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Wrinkled Thin Film Pressure Sensors for Pulse Pressure Detection

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Wrinkled Thin Film Pressure Sensors for Pulse Pressure Detection

THESIS

Submitted in partial satisfaction of the requirements for the degrees of

MASTER OF SCIENCE

in Biomedical Engineering

by

Nicole Leigh Mendoza

Thesis Committee:
Associate Professor Michelle Khine, Chair
Associate Professor Elliot Botvinick
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I would like to express my sincerest gratitude to…

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ABSTRACT OF THESIS

Wrinkled Thin Film Pressure Sensors for Pulse Detection

By

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Master of Science in Biomedical Engineering

University of California, Irvine, 2016

Professor Michelle Khine, Chair

The emergence of wearable electronics provides opportunity for continuous health monitoring. Current wearable devices are unable to continuously monitor blood pressure, primarily due to poor wearability and sensing limitations of the sphygmomanometer. The development of conformal pressure sensors enables continuous measurement of pulse pressure, offering an alternative to the outdated cuff. Although conformal sensors show promising sensing properties at low pressures, expensive fabrication and unrealistic applications have hindered their adoption. This work provides a simple, low cost, scalable method for fabricating piezoresistive sensors by combining wrinkle-patterned polydimethsiloxane and highly conductive carbon nanotubes. These flexible pressure sensors demonstrate high sensitivity ($0.65 kPa^{-1}$), fast response time (<10ms), and low power consumption, with an operating voltage of ~1V. This work further demonstrates the application of these sensors for real-time monitoring of pulse pressure, as well as their compatibility with wearable electronics.

CHAPTER 1

INTRODUCTION

1.1 Wearable Devices for Health Applications

Wearable devices can be useful for health monitoring applications. The Bluetooth capabilities of wearable devices can provide continuous long-term data, as well as remote patient monitoring. This extended data can give healthcare professionals greater clinical analytics, which can lead to better diagnosis and treatment[1-4]. However, current wearable devices on the market are limited in their sensing capabilities, and fail to capture major health vitals, such as high blood pressure.

1.2 Importance of Monitoring Blood Pressure

One of the principal vitals signs is blood pressure, which is the force exerted by blood onto artery walls when the heart pushes blood through the arteries. Blood pressure naturally varies throughout the day, depending on activity and stress levels. However, an abnormally high blood pressure for long periods of time can cause damaged blood vessels and strain to the heart. This condition is known as hypertension. Hypertension affects one third of Americans and contributes to the number one cause of death in the world: heart disease[5]. Hypertension is known as the silent killer because it carries no obvious signs or symptoms. In order diagnose and monitor this disease, blood pressure must be physically measured.

A typical blood pressure measurement is taken at the brachial artery in the upper arm, and consists of two values: systolic pressure and diastolic pressure. Systolic pressure is the peak pressure that occurs when the heart's ventricles contract and create a sudden influx of blood through the arteries. Diastolic pressure is the minimum pressure in the arteries that occurs between heart contractions. These values are typically reported as systolic pressure over diastolic pressure in millimeters of mercury (mmHg) since arterial pressure, historically measured with the sphygmomanometer, used a mercury column to reflect pressure.

1.3 Traditional Blood Pressure Monitors

The technology to measure blood pressure has barely changed since the invention of the sphygmomanometer, over a century ago. The sphygmomanometer requires an inflatable cuff and a stethoscope to listen to Korotkoff sounds[6]. In a normal person, when a stethoscope is placed over the brachial artery, no sounds should be audible. However, if an inflatable cuff restricts blood flow in any way, Korotkoff sounds are produced due to the turbulent blood flow. The cuff is placed on the upper arm and inflated to completely restrict blood flow to the brachial artery. The cuff is then slowly deflated, and Korotkoff sounds are recorded. Korotkoff sounds can be heard as long as the cuff's pressure is between systolic and diastolic pressure. Therefore, the pressure, at which Korotkoff sounds are first heard, is recorded as systolic pressure and the pressure, at which the Korotkoff sounds cease, is recorded as the diastolic pressure, as seen in Figure 1.1. Modern devices no longer require a stethoscope, because the turbulent blood flow caused by the cuff also produces oscillations. These oscillations can be detected with a transducer and used to determine systolic and diastolic pressures[6].

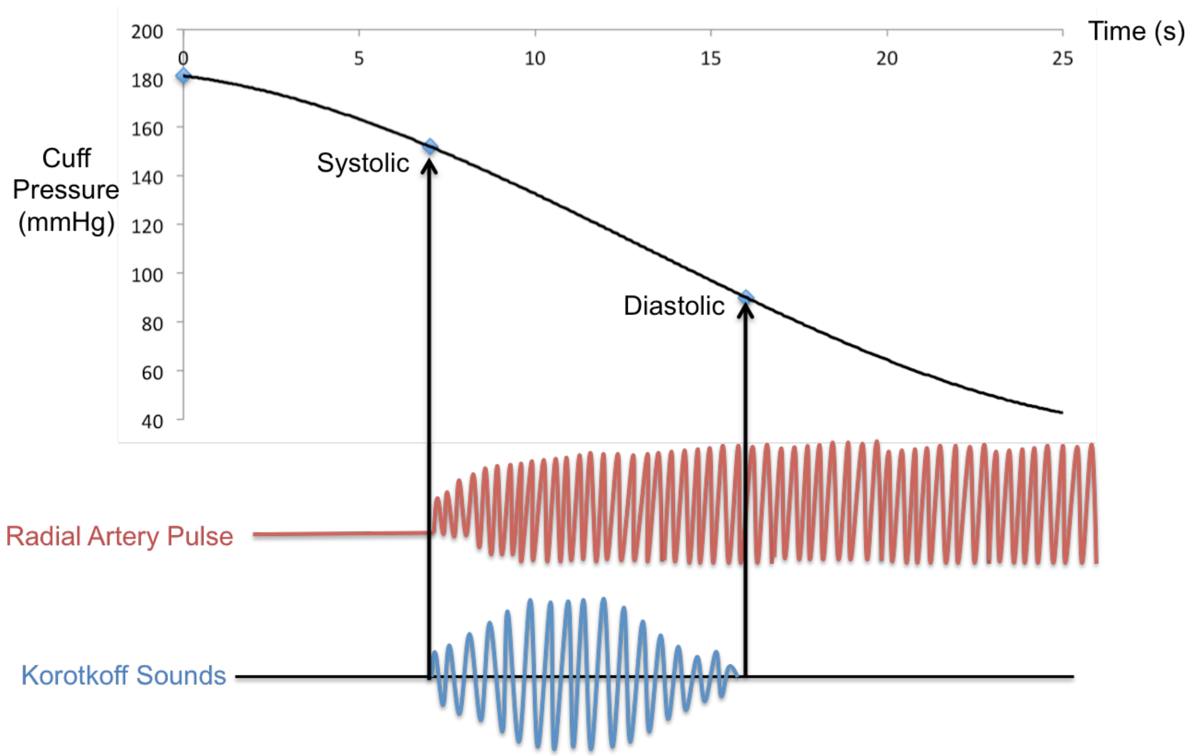

Figure 1.1: A graphic illustration of the sphygmomanometer technique for using Korotkoff sounds to measure blood pressure.

