelectric harvesters due to their COTS availability, small form-factor, and potential for mechanical energy capture.

In this paper, we demonstrate a podiatric sensing shoe system that is powered completely by the movement of the wearer. Off-the-shelf electronics are used for energy-harvesting capability, and to obtain distribution data of foot pressure. The proposed system is also vertically integrated, including not only the hardware, but also the coordinated visualization and database back-end.

The paper is organized as follows. Section II will highlight each of the subsystems and their functionality. Section III will discuss the preliminary results of this prototype system, and Section IV which will specify some of the further testing and future capabilities necessary to move this research forward. Finally Section V will conclude this work.

## II. System Description

### A. Energy Harvesting

The energy-harvesting capability of this system was designed to maximize energy capture by harnessing multiple excitation sources. Since piezoelectric transducers produce electrical energy only by physical deflection, we sought to harness energy from both foot strikes and bending. As seen in Table I and Figure 1, there were two piezoelectric transducers utilized — a rigid energy-harvester and a flexible energy-harvester. The rigid transducer was enclosed in a low-profile, custom 3D printed enclosure that allowed it to vibrate freely without breaking. The flexible-energy harvester was placed strategically at the ball of the foot to maximize foot strike excitations as well as bending excitations, via downward compression and foot flexion, respectively. Ultimately, the goal is to capture otherwise wasted energy generated by the natural movements of walking, running, and general athletic activity.

### B. Power System

This subsystem contains the power conditioning circuitry that allows the shoe system to operate. Since the exact amount of energy harvested by the piezoelectric transducers was unknown during the design phase, the power circuitry was created with operational flexibility in mind. Three distinct operating modes were chosen in order to allow for

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TABLE I
PRIMARY SYSTEM COMPONENTS

| COMPONENT | PART NO. | DESCRIPTION | ACTIVE POWER |
|---|---|---|---|
| **Energy Harvesting** | | | |
| Mide Volture - PZT Piezoelectric Element | V25W | Rigid Vibrational Transducer | $20 - 40$ $\mu$W † |
| Physik Instrumente Durract - Processed PZT | P-876.A11 | Flexible Piezoelectric Transducer | $5 - 10$ $\mu$W † |
| **Power Circuitry** | | | |
| Linear Technologies - Integrated Circuit | LTC-3588-1 | AC-DC - Piezo Power Conditioning | $1.75 - 45$ $\mu$W |
| Cymbet Enerchip - Integrated Circuit | CBC-3150 | Solid State Battery & Power Control | 11.55 $\mu$W |
| CDE Acrylic Capacitors | FCA1210C105M-G2 | Low Leakage Energy Storage | -NA- |
| **Microcontroller and Communication** | | | |
| Texas Instruments - Microcontroller | MSP430FR5739 | CPU & Data Storage | 6.44 $m$W |
| FTDI - Integrated Circuit | FT232RQ | UART to USB 2.0 | 49.5 $m$W |
| **Sensors** | | | |
| Tekscan - Resistive Sensors | A201 | Flexible Force Sensors | $0.098 - 0.99$ $m$W |
| CUI Inc. - Piezoelectric Diaphragm | CEB-35D26 | Passive Piezoelectric Sensor | -NA- |

† Highly dependent on: frequency of steps, user weight, transducer loading, cantilever tip mass, general mechanical stress/deflection.

