CONDUCTIVE POLYMER SENSOR IN-SHOE DISCRETE PLANTAR PRESSURE MEASUREMENT SYSTEM

ECE 314 FINAL PROJECT

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Abstract

The time/pressure distribution of a foot in motion has been the study of many scientists over the past twenty years. In this article, a system was built to measure these pressures. Piezoelectric sensors convert foot pressures to a resistance, which is placed in a voltage divider. The ensuing voltage drop is filtered at a high corner frequency of 50 Hz, and recorded in an oscilloscope through HPVEE. The system was used to gather data at 100 Hz from the gait of one female subject, moving at rates of 60 beats per minute, 120 bpm and 180 bpm and using shod and bare feet. Although the pressures experienced by individual sensors differed from results in the literature, an overall trend of greater mid- and fore-foot pressure with increased speeds was observed in the bare foot case. For the shod foot, there were high pressures in the heel region at low speeds, but an uncharacteristically large pressure in the toe was present for all three speeds. By considering the differences between the shod and bare foot data, a virtual transfer function of the shoe was calculated. From this function it can be seen that the shoe attenuates all frequencies equally. Sources of inaccuracy include sensor attachment, measurement synchronization, locomotion difficulties induced by the test, and the low repeatability of the trials.

Introduction

Spear in hand, a bushman ambles through an otherwise empty expanse of the Kalahari. Cradling a mobile telephone, a socialite darts across a crowded Manhattan intersection. For millions of years before and quite possibly for millions after the advent of planes, trains, and automobiles; mankind has depended predominantly on its feet for a vast range of daily activities. Because these primary means of locomotion often fall victim to afflictions both genetic and acquired; both the general public and the scientific community possess a keen interest in the pressures to which they are subjected and in how these pressures vary with factors such as shoe design and walking surface. Gaining such insight enables researchers to develop methods and devices for minimizing the risk of foot ulceration in diabetic patients; for monitoring the effects of degenerative foot diseases such as leprosy; for assessing the effectiveness of pressure relief insoles; for analyzing changes in gait due to injuries and to deformities such as spina bifida; for investigating the efficacy of reconstructive surgery; and for evaluating variations in load distribution as a result of limb length discrepancy (Cobb and Claremont).

The types of sensors deployed to confront these challenges are as diverse as the feet they measure; and since each approach offers a unique set of benefits and limitations, no single technique is appropriate for all possible applications. For example, the preeminent Kistler force plate achieves high specification, good repeatability, and long-term stability, but cannot measure plantar load distributions; pedobarographs render unequaled resolution for barefoot measurements, but are compromised by extremely temperature-dependent sensitivities (Patil and Srinath); the seven discrete sensors of the Electrodynagram allow both barefoot and in-shoe measurement, but also exhibit high nonlinearity, hysteresis, and drift (Cobb and Claremont); and a system consisting of discrete piezoelectric ceramic transducers provides moisture shielding, uniform load distribution, and minimal sensitivity to lateral strain, but requires calibration of individual transducers (Nevill *et al*.). Even a piezoelectric copolymer film configuration that boasts a measurement range of 0-1 MPa, linearity to within 1.5%, a hysteresis error of less than 1.5%, sensitivity of ±1 kPa within the calibration range and a worst-case reduction of 3%, and a frequency response of 0.008 to 250 Hz is restricted from widespread clinical use by the need to fashion insoles for each individual subject (Cobb and Claremont).

Differences in calibration methods and inconsistent definitions of such terms as "force," "pressure," and "load" make it difficult to quantitatively compare results from myriad methodologies, but qualitative trends from past studies generally agree.

The objective of this project is to utilize conductive polymer sensors in qualifying discrete in-shoe plantar pressure measurements for a single, female subject during both barefoot and shod walking and for speeds varying from a slow walk to a jog. These results are then compared to known data.

DESIGN

Sensor

Selection

In part for its economic viability, but primarily for its ready accessibility, the Liberty Touch Pad was selected for this project.