Although the sphygmomanometer has been the gold standard for taking blood pressure measurements for many decades, there are several shortcomings to this method, including discomfort to the patient and the white coat effect. Using an inflatable cuff to completely cut off circulation in the arm can be painful, especially if the measurement needs to be repeated. The white coat effect refers to an increase in blood pressure during a clinic visit – most often due to patient anxiety. Since the white coat effect can lead to a misdiagnosis of hypertension, many patients are required to participate in ambulatory blood pressure monitoring[7]. Ambulatory monitoring consists of blood pressure measurements with the cuff device at repeated intervals over a 24-hour period. Ambulatory monitoring helps give physicians more comprehensive data for basing his or her diagnosis and treatment. However, ambulatory monitoring is also a huge inconvenience for patients, causing bruising and sleep disturbances and leading to poor patient compliance[8].

Therefore a wearable blood pressure monitor - capable of taking continuous measurements - is needed to provide physicians with better diagnosis analytics and to aid patients in managing hypertension. In order to obtain continuous, noninvasive blood pressure measurements, the traditional cuff method must be discarded and new methods must be explored. One such potential method, is improving physiological pressure sensors to detect pulse pressure through the skin without restricting blood flow[9].

1.4 Arterial Tonometry

This method of using pressure sensors to measure physiological pressure through the skin is known as tonometry. Applanation tonometry uses a hand held strain gauge pressure sensor, applied to the radial artery with mild pressure to detect the arterial pressure waveform. The pulse pressure waveform actually gives more insight on cardiovascular function than the central blood pressures alone[10-13]. Central blood pressures can still be derived from this waveform with an FDA approved algorithm[14]. Many physicians are unaware of applanation tonometry, thus its adoption into clinical practice has been slow[13]. Applanation tonometry validates the potential of moving away from the cuff device toward sensitive pressure sensors to measure pulse pressure at the radial artery.

In order to apply the principles of applanation tonometry to a wearable blood pressure monitor, pressure sensors must be optimized. These sensors must be highly sensitive, have a fast response time, and low power consumption[4]. In addition, it is desirable for these sensors to be flexible, lightweight and low cost[15]. Recently, flexible and stretchable thin film polymer sensors have demonstrated great pressure sensing performance[4, 15]. The use of nanostructured nanotubes and nanowires as the functional conductive materials in these sensors has shown the potential to further enhance sensitivity[16-20]. Despite these advances, efforts are still being made to find a cost-effective method for fabricating these types of sensors to be integrated into wearable devices[15, 17].

To contribute to these efforts, this work presents a simple and scalable process for fabricating flexible, thin film, carbon nanotube (CNT), resistance-based, sensors for detecting pulse pressure. The following chapters describe the process of choosing an appropriate sensor design and fabrication techniques for producing highly sensitive, conformal pressure sensors. This includes a highly scalable technique for creating wrinkled multi-scale structured surfaces and patterning polymer thin films. CNT solution is deposited onto these multi-scale structures and the thin films are assembled to create a resistance-based pressure sensor. This thin film CNT pressure sensor is characterized by its sensitivity, cycling behavior, and detection limit. This sensor is proven capable of detecting pulse and compatible with microcontrollers for wearable electronics.

CHAPTER 2

FUNDAMENTALS OF PRESSURE SENSORS

2.1 Applications of Pressure Sensing

Pressure is defined as the force exerted over a particular area. Gravity, touch, and even internal blood flow are all forms of naturally occurring pressures. Pressure sensors usually measure displacement and generate signals that indicate an applied pressure. These displacement signals can be collected visually, such as with a barometer or a gauge, or electrically. Converting this physical displacement to an electric signal is known as transduction. Transduction is useful for integrating pressure sensing with electronic devices[21].

In the past, pressure sensing was mostly reserved for civil and mechanical engineering applications. Pressure sensors are used in aviation for altitude sensing and in marine industry for measuring depth. They are also used in automobiles for sensing the accelerator, brakes and air bags as well as regulating fluids and engine power. More recently, pressure sensors have been integrated into phones and computer devices to provide touch screen capabilities[22]. The pressure sensors used for the previous applications are rigid and planar. In order to make pressure sensors appropriate for measuring human physiological pressures, they must be conformal, flexible and stretchable[23-25].

The sphygmomanometer, still the gold standard for measuring blood pressure, consists of two sensing devices: a pressure gauge and a vibrational sensor. The original sphygmomanometer used a column of mercury to gauge pressure. Current day devices use an aneroid, or barometric, gauge. In order to identify the systolic and diastolic pressures, doctors use a stethoscope to listen for Korotkoff sounds. These Korotkoff sounds are caused by the turbulent blood flow that causes vibrational sounds. These vibrations can also be detected using electric sensors, which measure oscillations. In all cases, the sphygmomanometer only is able to measure the peak pressures (systolic and diastolic). In order to obtain a complete pulse profile, pressure sensors must be applied directly over the artery with minimal force[6].

Arterial tonometry is the method of applying strain sensors directly against the skin to measure the blood pressure. The advantage of this technique is a complete pulse profile and continuous measurement. This technique can give greater data for better clinical analytics. However, arterial tonometry has been slow to adoption in the clinical environment, mostly due to unawareness and lack of studies to support the accuracy of its sensors[13]. The follow section gives an overview of how electrical pressure sensors are characterized and how to determine which sensor would be optimal for use in arterial tonometry and wearable devices.

2.2 Characterization of Sensor Performance

Sensors that function through electrical transduction have key parameters that summarize their performance. These include linearity, sensitivity, limit of detection, response time, stability, and operating voltage[4].

Linearity is the percent deviation from a direct correlation between applied pressure and the corresponding electrical signal. Sensors that perform linearly are generally more reliable and predictable since pressure and signal are more or less proportional[4]. Linearity often an indicator of the optimal pressure ranges for a particular sensor[26]. While pressure sensors may have the ability to detect wide ranges of pressures, focusing on the linear regions helps to identify that sensor's pressure regime. Typically these pressure regimes are measured in Pascals. Sensors in the medium pressure regime (10-100kPa) are used in current blood pressure devices while sensors in the low-pressure regime (<10kPa) have shown promise in measuring pulse pressure without the cuff[27].