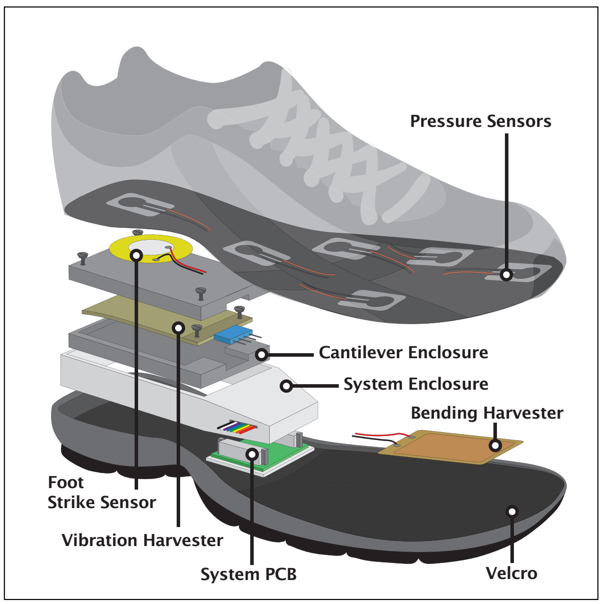


Fig. 1. Expanded view of the shoe system and all integrated components.

flexible duty cycling of data capture. Mode-1 characterizes the system in a fully awake state that is capturing sensor data, and is powered completely from the stored piezoelectric energy (while going into an additional sleep state between samples). Mode-2 characterizes the system in a sleep state, where the harvested piezoelectric energy is not sufficient to capture the sensor data. Both Modes 1 and 2 occur while the user is wearing the shoe system and is exhibiting some kind of foot movement. Finally, Mode-3 is defined as the non-user mode, where the shoe system is connected via USB for data download and is no longer capturing sensor data.

In order to power the system in all three modes, the power circuit uses the two integrated circuits and low leakage capacitors annotated in Table I. The first chip is a Linear Technologies IC that provides a combination of both an AC/DC rectifier and buck/boost converter. The buck/boost converter is chained to low-leakage acrylic capacitors that store the captured piezoelectric energy during Mode-2 until sufficient charge is accumulated, and then consumed during Mode-1. A CBC-3150 (Enerchip) provides logic signals as well as a small 50 $\mu$Ah battery which is used in Mode-2 (sleep state) to keep the time. During Mode-3 (USB operation), the power provided by the USB is utilized to run the system during data download as well as recharge the solid state battery (Enerchip).

### C. Microcontroller and Communication

The microcontroller subsystem is the main hub for controlling communication, storage, and processing within the system. The main control signals of the microcontroller include: sensor ready signals that indicate if the wearers foot is on the ground and thus whether sensor readings should be captured, power ready signals that indicate when there is enough energy accumulated to perform data read and store, and finally USB communication controls that send and receive data while in Mode-3.

During the design phase, it was deemed extremely important that the microcontroller consume ultra-low power, as the energy-harvesting system only produces power on the level of microwatts. Furthermore, the microcontroller also needed to incorporate at least six ADC channels to allow for parallel conversion of the sensor data rather than serial, in order to reduce the amount of time spent in energy-consuming Mode-1. A UART interface was also required in order to communicate with the USB chip in our system. Lastly, the microcontroller was required to incorporate non-volatile data storage, as off-chip data storage (such as Flash RAM) demanded too much power. The specific Texas Instruments microcontroller that was used can be seen in Table I. This microcontroller provides the ability to store 16 KB of data in on-chip FRAM, consume only 1.2 $\mu$A (3.3 V) when idle, while incorporating 33 general purpose I/O pins, a UART interface, and 14 ADC channels. Therefore, it sufficiently meets all of the above requirements. Since FRAM is still an emerging technology, limited sizes are available (maximum of 64 KB); however, with our data storing implementation (4 B for timestamp, 2 B for each sample) allows for approximately 1000 steps to be stored (assuming 5 samples per measurement).

### *D. Sensors*

The sensor subsystem consists of the final data-capturing circuitry, and has two main purposes. The primary function of the sensor block is to provide an analog signal for each of the resistive force sensors placed within the insole of the shoe. Force exertion is converted to a voltage signal via these sensors, with this data routed to the I/O pins of the microcontroller for processing and storage. In the final prototype, we use flexible sensors (shown in Table I) that allow for measurement of both static and dynamic forces. Flexibility was important in order to maximize safety, minimize walking impediment, and prevent the sensors from breaking when in use.