Characteristics

Inside the sensor, a series of interdigitated, metal "fingers" sit above a conductive polymer whose resistance is:

$$R = \rho * L / A,$$

Where ρ, L, and A are its resistivity (Ω-m), length (m), and internal contact area with the "fingers" (m^2), respectively.

The sensor always measures pressure, but if external contact area with the load is held constant, can also measure force. When the sensor is unloaded, the internal contact area is zero, theoretically generating an infinite resistance; in practice, however, the resistance is simply very large. As the applied load increases, so does the internal contact area, thereby decreasing sensor resistance.

When a soldering iron increases the ambient temperature of the sensor, its resistance increases perceptibly, but not significantly, thus indicating a low temperature coefficient. This feature is particularly advantageous because the linoleum floor and insole against which the sensor is tested differ significantly in temperature.

Using available equipment, it was not possible to calculate the resolution of the sensor; doing so requires instrumentation that can both hold a constant pressure and change that pressure by small, known increments. However, this drawback does not affect the qualitative focus of this project.

Bending the sensor brings the "fingers" and polymer into contact and, thus, decreases its resistance. Because a foot is a curved surface, this characteristic offers both the advantage of increased sensitivity and the disadvantage of requiring extremely accurate calibration techniques to prevent measurement error.

Calibration

The transducers used in this study change resistance as a function of the average pressure exerted on them. The resistance of the transducer at any point in time is estimated from the voltage drop across the transducer, while assuming knowledge of the power supply and the resistance of the other resister in the voltage divider. Once the resistance is estimated, a systematic method was used to infer the average pressure exerted on the transducer as a function of the estimated resistance. Empirical data was collected that develops a relationship between the resistance of the transducer and its average pressure. Plot 1 represents that imperial data. Note the log-log scale on each axis. The extreme non-linear nature of the transducer was motivation for the choice of scale. Even when plotted on log-log scale the relationship displays a superlinear trend. In actuality the pressure/resistance relationship is "superlogarithmic".

The calibration process was conducted in a different manor than that conducted in the ECE 315 laboratory. In our procedure the transducer was placed on a scale. The scale was then zeroed. Pressure was distributed across the sensor by the thumb. Incremental pressures were applied to the sensor. At each increment that resistance across the sensor was recorded. When comparing the data collected from this method to that used in the ECE 315 laboratory the results differed by a factor of two. The authors believe that our calibration method is superior due to the fact that type of loading that we put on the sensors (thumb pressed against the sensor) is closer to what the sensor will see.

Once the calibration data was established, a method of interpolation was needed to determine pressures across a continuous range. Attempting to fit least-square functions to the data had little success. Due to the extreme non-linearities of the data, a single function could not adequately represent the data. By far the best alternative was fitting a simple linear spline between each of the data points.

Placement

Previous studies indicate that while walking, the hindfoot experiences higher and more sustained pressures than the mid- and forefoot. As speed increases, more pressure is exerted on the lateral four toes at the expense of the medial midfoot. The fact that feet have thicker heels than metatarsal pads implies that human beings were physiologically designed for walking. While running, the following areas – listed in descending order – encounter maximal peak plantar pressures and plantar pressure integrals: second metatarsal head (MTH), first MTH, third MTH, great toe (Guten).

Based on the results above, sensors were placed on the first, second, and third MTHs, on the great toe, and on the inner and outer heel. Relocating the sensors to vary lateral and longitudinal coverage would enable the assessment of foot conditions such as pronation and flat-footedness.

<u>Amplifier</u>

Because changes in applied pressure induce changes in sensor resistance, each sensor was first incorporated into a voltage divider circuit containing a 10-kΩ series resistor, so chosen because its value lies approximately in the middle of the range of the sensor (Figure 1).

The linear working region of the sensor can be increased by varying the value of the series resistor, but doing so sacrifices resolution.

Based on previous experimental results (Zhu *et al*.) and on sampling theory (Nigg), a sampling rate of 100 Hz was chosen. To minimize interference from high-frequency noise, the voltage across each sensor was fed through a low-pass filter with unity gain to avoid saturating the output. Based on the aliasing phenomenon associated with the Fast Fourier Transform, the cut-off frequency was established at approximately 50 Hz:

$$f_c = 1 / (2\pi * 1.15\ k\Omega * 2.71\ \mu F) = 51.1\ Hz$$

Finally, the output from each filter was connected to an oscilloscope input.