Sensitivity is arguably one of the most important parameters to take into consideration when evaluating sensor performance[26]. Sensitivity can be defined as:

$$S = \frac{dX}{dP}$$

Where S is sensitivity, dX is the change in output signal, and dP is the change in pressure. A higher sensitivity is desirable for more responsiveness and accuracy. Therefore a big change in the electrical output (numerator) will result in a more discernable change or sensitivity for a given applied pressure.

Limit of detection (LOD) is another useful parameter to characterize a sensor, particularly when working in the low-pressure regime, where a low limit of detection is desirable[27]. LOD, also known as the sensor's threshold, is commonly defined as the lowest pressure that produces a detectible electrical signal, with a noise to signal ratio of 3 to 1.

Response time refers to the time delay between applied pressure input and electrical signal output. Usually a low response time is desired, especially in real time physiological sensing applications, where signal timing is an important parameter in dynamic signals. Typical quick response times for real time health monitoring are considered to be <100ms[28]. Response time is usually captured through cycling, or periodic change in pressure by a stepper motor. Cycling can also give insight to stability, or a sensor's ability to maintain its signal production after repeated use. Good stability is essential for sensor application in biomedical devices[29].

Operating voltage is an important parameter to consider for sensor integration into wearable devices. Operating voltage is the voltage required to operate a sensor and is a determinant of power consumption. Since wearable electronics rely on small electronic components and small batteries, a low operating voltage is desired to minimize battery consumption.

2.3 Mechanisms of Transduction

Pressure sensors act as transducers, describing the process in which applied pressure produces an electrical signal. There are various methods of transduction, but piezoresistive sensors and capacitive sensors in particular have shown promising sensing properties for wearable applications[4].

Piezoresistive sensors respond to pressure by changing in electrical resistance. Electrical resistance is a measure of how a device reduces the flow of current. This response is usually generated by a change in contact resistance between two functional materials, created by the mechanical deflection caused by applied pressure. These resistance changes can be measured directly with a digital multimeter, or using Ohm's law:

$$V = iR$$

Where R is proportional to both variables voltage (V) or current (i), and can be indirectly measured by keeping one of these variables constant. Piezoresistive sensors are known for their simple design mechanisms, high sensitivities, fast response time, and low operating voltages – making these sensors great candidates for wearable device integration[31].

Capacitance sensors act through capacitance, or the ability to store and electrical charge. Most common are parallel plate capacitors, in which a dielectric material separates two functional materials. Applied pressured causes a deflection of of this dielectric area resulting in a change in capacitance. This capacitance, C, can be calculated with:

$$C = \frac{k\varepsilon_0 A}{d}$$

Where k is the dielectric constant, or the dielectric material's relative permeability, ε_0 is the electric permittivity of space, A is the area overlap between two plates (in square meters), and d is the distance between to plates (in meters). These sensors tend to exhibit a low change in capacitance, thus have relatively low sensitivities. However, organic thin film transistors (OTFTs) also use a capacitance mechanism but are able to achieve higher sensitivities due to their unique amplification function[15]. OTFTs require many more components than piezoresistive sensors, including a semi-conducting layer, gate dielectric layer, gate electrode and source drain electrodes. A compilation of literature values characterizing current flexible pressure sensors can be found in Table 2.1.

Type of Sensor	Materials	Sensitivity	LOD	Response Time	Working Voltage	Reference
Capacitive/OTFT	PDMS/Pil2TSi	8.4 kPa^{-1}	-	<10ms	100V	15
Capacitive/OTFT	PDMS/Ruberene	0.55kPa^{-1}	3 Pa	<10ms	80V	32
Capacitive	PDMS/Nanotube	0.23kPa^{-1}	50kPa	<125ms	-	19
Piezoresistive	PDMS/SWNTs	1.8 kPa^{-1}	0.6 Pa	<10ms	2V	16
Piezoresistive	Tissue/ Gold nanowires	1.14 kPa^{-1}	13 Pa	<17ms	1.5V	31

Table 2.1: Summary of flexible pressure sensors and their performance.

While piezoresistance and capacitance/OTFT have both proven successful in achieving high sensitivities and response times for pressure sensing flexible devices, we chose to move forward with piezoresistance sensors, due to simple design criteria and low operating voltage.

The next chapter will describe both the fabrication methods of these piezoresistive sensors as well as the role of topography as a sensing mechanism.

CHAPTER 3

FABRICATION OF WRINKLED THIN FILM SENSORS

3.1 Substrate and Functional Materials

In order to develop sensors with superior sensing capabilities, it is important to choose suitable materials. Unlike OTFTs, which are composed of many materials and layers, piezoresistive sensors are simpler in design and only require two main components: a substrate and a functional material. The combination of these two materials will affect the mechanical properties and the electrical conductivity of the sensor[34].

Traditional pressure sensors are largely non-conformal because their substrates are rigid and planar. The development of conformal sensors requires a highly flexible, soft substrate[24]. One of the most common elastomers used as a flexible substrate is polydimethylsiloxane (PDMS). This commercially available rubber silicone is biocompatible and readily available in most laboratories[30]. It has high elasticity, with a Young's Modulus of 1.8MP. PDMS is also optically transparent and thermally stable[26]. In addition to these many advantages that PDMS offers, it can also be easily molded to fit various geometric configurations, which is one of our main criteria since modifying structural configurations will be instrumental to the design of our sensors.

Functional materials largely contribute to the electrical and mechanical properties of sensors. Initially, we chose gold as the functional material and electrical conductor for our piezoresisitve sensor. Early on we discovered that gold did not possess the electrical properties required for high sensitivity. We replaced gold with single-walled carbon nanotubes (CNTs), which have an electrical current carrying capacity one thousand fold greater than gold[35]. CNTs are cylindrical carbon molecules that naturally align into rope-like anisotrophic networks[36]. In addition to their exceptional electrical properties, CNTs exhibit superior mechanical properties, with a Young's modulus higher than steel. Due to the nature of their alignment, CNTs are also flexible and compatible with elastomer substrates[37].

3.2 Wrinkled Structural Configuration

Recently it has been shown that introducing microstructures into the PDMS film of piezoresisitve sensors results in higher sensitivities and response times than unstructured PDMS[11, 13]. Unstructured, planar PDMS films are subject to amplified visco-elastic behavior, such as decreased response times and inhibited deformation. High-density microstructures have proven useful in combatting these effects by providing features that will elastically deform, which improves the mechanical response to applied pressure. Work by Schwartz et al.[11] reported that pyramidal PDMS structures gave a 30-fold increase in sensitivity compared to unstructured film and improved response time more than tenfold. However, fabrication of uniformly patterned PDMS microstructures traditionally expensive fabrication processes, namely photolithography.