The second function of this block is to provide an interrupt signal to the microcontroller, asserting when the foot is placed on the ground. The sensor chosen to provide this "sensor-ready" signal is a passive piezoelectric sensor. This sensor is ideal because it requires no power or amplification, and it senses dynamic force only. This latter characteristic makes it ideal for sensing foot strike and liftoff.

### *E. Computer Software*

In Mode-3, all data from the device is transferred to custom PC software via USB. The software is responsible for processing the sensor data, storing it in a database, and displaying various data visualizations (pressure map, graph, datatable) to the user. Post-processing is performed to convert the 12-bit ADC values to the corresponding force value in Newtons. Each set of samples is saved with a timestamp for later visualization. During analysis, the pressure map is the most useful representation (Fig.2), which shows a RGB gradient of the pressure mapped onto the sole. The data can be redisplayed in the time-domain using a timeline, showing the pressure transitions of the foot within the shoe. In addition, when the shoe is connected to the computer the user can change any programmable settings (such as sampling period and sample size) and view/record sensor data in real-time.

## III. Results

### *A. Power Generation and Delivery*

The complete prototype, including the energy harvesting and power circuitry, was able to sufficiently power the shoe system at different sensor capture rates, depending on the activity of the wearer. For typical walking situations, approximately $10-20$ steps were needed during Mode-2, in order to have enough energy to enter into Mode-1 and capture sensor readings. During running, duty cycle relating to the number of steps required for each sensor reading was varied between 1 and 5 steps. These step numbers would likely vary slightly depending on the variance of individual user. In our lab tests, a 90.7 kg male of height 1.8 m completed the tests. Because

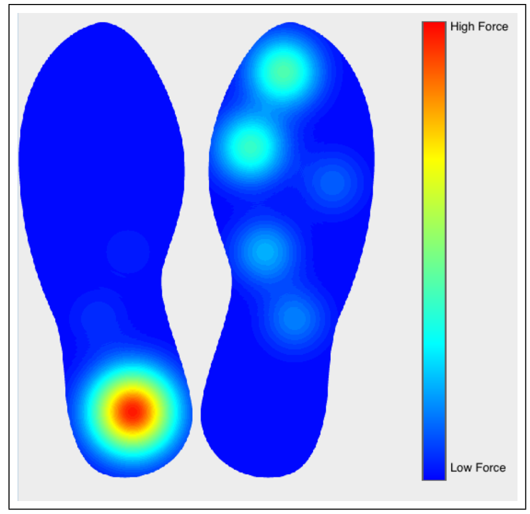


Fig. 2. Custom GUI visualization of force distribution of the foot while walking. Note the location of the 6 sensors.

the piezoelectic elements utilized are characterized according to their frequency of oscillation, it is difficult to exactly quantify the energy capture, as human movement consists of a superposition of many frequencies. Sudden changes in movement can cause both constructive and destructive interference during oscillation. Despite high variability in operating amplitude and frequency, we consistently observed an average of $10-20$ $\mu$J of energy capture per step. Since this value is based on counting the capacitor charge, these measurements also include all efficiency and parasitic losses in the integrated circuits, shown in Table I.

One example of a charging test is depicted in the graph seen in Fig. 3. This shows the results of typical walking movement. The raw piezoelectric voltage waveform (labeled and in the color teal) is characterized by an AC oscillation between $-1$ and 6 V. The high-frequency vibrational transducer is seen superimposed on the low-frequency bending transducer. After rectification, the energy is stored on low-loss input capacitors that are used to regulate the input to the Linear Technologies energy-harvesting IC. As charge accumulates, the input capacitors move energy across the boost converter to the output capacitors, which are then regulated by the Enerchip to supply a consistent 3.3 V operating point for short durations of time. The Enerchips power ready signal (labeled and seen in green) alerts the microcontroller when enough charge has accumulated on the output capacitor bank. Hence, the Mode-1 operating state (described in Section II-B) can then be entered to capture the force sensor data.