Data Acquisition System

Initial Approach

The initial data acquisition system was comprised of the HP E1347A multiplexer (MUX) and HP E1326 digital multimeter (DMM), both of which are VXI instruments controlled by HP VEE software. The voltage drop across each sensor was read into MUX channel 0, 1, 2, 3, 8, or 9; after which a sequencer in VEE successively cycled through the switches, closing one at a time, then sending the signal to the tree bus via switch 90. The data were then transferred from the tree bus to the DMM, which measured the voltage, the results of which were stored in an array in VEE.

The primary benefit of this setup was that data from all six sensors could be simultaneously gathered during each sampling cycle of each trial, thus enabling easy observation of the timing relationships among discrete pressures at different locations. In addition, the data could be manipulated into any organizational format desired; and the DMM would provide accurate measurements.

The first obstacle presented by this configuration was that the MUX was hard-wired with various impedances for previous lab use; thus, only three channels were available. This problem was easily solved, however, by removing said impedances and constructing a new cable interface to the MUX. In addition, the time overhead for the software limited the sampling rate to 1 Hz. After optimizing the software for both the sequencer and user-defined functions, the maximum sampling rate increased, but still fell far short of the desired 100 Hz. Consulting Prof. Norm Miller of the Department of Mechanical and Industrial Engineering at the University of Illinois, however, eventually revealed that the relay MUX simply cannot sample at such high frequencies.

Finally, when this system was tested using a +5-V source at its input, the MUX output values were measured inaccurately: although only one distinct switch was closed, all channels seemingly outputted the 5-V signal. Occasionally, despite the +5-V input, all channels would output 0.1 V. After deploying a variety of different tree buses continued to yield inaccurate results and after several months of effort, this approach was abandoned.

Final Approach

The data acquisition system eventually utilized (Figure 2) consisted of the HP 54600A oscilloscope, which was manipulated using VEE. The voltage drop across the two sensors was input into Channels 1 and 2 of the oscilloscope. The oscilloscope was configured to take 500 samples with a time base of 500 ms:

500 samples / (10 units * 0.5 seconds per unit) = 100 Hz

VEE captured the waveforms as a list of data points and wrote them into files.

The advantages of this system were its straightforward configuration and the ease with which it was debugged. Aside from several instances of conflicting settings and determining exactly how to manipulate the sampling frequency, no real problems were encountered. Its drawbacks, however, far outweighed its benefits: only two sensors could be tested per trial, thus requiring more trials and sacrificing the determination of exact time relationships among peak pressures at the six recording points. Tester synchronization enabled an approximation of the time-pressure continuum that was still far from exact.

Sensor Placement

The areas of maximal pressure (both peak pressure and duration of pressure) experienced by the running foot were (in order), the second metatarsal head, the first metatarsal head, the third metatarsal head, and the great toe. The heel bore substantially less pressure. It should be noted that as a runner increases speed, more pressure is exerted on the lateral four toes at the expense of the medial midfoot. While walking, a foot experiences more pressure at the heel. The foot is physiologically designed to have a thicker heel pad than metatarsal pad, implying that humans are designed to walk (Guten, 20).
Using this research, the sensors were placed at the first, second, and third metatarsals, the big toe, and the inner and outer heel (Figure 3). Also, with a wide

range of lateral and longitudinal area covered, pronation and supination measurements could be made.

Procedure

To ensure exact placement of the sensors, the subject's foot was labeled with chalk and pressed into the test shoe (Figure 4). The appropriate points were labeled with pen.

Tests were performed for two cases: the subject wearing Nike Zoom Air (cushioned heel. Figure 5) running shoes, and the subject walking barefoot. For the former configuration, the sensors were duct-taped to the insole of the left shoe. The wires were fed out of the shoe and fascinated to the subject's entire leg using rubber bands. The subject held the wire while moving to avoid tripping. For the bare-foot condition, the sensors were duct-taped to the sole of the subject's left foot, and the wires were configured as described above.