Shrink film technologies, originally utilized by Fu et al., offers a much simpler and cost efficient alternative technique for creating microstructures[38]. This platform utilizes pre-stressed thermoplastic, known as commodity shrink film, or polystyrene (PS). Upon exposure to heat, PS shrinks down to its original size. When a thin metal film is deposited onto the surface of PS, thermal shrinking creates a buckling effect in the metal film, creating hierarchical, self-similar wrinkled structures. These wrinkles are tunable, with a thicker metal film resulting in wrinkles of higher amplitude[39]. In addition, wrinkled features can be directly molded into flexible polymer substrates, including polydimethylsiloxane (PDMS). Importantly, these wrinkled polymer substrates are roll-to-roll compatible so the fabrication is scalable for manufacturing[40].

Increased surface area is a particular advantage of these thin film wrinkled surfaces. The high surface area and high-resolution structures help to increase conductivity, and thus contributes to a better sensitivity and lower limit of detection for electrodes[41]. In this work, we show that patterned wrinkled structures can act as contact sites for the conducting networks of our piezoresisitive sensors. The following section provides the fabrication method for creating wrinkled structures and assembling piezoresisitve sensors.

3.3 Fabrication Methods

To create wrinkled topographies, commodity shrink film is coated with a thin layer of gold (~40nm thick) via sputter deposition. This substrate is then placed in an oven at 160 degrees Celsius, where it undergoes shrinking. The surface mismatch between the soft plastic and the brittle gold film causes a buckling effect on the surface. This buckling creates multi-scale wrinkles. The amplitude of these wrinkles is tunable by varying the thickness of the gold film. Optimization of this step resulted in the use of 40nm thick gold films for the formation of wrinkles that are have an amplitude ~10um and a peak height ~3um. The peaks of these wrinkled structures provide effective contact sites for our piezoresistive pressure sensors.

Wrinkled structures are then molded into a PDMS elastomer. PDMS is prepared by mixing the elastomer base and the curing agent in a 10:1 ratio. PDMS is then spin coated directly onto the wrinkled substrate at 150RPM for 30 seconds, yielding a PDMS thickness of ~300um. The entire substrate is then placed in the vacuum for 15 minutes and then placed in the oven to cure overnight. After the PDMS is cured, it is carefully pealed off of the wrinkled gold/PS substrate. Next, commercially available CNT solution is deposited directly on top of the PDMS wrinkled surface using an air-spray gun. The entire fabrication process is outlined in Figure 2.1.

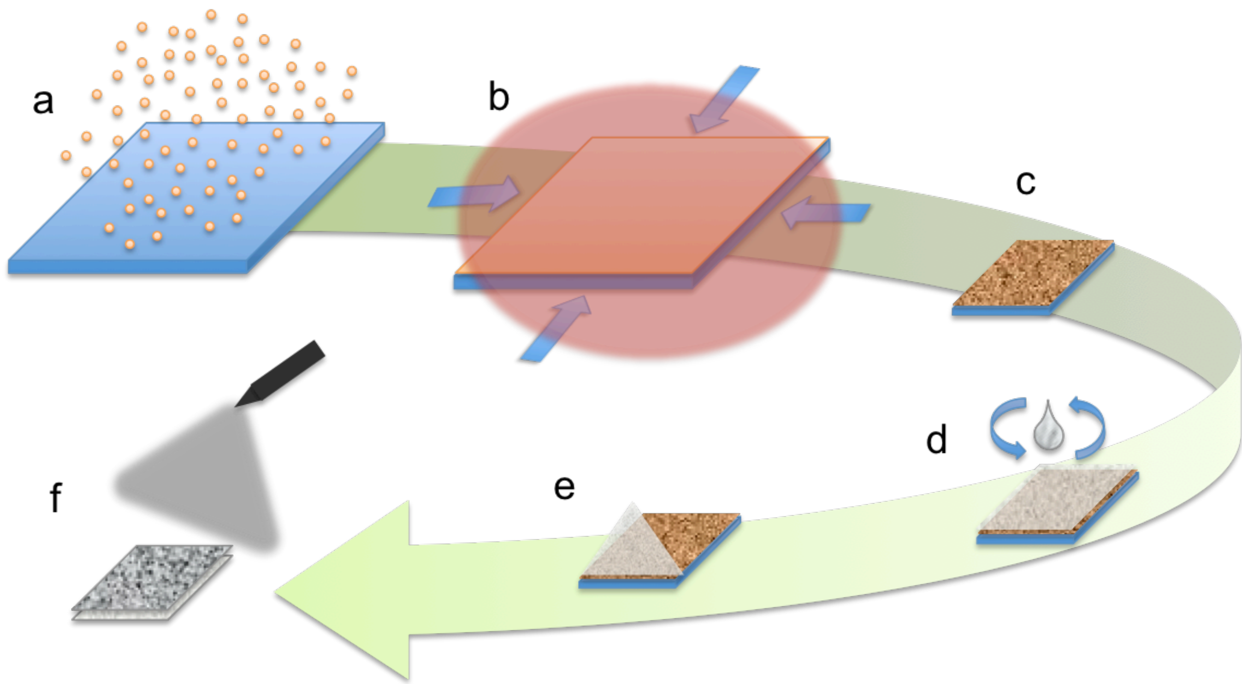

Figure 2.1: Process Flow Diagram. a.) Gold thin film deposited onto pre-stressed polystyrene b.) Substrate is heated c.) Substrate shrinks and forms wrinkled topography d.) PDMS casted onto wrinkled substrate e.) Peal PDMS f.) Spray coat PDMS with CNT solution

The wrinkled features unexpectedly promoted CNT adhesion to the PDMS surface, opposed to the non-wrinkled PDMS surface, which showed poor adhesion as the CNT films easily detached from the surface. To further improve CNT adhesion and stability, the CNT/PDMS thin film was annealed at 200 degrees Celsius for 30 minutes. Thermal treatment also improves the conductivity by enhancing the junctions between CNTs. The resulting CNT layer is tightly adhered to the wrinkled topography, as seen in the scanning electron microscope image (Figure 2.2).