### *B. Recorded Measurements*

The final prototype consisted of six flexible force sensors placed at the Calcaneus (heel), Cuboid (lower outside), Navicular (lower inside), the head of the first and fifth metatarsals (upper inside and outside), and the head of the proximal phalanx of the big toe (Fig. 2). Various tests were performed, such as jumping, walking, and jogging tests. Jumping tests

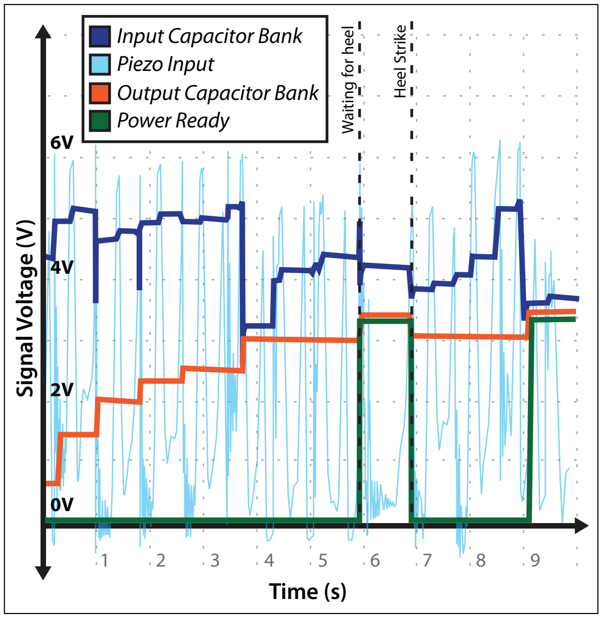


Fig. 3. A time-domain graph of energy capture in lab walking test. Note, the device is usually in Mode-2, only transitioning to Mode-1 for a brief amount of time after heel strike. Piezo energy is consumed in Mode-1 only.

Fig. 4. Completed and functional piezoelectric energy-harvesting shoe prototype (same size and height as unaltered shoe).

consisted of the wearer jumping up and down for 5 seconds in order to verify accurate data capture. The more useful tests consisted of walking or running for a discrete time interval. These tests were conducted in an urban environment, such as on pavement or linoleum. The tests spanned anywhere from 10 seconds, to 15 minutes, and were performed in a casual, conventional daily routine environment. Our recorded results were as expected in terms of relative force. As the wearer walked around, the force transitioned from the heel to the toes, with most of the force occurring on the initial impact of the heel. This is in contrast to running or walking down stairs, where the force is mostly centered around the ball of the foot, with less impact on the heel. While our system measures force in Newtons, actual force measurements inside the shoe were not verified by another device. Therefore, the sensor data recorded is currently qualitative, though with further calibration and refinement of our sensor voltage versus force correlation (currently an exponential best-fit), it is possible to obtain accurate, quantitative force data.

## IV. Future Work

While an initial prototype for podiatric sensing using energy harvesting has been created, there is still much validation and calibration that needs to be performed in order to obtain precise (absolute) medical analysis. The sensors need to be calibrated within the shoe and validated against conventional methods for podiatric/gait analysis. Similarly, the software could be improved to include features requested by the medical community who may be using the device and software (e.g., physical therapists or sports scientists). Other improvements include increasing the energy capture efficiency as well as optimizing the piezoelectric transducers to capture the most energy possible while minimizing parasitic losses in the circuitry. Improving the energy-harvesting efficiency would enable wireless data collection, such as Bluetooth 4.0 Low-Energy, thereby enabling smart-phone applications or wireless, real-time monitoring of foot pressure.

## V. Conclusion

The system described in this paper combines novel energy-harvesting techniques with force-based sensors to deliver an innovative solution to conventional in-lab equipment. The system is designed to be robust, mobile, and fully embedded in the patients normal routine, allowing for podiatric analysis in a variety of environments. Due to the low-volume and low-maintenance features, the device can be targeted for athletes, physical therapy patients, amputees, and those with muscular or nervous system disorders.

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