The test subject, a 21-year-old female, propelled herself across the test field at rates of 60 steps per minute (slow walk), 120 steps per minute (fast walk), and 180 steps per minute (run). After first attempting to use Lauryn Hill's "Doo Wop" song to keep a constant walking rhythm, a metronome was used to ensure that she maintained these rates. For each rate, three trials were performed. This procedure was executed twice, with and without shoes.

The female moved for a total of five seconds during each trial. Her ending location was measured, and her velocity recorded. Speed influences pressure distribution (Nigg, 20), and thus it was important to take the velocity into account when analyzing data. The standard deviation of velocity was usually greatest during the 180 bpm trial, and smallest during the 60 bpm trial. The difference in standard deviations is larger for the barefoot trials than for the shoe trials.

To take in the desired window of data points in HPVEE, it was necessary to understand the recording process of the oscilloscope. Once the "start" button of the program was pressed, the software spent approximately one second retrieving data, and then displayed the waveform from the past five seconds in HPVEE. Thus, upon commencement of locomotion (synchronized at the start of locomotion with the metronome), a four-second delay was observed before the "start" button was depressed.

Results

The effects of velocity on pressure

When comparing the sensor measurements taken at 60 bpm and 120 bpm, there seems to be little variation.

For the trials with the running shoe, the outer heel exerts more pressure over a longer period of time than do the inner heel and third and second metatarsals. The first metatarsal has the same magnitude and longevity of the outer heel. The big toe experiences larger forces than any of the other parts of the foot, and with a longevity equal to that of the inner heel. This data does not agree with the above research, which stated that the metatarsal would receive the greatest force over the longest period of time.

When comparing the 60 bpm and 120 bpm measurements to the 180 bpm measurements, there are obvious differences. The duration of pressure exertion is now larger for the first, second and third metatarsals, and the magnitudes of these pressures are greater than at the slower rates. This confirms the above research, which states that at faster speeds, pressure is transferred farther up in the foot. The heel shows a marked decrease in duration of pressure from previous trials.

For the barefoot trials, the trends are much different. At 60 bpm and 120 bpm, the maximum pressure and force duration seem to be the same for the inner and outer heel, and the first, second, and third metatarsals. For 60 bpm, there is comparatively no pressure on the toe, which contrasts greatly with the large pressure seen in the shoe trial. At 120 bpm, the toe pressure has increased to rival the other sensors. At 180 bpm, the inner and outer heel and first metatarsal show a lower peak force and smaller pressure window, while the second metatarsal now bears the most pressure, as predicted by Guten. The toe and third metatarsal seem unchanged.

In summary, the shoe seems to lessen the pressures felt by every part of the foot except the big toe, which experiences a vastly larger pressure. As speeds increase, the bulk of the pressure shifts toward the middle and upper foot.

Pressure impulse

Some of the literature discusses the topic of the "pressure integral" or "pressure impulse". It is defined as the area under the pressure curve of a time history, and is thus the integration of pressure with respect to time. An impulse can be defined as the integration of force with respect with time. Pressure can be interpreted as a normalized force with respect to effective surface area. Investigation of the nature of an impulse reveals that an impulse is simply a change in momentum. Likewise, one may interpret the pressure impulse as a change in momentum per unit area.

A pressure impulse calculation was performed for each of the runs, and the results are summarized in the following table:

	shoe		
	pressure impulse [kPa*sec]		
sensor	**60 BPM**	**120 BPM**	**180 BPM**
1	**22.7**	**8.8**	**4.1**
2	**14**	**3.4**	**4.1**
3	**9.6**	**5.2**	**6**
4	**8.8**	**5.4**	**5.1**
5	**29.1**	**5.4**	**4**
6	**120**	**57.9**	**104.6**

	barefoot		
	pressure impulse [kPa*sec]		
sensor	**60 BPM**	**120 BPM**	**180 BPM**
1	**36.4**	**22.1**	**4.1**
2	**47.4**	**19.4**	**6.2**
3	**59.6**	**19.6**	**14.8**
4	**62.5**	**44.6**	**31.9**
5	**47.4**	**24.1**	**4.9**
6	**15**	**24.4**	**11**