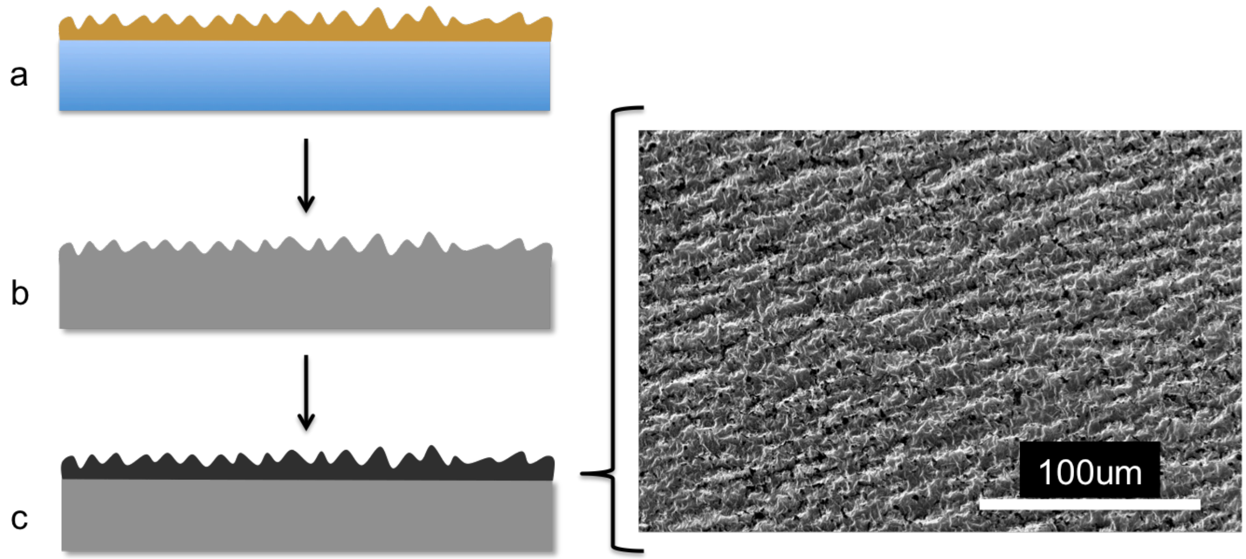

Figure 2.2: Schematic of wrinkled structures. a.) Wrinkled gold film formed by heat shrinking PS b.) Molded wrinkles in PDMS c.) CNTs deposited onto wrinkled PDMS surface accompanied by SEM images of wrinkles.

Piezoresistive, or resistance-based, pressure sensors are assembled by taking two CNT/PDMS wrinkled thin films (20mm x 20mm) and placing them together, so that the two wrinkled CNT faces are in contact. Ag paste was used to attach thin copper wires along opposing ends of the films to create the source and drain electrodes. The initial contact points between the two wrinkled CNT faces are the points at which electrical current passes through the sensor. The resistance of the sensor is the combined value of all of these contact points. As pressure is applied to the sensor, the wrinkled surfaces compress against one another, creating more points of contact for electrical current to pass through. This decreases the overall electrical resistance of the device, with an increase in pressure. Therefore for this piezoresistive sensor platform, a decrease in resistance is correlated with an increase in pressure. A schematic, along with a photograph of the piezoresisitve sensor device is shown in Figure 2.3.

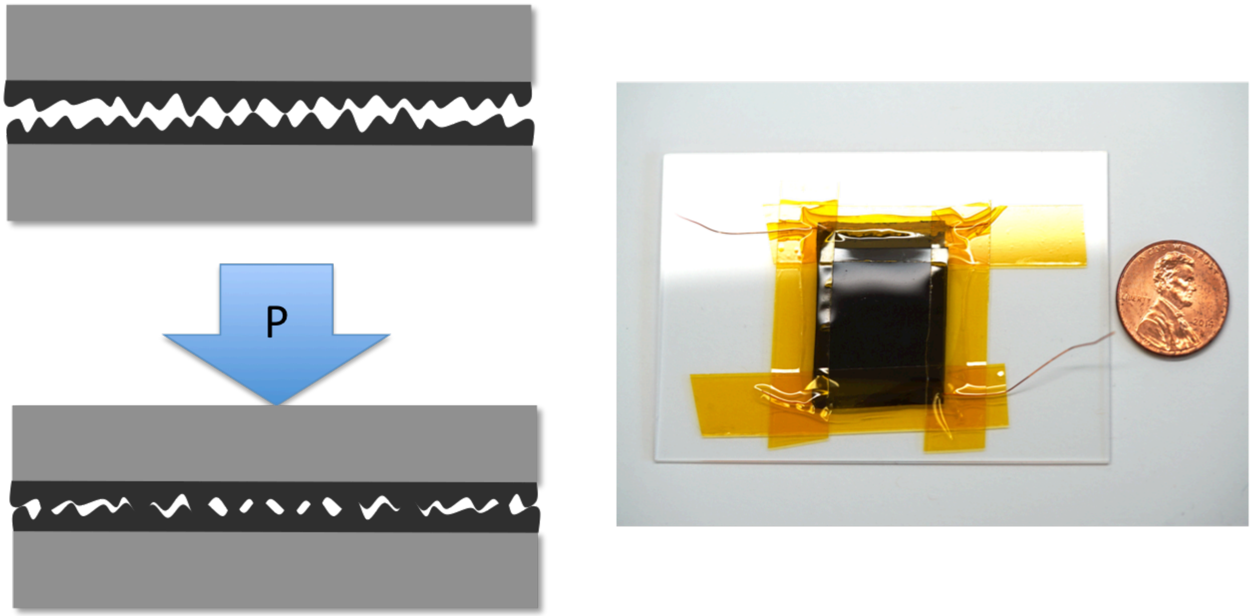

Figure 2.3: Schematic (left) and photograph (right) of piezoresistive sensor.

CHAPTER 4

SENSOR CHARACTERIZATION

4.1 Linearity and Sensitivity

Piezoresisitve pressure sensors translate pressure into an electrical change in resistance. This resistance can be measured directly, or indirectly by measuring voltage or current (while keeping the other constant). For our characterizations, we measured resistance directly using an LCR meter, an instrument named for its capability of measuring inductance (L), capacitance (C), and resistance (R). Input voltage, although not constant, was set at 1V. An automatic force gauge was used to measure applied pressure. To determine the dynamic range and sensitivity, pressure was applied in a stepwise fashion. As applied pressure increased from 200Pa to 1800Pa, there is a consistent decrease in resistance. In order to better visualize this relationship, these resistance measurements are reported as resistance change ($\Delta R/R_0$) and then the inverse $(\Delta R/R_0)^{-1}$ is plotted against pressure in Figure 4.1. This curve shows two linear regions (S_1 and S_2). The sensitivity between 200 and 300Pa was found to be 0.65kPa^{-1}, which is within the reported literature sensitivities found in Table 2.1.