The pressure impulse was estimated from numerical integration of the data. A representative portion of data from each data set was used to estimate the pressure impulse to which each sensor was subjected. Noteworthy is the fact that the pressure impulse actually decreases with increasing speed. Initially this may seem counterintuitive. But, as the subject increased her speed, less energy needed to be exhausted to fluctuate her center of mass in the vertical direction. For the center of mass to increase in elevation its momentum must change. This change of momentum is represented by the pressure impulse inferred from the time history of the pressure. As velocities increase, more salience is placed on minimizing non-conservative energy losses. Almost none of the potential energy lost during a decrease in center of mass is ever regained during human locomotion.

Frequency domain analysis

Frequency domain analysis is an excellent compliment to the time domain analysis. Plots 8 through 13 represent a Fourier analysis for each of the test conditions and each of the sensors. Each graph was averaged over 3 data sets in attempt to minimize the noise inherent to frequency domain analysis. The analysis utilized Matlab's fast Fourier transform. Each graph shows the frequency contribution to the signal.

Most of the results agreed with the literature. Much of the data follows similar qualitative trends. Much of the signal's bandwidth is below 20 Hz, as expected. This is consistent with mechanical systems of this scale. The low-pass filter incorporated in the data acquisition process has a corner frequency of 50 Hz. This frequency was chosen to minimize aliasing and noise while maintaining broad enough bandwidth to capture most of the dynamics.

Another salient characteristic present in the Fourier analysis data is the presence of spikes. Initially, these spikes were thought to be some sort of resonating due to exiting some natural frequencies. These spikes could have been picked up anywhere along data acquisition process (caused from shoe dynamics, sensor dynamics, noise, etc.). Further investigation discredited these sources. The spikes occurred at constant intervals. This is most likely due to harmonics of the same phenomenon.

Shoe Dynamics

One of the main objectives of this project is to gain an insight to the dynamics of a running shoe, and how it affects foot/ground reaction forces. The final three plots (Plots 14 through 16) address that issue. These plots represent the attenuation (or amplification) of pressure at each sensor caused by the dynamic properties of the shoe. In this sense the shoe was treated as a "black box" and empirical input-output data was collected to determine its dynamics. The input signal was the pressure between the foot and ground. This was the data taken during the barefoot trials.

The output signal was the pressure between the foot and shoe. So, the ratio of these signals reveal the dynamics of the shoe. An approach similar to that of developing an empirical Bode plot was utilized. The Bode plot, thus, gives insight to the input-output transfer function. While the ratio of the output to input magnitudes at each frequency give rise to the magnitude dimension of the Bode plot, phase could not be determined. Phase losses caused by the shoe dynamics could not be estimated because the input-output data was not taken at the same time. Care was taken to replicate the experiment with and without wearing a shoe, but they could never be close enough to accurately compare phase differences.

A few interesting and a lot of not-so-interesting conclusions were drawn from this analysis. Probably the most obvious observation is the lack of frequency dependence. While the shoe transfer function is non-linear, a simple linear model could be fit to the model. In its simplest form the linear model would be a zeroth order system. This means no energy storage, and the system can only dissipate nonconservative energy. But, without an estimate of the behavior of the phase, conclusive analysis could not be performed.

The attenuated magnitudes across all frequencies can be a double-edged sword. Shoes are designed to absorb shock, which means dissipating energy. It is apparent that this is in fact true. This alleviates potentially traumatic forces the body is subjected to. But, by absorbing shock less force can be transferred from the foot to the ground. Sprinting shoes are designed to dissipate as little energy as possible. While this is not recommended for extended duration, discomfort is compromised for performance.

<u>Limitations</u>

Inaccuracy

There was a great deal of inaccuracy in most of the aspects of this system. All sensors were affixed to the shoe/foot with duct tape. This duct tape introduced a tension to the surface of the sensor, which provided at DC voltage offset in the readings. To solve this problem, a narrow fringe around the edges of each sensor was secured with duct tape, allowing the sensor to remain exposed (Figure 4). However, a better type of attachment is necessary, perhaps with the introduction of a re-usable adhesive on the sensor.

While the insertion of the sensors did not alter the stride of the subject, the barefoot runs affected the stride of the subject. The average velocity (and hence stride) of the barefoot trials was consistently lower than that of the shoe trials, with differences ranging between 0.6 percent and 8.7 percent. This is due to the decreased amount of friction between the duct tape and the smooth linoleum floor, compared to the large amount of friction between the rubber shoe sole and the linoleum. Thus, the runner needed to adjust his/her stride so as to provide greater normal force to the floor. The step rate did not seem to influence the velocity differences.

The choice of shoe may have influenced the pressure measurements obtained. Only the heel was cushioned, and thus the pressures measured by the heel sensors were smaller in comparison to the metatarsal pressures than they normally would be. Measurements were taken as the subject started walking from rest. These initial few steps were uncharacteristic of the subject's normal stride, and thus the data obtained from these steps was inaccurate. In the future, the subject should be walking prior to the data collection window.

Only two sensors could be tested per trial, which meant that not only were more trials necessary, but the exact time relationships between peak pressures at the six sensor points were not recorded. Through tester synchronization, the time-pressure relationships were approximated, but were not exact.

Finally, the actual recording procedure was subjective. Assuming the data retrieval time to be one second, the recorder pressed the "start" button four seconds after the commencement of each trial. However, the retrieval time was never precisely measured. In addition, the exact point when HPVEE stopped accepting new data values and started displaying old data values was unknown. All of these uncertainties caused a loss of synchronization between the data and the measured time intervals, and thus lent error to the synthesis of all six sensors into one comprehensive pressure distribution model.

Repeatability

The plantar pressure measurement system yields a poor repeatability. Often, a trial would inexplicably yield unusual data, and that trial needed to be re-done. When fixing sensor problems, there was little or no way to guarantee exact re-placement of the sensors. Also, subject fatigue and mental concentration greatly altered walking patterns.

As the tests were being carried out, the wires occasionally became dislodged from their sensor contacts, causing poor data collection. Although duct tape was used to secure them, a better method of attachment should be developed. Also, with motion, the wires would touch each other and short out the sensor. Insulation of the wires with duct tape solved this problem.

Flexibility

There are many limitations to the size of the plantar pressure measurement system. The configuration requires a computer, bulky circuit board, and long wires that are conducive to tripping. Thus, it is not easily portable, which limits the applications of such a system.

Data analysis

The effects of velocity on pressure

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In summary, the shoe seems to lessen the pressures felt by every part of the foot except the big toe, which experiences a vastly larger pressure. As speeds increase,

Future tests

The foot sensing system is versatile, and could be used in a wide range of applications. For example, the system could test for pronation and supination disorders, and thus be used to supplement optical tests. By measuring the time change in pressure across the metatarsal heads and the heel regions, the degree of pronation could be measured. In measuring the time change in pressure from the rear heel through the midfoot to the toe region, the strength of supination of the subject could be determined .

Also, the system could be used to test the pressure-dampening effects of various models of shoes, and the true worth of pronation vs. non-pronation insoles.

the bulk of the pressure shifts toward the middle and upper foot.

Conclusion

The system described in this report was successively used to gather data at 100 Hz from the gait of one female subject, moving at rates of 60 beats per minute, 120 bpm and 180 bpm and using shod and bare feet. Although the pressures experienced by individual sensors differed from results in the literature, an overall trend of greater mid- and fore-foot pressure with increased speeds was observed in the bare foot case. For the shod foot, there were high pressures in the heel region at low speeds, but an uncharacteristically large pressure in the toe was present for all three speeds. By considering the differences between the shod and bare foot data, a virtual transfer function of the shoe was calculated. From this function it can be seen that the shoe attenuates all frequencies equally. Sources of inaccuracy include sensor attachment, measurement synchronization, locomotion difficulties induced by the test, and the low repeatability of the trials. This system could be further applied to analyze the presence of pronation and supination disorders, and the values of different models of running shoe.

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