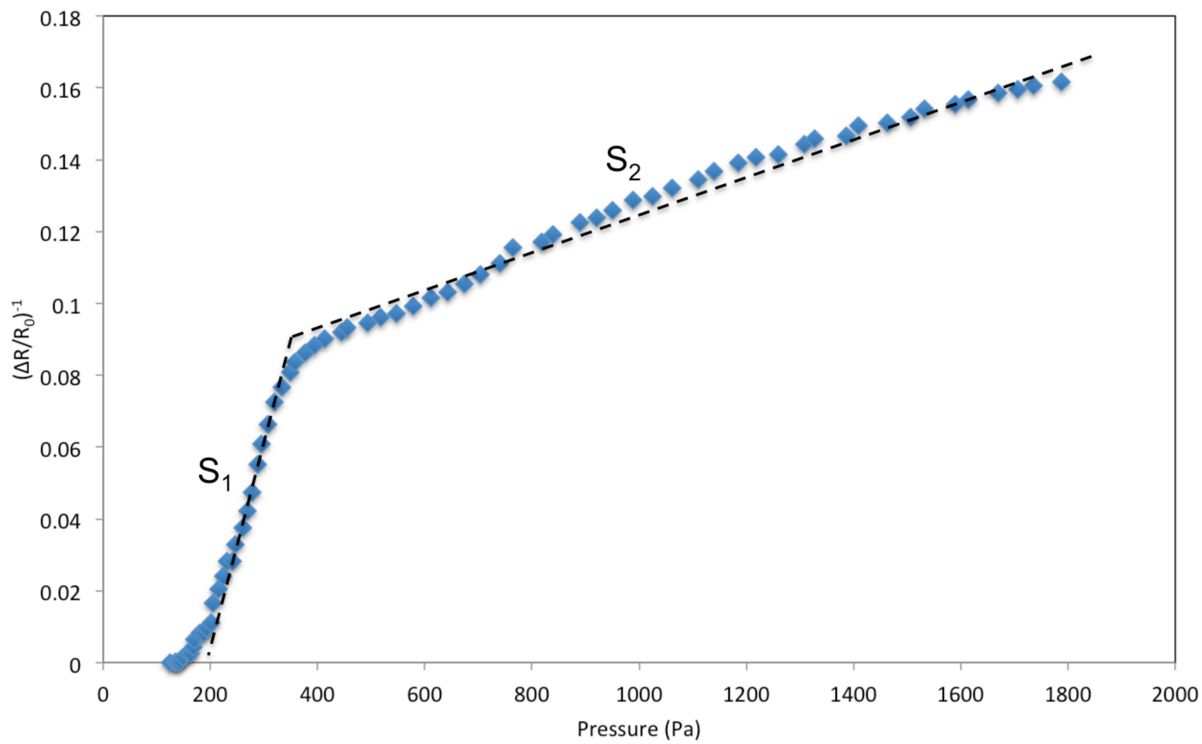

Figure 4.1: Sensitivity of wrinkled piezoresisitive pressure sensor.

4.2 Cycling and Response Time

The force gauge setups and LCR meter were also used to measure cycling. In order to assess the stability of the pressure sensor, the resistance change for repeated loading and unloading of pressure (600Pa) was measured for 10 cycles (Figure 4.2). Although the pressure sensor exhibits good repeatability and stability, there is slight hysteresis that occurs between loading and unloading. This could be due to inadvertent resistance changes produced by deformation of the electrodes or possibly the elastic nature of PDMS, both which can be corrected for with further sensor design optimization. The response time is <10ms, according to the sampling rate of the LCR meter. This is consistent with literature values for response time (Table 2.1).

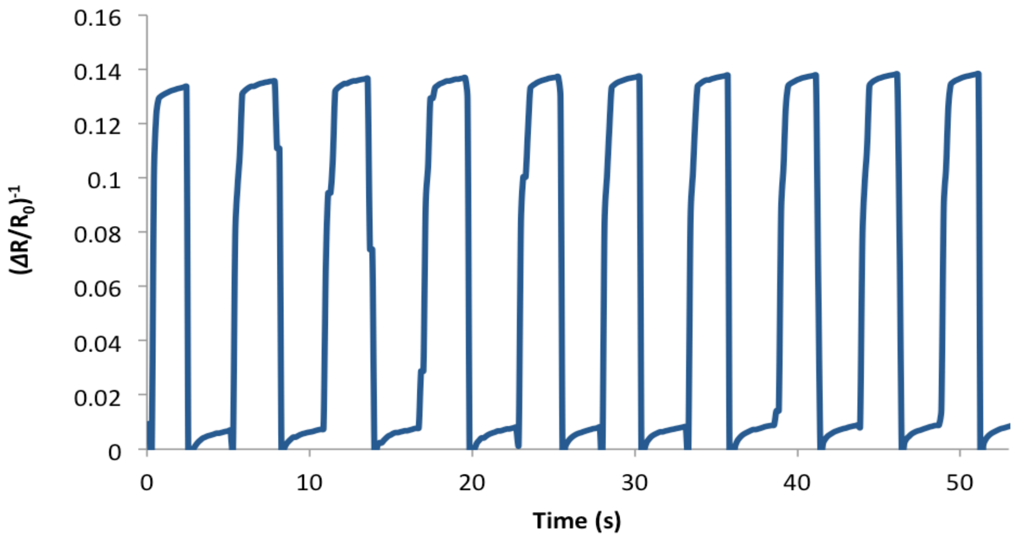

Figure 4.2: Real-time curves for loading/unloading cycles.

4.3 Pulse Pressure Detection

The wrinkled thin film piezoresistive sensors demonstrated appropriate performance in the force gauge tests for physiological monitoring. To access the capability of these sensors to measure pulse pressure, sensors were placed on the carotid artery in the neck (Figure 4.3) and the radial artery in the wrist (Figure 4.4). The resistance change signal at the carotid artery provided a dynamic display of pressure change. Although there was slight drift in the overall signal, the repeated pulse signals were similar in amplitude. The resistance change signal in the radial artery in the wrist was slightly noisier, with a more noticeable drift, most likely due to the relatively low pressures changes in the radial artery as compared to the carotid. This signal contains consistent pulse waves, however the wrist pulse contours, comprised of the percussion wave (P-wave), tidal wave (T-wave) and diastolic wave (D-wave), cannot yet be distinguished.

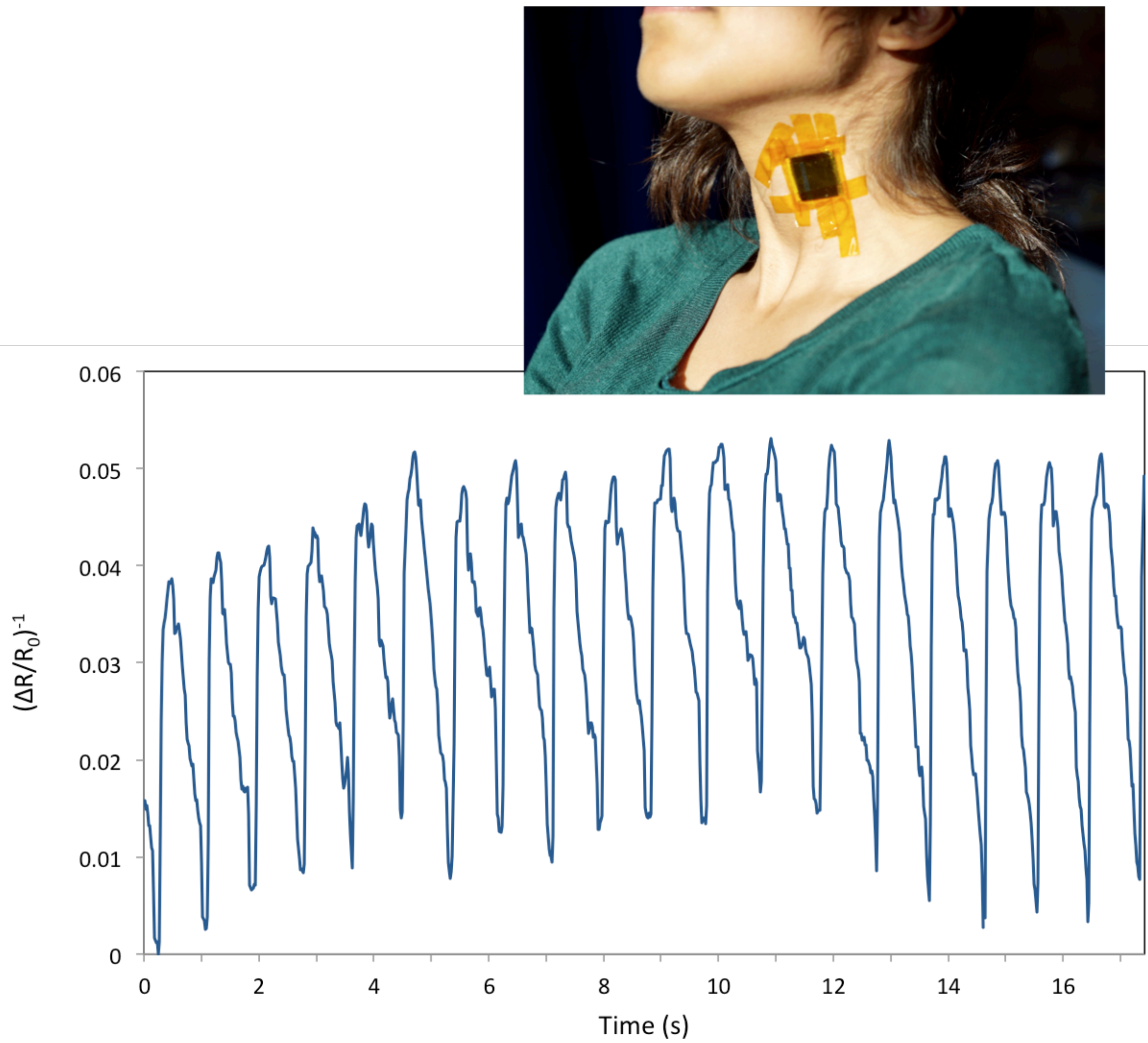

Figure 4.3: Real time curve for measuring pulse at neck.

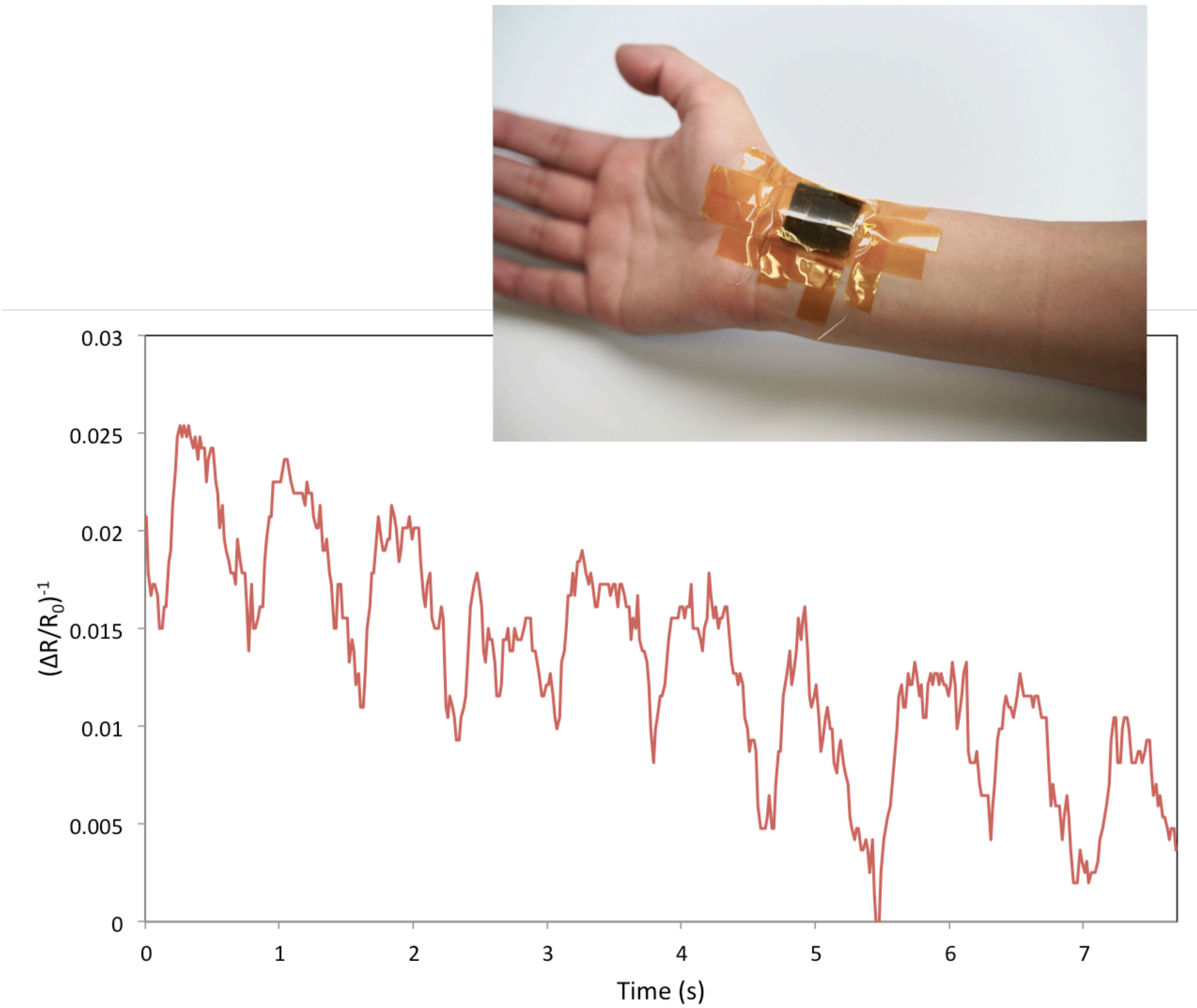

Figure 4.4: Real time curve for measuring pulse at wrist.

CHAPTER 5

Sensor Integration with Wearable Devices

5.1 Compatibility with Microcontroller

To further explore the integration of piezoresistive sensors with wearable health monitoring devices, piezoresistive sensors were combined with the other elements of wearable electronics: an electrical circuit and microcontroller. The microcontroller (Arduino 101) has an operating voltage of 3.3V, so it is relatively low power consumption and comparable to the operating voltage of the sensors during testing (~1V). The microcontroller supplies the power and acquires a signal. The acquired signal is an arbitrary value between 0 and 1024. In order to optimize the input voltage to output an appropriate signal range, a voltage divider was implemented. A photograph of the microcontroller, voltage divider and sensor setup is shown in Figure 5.1. Although this setup was capable of detecting pulse at the radial artery in the wrist, resolution was poor. This is due to the limited sampling rate of the microcontroller.

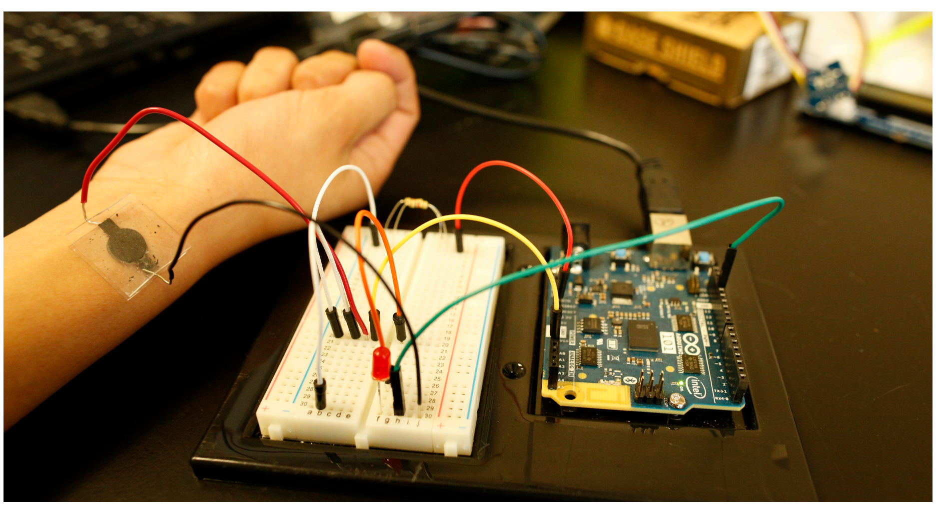

Figure 5.1: Integration with a Microcontroller

5.2 Wearable Vitals Monitor

Advancements in wearable technologies are largely due to the miniaturization of the electronic components. Shrinking electronics allows for small and lightweight integration into wearable items, such as clothing and wristbands. After successful integration of conformal pressure sensors with electronic components, miniaturized electronic components were implemented to fit in a wearable wristband. Figure 5.2 shows the original microcontroller (Arduino101) and electronics compared to the miniaturized microcontroller (Radon, Intel Corp) with the electronics soldered directly onto the backside of the board. This platform was housed in a custom 3D-printed compartment and attached to a silicone wristband. Sensors were placed on the inside of the wristband, over the radial artery. This device showed similar performance to the original microcontroller platform. However, the pulse signal was easily overridden by noise, especially noise caused by human motion. In order to fully employ conformal pressure sensors into wearable devices, filters must be applied and algorithms must be developed to extract pulse signal from noise.

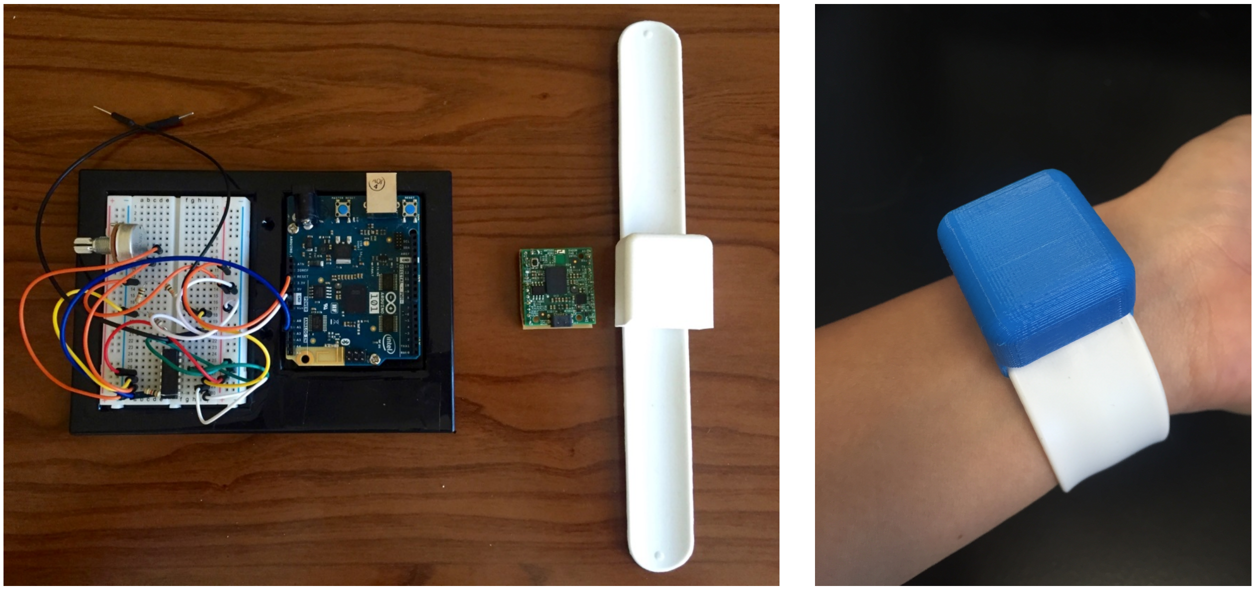

Figure 5.2: Reducing size of microcontroller and electronics (left) for integration into wearable band (right).

CHAPTER 6

Summary and Conclusions

The recent popularity of wearable devices gives opportunity for health monitoring applications. Current wearable devices are unable to measure important health vitals, such as blood pressure. A device that measures blood pressure continuously could give greater clinical insight to both health professionals and patients. However, the current gold standard cuff device, the sphygmomanometer, is not suitable for continuously monitoring. This work summarizes the rapid progress that has been made in the development of highly sensitive conformal pressure sensors. In addition, this work offers a scalable, low cost alternative for creating these sensors with microstructures, for better sensor performance. These piezoresistive sensors not only demonstrated good sensitivity with fast response times, but also operate at a relatively low voltage. This low operating voltage allowed for the integration of these sensors with a microcontroller, and even a complete wearable platform. These sensors still need optimization to reduce hysteresis, as well as advanced algorithms to combat the signals susceptibility to noise. Most importantly, this work shows that production of microstructured flexible pressure sensors is scalable and can be implemented for realistic applications.

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